PROCEEDINGS 1980

INTERNATIONAL CONFERENCE ON REHABILITATION ENGINEERING

JUNE 16-20, 1980
TORONTO, ONTARIO, CANADA

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FOREWORD

This proceedings of scientific papers relates to the scientific sessions which constitute a vital programmatic component of the International Conference on Rehabilitation Engineering. ICRE-80 also embraces workshops on issues of current concern, and instructional courses relevant to the provision of fundamental knowledge in a variety of important sectors of rehabilitation endeavour. The Conference includes an exhibition of both scientific and commercial efforts relating to the field, and as well, student design and paper competitions.

The organizers of each of the programmatic components are to be lauded for their outstanding efforts in reaching for and attaining the levels of excellence that have resulted from their energies. A perusal of this proceedings will readily reflect the superb efforts of Dudley Childress and Bob Scott who have ensured the high quality of scientific presentations through their careful and systematic organization. Each paper submitted was reviewed by at least one of them and has been submitted for critical review to two external reviewers in the field. A broad range of rehabilitation engineering topics has been collected together to constitute three parallel scientific sessions at the conference. The international flavour is quite evident and the level of maturity of rehabilitation engineering endeavour marked.

Feeling certain that this proceedings will be a valuable reference work for a number of years to come, we wish all participants in the ICRE-80 "Bonne conférence".

Douglas A. Hobson
Program Co-Chairman
Memphis, Tenn. USA

Morris Milner
Program Co-Chairman
Toronto, Ont. Canada

EDITORS' NOTE

The Proceedings of the International Conference on Rehabilitation Engineering 1980 are a continuation of publications started by the Conference on Systems and Devices for the Disabled. Those programs, which began in Boston in 1974 as an outgrowth of the Carnahan conferences, published proceedings that have provided a valuable stimulant for technological development in rehabilitation. We hope these proceedings will continue that tradition.

We want the papers to foster ferment within the field and believe the Proceedings should not only be a place where logical concepts are extended but also a place where ideas that are daring and imaginative can be presented. Some of these ideas may not be completely worked out and others may not even be completely correct. Nevertheless, as Edward de Bono has observed in *New Think*:

> "The need to be right at every stage and all the time is probably the biggest bar there is to new ideas."

In this sense, we hope these papers will encourage ideas and that the Conference will be an effective forum for lateral thinking, for constructive criticism, and for learning.

We wish to thank the Canadian Medical and Biological Engineering Society for publishing these Proceedings. This is a significant contribution toward furthering the emergence of rehabilitation engineering. Thanks are due of course to those who contributed papers, and to those who assisted in the review process. Finally, we wish to record particular appreciation to Bonnie Collard and Dianne Hughes, our secretaries, for many hours of extra work associated with these Proceedings.

We hope these Proceedings and the Scientific Program will be of value to the rehabilitation community, and most of all that they will help enable disabled people to further realize their human potential.

D.S. Childress

R.N. Scott

The Conference organizers, publishers, and Scientific Papers Committee are not responsible for statements or opinions advanced by papers in these Proceedings, or for errors therein.

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TOWARD A BRAILLE COMMUNICATION AND INFORMATION SYSTEM

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Palo Alto, CA 94304

ABSTRACT

Braille code is central to effective communication and information handling by blind people. With the advent of dynamic braille displays, storage of braille on magnetic media, and portable packaging, tasks previously known to be onerous if not impossible for blind individuals can be accomplished with efficiency and speed. As significant as these improvements are, braille remains a closed system which requires that written communication be processed in some fashion by a sighted person.

Technologies now exist which make a low cost general purpose braille communication system feasible. The recent proliferation of the microprocessor and associated software tools along with non-volatile mass storage media have reduced the cost of the system components to the point where they can be incorporated into an affordable consumer product. New technologies have changed the ways in which public information is stored, retrieved, and handled by sighted persons. Future braille communication systems will be able to adapt easily to public information networks, providing fast and convenient data processing. Although braille will remain a system used only by the blind, the existing communication gap between the sighted and the blind world will be eliminated.

BRAILLE -- A COMMUNICATION SYSTEM FOR THE BLIND

Developed during the last century, braille today remains the primary mechanism for reading and writing for the blind. The slate and the stylus or mechanical braille writer are, for the blind, what pen or pencil are for the sighted. Personal notes and day to day writing are done using braille. Braille is used on the job by most blind employees where writing, reading, and any information handling is required.

Braille is based on a code involving a six-dot cell arranged in a 2 x 3 matrix. Various combinations of dots are used to represent different alphanumeric characters, symbols, and punctuation marks. In 1959, standards for English-American braille were compiled and adopted. Foreign countries have adopted similar standards for braille in other languages.

English-American braille exists in three basic forms. These forms are referred to as "grades". Grade I braille is essentially a symbol for symbol form, where each print letter, number, symbol, or punctuation mark corresponds to a particular braille cell. Grade II braille employs approximately 200 "contractions" or abbreviations for various words and letter combinations. Grade II braille is by far the most common form of braille. All textbooks are produced in Grade II braille. The Library of Congress produces books and periodicals in Grade II braille for a population of over 20,000 active braille readers. Grade III uses a larger number of contractions than Grade II, and is thus a more "compact" form of braille. Use of Grade III is not widespread, however, being popular only among a small percentage of the most proficient braille readers.

The production of braille material is essentially a two-step process. The first step involves translation from print text to braille code. The second step concerns the actual embossing process. The production cost of braille material is very high. The average title produced by a large braille printing house costs $44 per copy while the average cost of a hand copied title is $80 per copy. Because of the great expense involved, the limited centralized production facilities and equipment, and the need for specialized transcriber skills, less than 5% of ink print materials is made available in braille.

NEW TECHNOLOGIES FOR BRAILLE

During the past decade, technologies for "dynamic" or "volatile" braille have been developed. The dynamic braille display consists of a number of braille cells with movable dots driven by electromechanical actuators. The individual dots can be raised or lowered to represent different characters, and braille information can be read from a one-line dynamic braille display by presenting each line of braille characters in succession. Original work in the development of a piezoelectric braille display was conducted in France. Work has been carried out in Germany, England, and the United States on solenoid driven displays. The VersaBraille™ system developed by Telesensory Systems, Inc., and the Digicassette produced by Elinfa both employ a 20-character dynamic braille display using piezoelectric actuators.

Accompanying the development of the dynamic display is the new capability of using magnetic storage media, such as cassette tapes, to store

braille information. Standard cassettes can now be used to drive dynamic braille displays. This combination of dynamic display and magnetic storage medium has commonly been referred to as "paperless braille".

A shift from embossed paper braille to magnetically stored dynamic braille brings about a dramatic decrease in the cost of reproducing braille. For example, a press braille title which currently has an average cost of $44 per copy can be reproduced on cassette for less than $9 per copy. While cost reduction is a major advantage of this new technology, it is not the only one.

Perhaps the most important contribution of dynamic braille technology is that it increases the ease and efficiency of handling braille information for blind people and thereby expands the usefulness of braille as a form of information and communication. Braille is transformed from a static, medium constrained, often cumbersome system to one which is dynamic and flexible. Electronic storage, retrieval, and editing of braille information now becomes possible, thereby making available to blind people the kind of facilities that computer-based word processing brings to sighted people. For example, TSI's VersaBraille system enables the blind user to write and store up to 400 pages of braille on a standard C-60 cassette. Specific pages, paragraphs, and even words can be accessed automatically in a matter of seconds. Text can be deleted and inserted without rebrailling the entire text, the process normally involved when working with paper braille.

Other technologies not directly associated with braille can be easily adapted for such purposes. Floppy discs and disc drives are now popular and inexpensive. With such devices the user not only utilizes non-volatile mass storage media but also retrieves and stores information quickly. Along with floppy discs and low cost microprocessors, the variety of application programs for organizing files, word processing, language translation and control is increasing rapidly. The semiconductor industry emphasizes the fact that during this decade significant improvements in CMOS technology will be made. With CMOS components, electronic circuits can be designed to consume significantly less power and consequently increase the portability of the final product.

New technologies have already reshaped the entire industrial market. More users today rely on computer technologies for communication purposes. Publishing houses are adapting compositor tapes as a substitute for microfilm for the generation of magazines, books, newsletters, etc. Public services such as news, classified advertisements, and electronic mail are available and increasing in popularity.

THE BRAILLE COMMUNICATION GAP

As important as all the developed technologies discussed previously are, braille remains a closed system isolated from the mainstream of written communication. Dynamic braille and magnetic storage media, while reducing the cost of braille information, do not solve the problem of translation, which accounts for a significant portion of braille production costs. However, the fact that braille can now be directly written and read from a magnetic medium opens the possibility for an intermediate process to be introduced to make translation from written English to braille (or vice versa) an automatic process.

Some dynamic braille systems, such as the Digicassette and the VersaBraille, have the capability of translating print to braille, symbol for symbol. The VersaBraille system, for example, has a built-in Grade I braille translator which is bi-directional. As Figure 1 illustrates, print text can be entered using a standard keyboard or other device. The keyboard transmits the print text using ASCII code. The ASCII code enters the VersaBraille input-output (I/O) port and is translated into Grade I braille code which is displayed or stored in the VersaBraille. Likewise, Grade I braille code can be translated by the VersaBraille into ASCII code which can then be sent to an external device, such as a line printer, which will produce printed text. Here, then, for the first time, the translating process can be handled at the personal level, independent of a human transcriber.

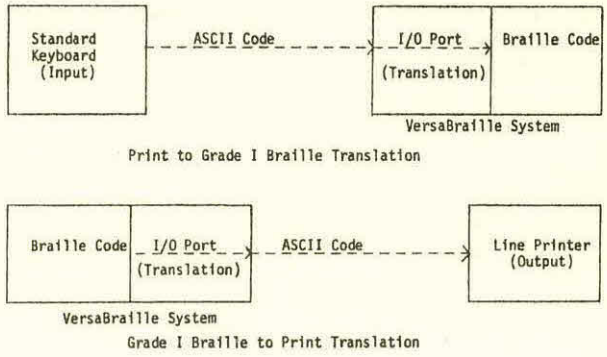

Figure 1. Print to Grade I Braille Translation

As previously mentioned, however, Grade II braille is the form most commonly used. The translation to and from Grade II braille is not a simple character-for-character substitution process as it is for Grade I. Use of the various contractions is based on a combination of grammatical and linguistic rules, context, and rules that have no grounding in linguistics but, which through use, have become standard convention. This complex transcription process has traditionally been handled by trained braille transcribers.

Since there are only 64 possible dot combinations with a 2 x 3 dot matrix, some combinations are used to represent a number of symbols. The contraction for the word "his" provides a good example of this problem. This three-dot contraction, when it stands alone, represents the word "his". When it is the first character of a character group, it represents opening double quotation marks. Preceded by a particular character, it represents an opening single quotation mark. Finally, taken in a numerical context, it represents the number "8". Depending on position and context, then, this single contraction has four different meanings.

A word like "sweetheart" provides a further example of some of the complexities of translation. The word contains three possible contractions. Contractions exist for the letter pairs "th", "ea", and "ar". The "th" is not used because the "t" and "h" belong to two different syllables. The "ea" contraction is not used because when "ea" and "ar" are both possible, the rule is that "ar" takes precedence. Thus, of three possible contractions, only one, "ar", is used in the correct Grade II translation of the word "sweetheart".

The availability of large computer memories has enabled a number of people to apply computer technology to the task of Grade II braille translation. The pioneering work in this area was conducted at MIT with the development of the DOTSYS program. More recent programs include that of Duxbury Systems, a modified DOTSYS program written in FORTRAN; ASI Teleprocessing's system, written in assembly language; and IBM's PL/I version. These are large, table-driven programs, requiring significant amounts of memory.

Existing Grade II translation software has been developed with the needs of large, centralized, press braille facilities in mind. Due to a lack of established performance criteria, little is generally known about the comparative quality of the existing programs. They are tailored or dedicated to specific hardware configurations which are, in turn, tied to press braille production systems. As a result, translation programs generally require large, expensive mainframe computers which are economically justified by institutional producers of braille but which are not affordable by smaller production facilities, such as schools, or by users of dynamic braille systems like the VersaBraille system or Digicassette. Furthermore, since the problem of translation has been approached with braille production as the ultimate end, the programs have been constructed only to perform translation in one direction -- from print to braille.

The fact that existing Grade II translation programs have been developed for, and are tied to, the process of press braille production also means that it has not been necessary to consider speed a vital element of the translation process. Consequently, there has been no need for a high speed translation program that translates as information is entered or received without noticeable delay. However, with the dynamic display, braille becomes an interactive code which can be used on-line to send and monitor information. Whether or not braille can be used in this way becomes dependent upon whether the Grade II/print translation process can be handled quickly enough.

Thus, in a number of ways, existing Grade II braille translation software is not suitable for use with dynamic braille systems and is therefore deficient in helping to make braille a much more interactive and powerful form of communication for the blind. That existing software is generally written for and depends on the capabilities of large dedicated computer systems renders it unadaptable for small, personally affordable braille systems. The high cost of the associated hardware also limits use of existing translation software to large institutions where the cost of translation is ancillary to the cost of braille production. Furthermore, the uni-directional nature of this software fails to meet the requirements that the translation process operate in both directions -- from braille to print as well as from print to braille.

It becomes apparent, then, that it is no longer the cost of braille production but the cost of braille translation which must be reduced before braille can become a truly interactive communication medium and before the effective handling of braille information is economically feasible at a personal level.

BRIDGING THE COMMUNICATION GAP

A proposed Braille Communication System, when implemented, should provide the means through which the communication gap can be bridged. It is important to examine the entire problem and to propose the general network before implementing the sub-system modules. This will insure some compatibility among products designed for braille users. Listed below are the requirements for the system modules which, when integrated, compose a basic Braille Communication System. The description of each module is left general enough to accommodate many implementations.

Braille Terminal

The braille terminal should include a dynamic braille display, braille keyboard, serial interface port, and should be portable. Interfacing to and from the terminal must be performed in ASCII, and the translation from ASCII to braille and vice versa should be done internally. An important element of the braille terminal should be the ability to be configured easily to a variety of data communication protocols.

Centralized Information Center

Given the relatively small number of braille readers, any dissemination program of braille material (either on paper or on cassettes) is extremely expensive. Braille books, magazines, or even a daily newspaper should not be mailed but rather disseminated through the telecommunication network. Requests for published material can be easily automated. Requests will be made by the user who utilizes his or her own braille terminal to scan the list of publications and to request the particular print copy (or portion of it) to be downloaded to a personal mass storage medium.

Similar inquiries can be made to a local or special information center for news and services. UPI and

other news organizations already provide services to owners of telecommunication equipment, and there is no reason why the blind user should be excluded from such an important service.

Non-Volatile Mass Storage Media

With the braille terminal, the need for a personal mass storage medium is naturally questioned. As described above, once the connection between the user and the central information center is completed, the braille material is accessed and can be read directly. Although technically speaking such interaction between the braille user and the information center is possible, it most likely will not be economically feasible. The reading process is relatively slow and complicated. Tasks like skimming through text, entering footnotes or reviewing, are all tailored to the specific user or to the type of text. Standardization of braille terminals for complicated reading control functions will be difficult and expensive. The connection time should be minimal in order to serve a large number of users simultaneously, and reduce system load. This can be achieved with a non-volatile mass storage medium (like cassettes or diskettes), not necessarily portable. The selected book or magazine can be downloaded at a very high speed, and terminated immediately. Two hundred pages of braille can be loaded within less than fifteen minutes. Special rates for long distance data communication are currently available and can reduce the braille dissemination process substantially.

Once the information is downloaded into the user's mass storage facility, it can be revised and edited at will.

Real-Time, Bi-directional Braille Translation

Translation from print (or ASCII) to braille is a necessity if any written communication between the blind world and the sighted is to be established. With the central information center and the remote data accessibility, braille can be easily disseminated. However, dependency on the manual transcribing process is a barrier to wider dissemination of information in braille (less than 5% of printed material).

An automated braille translation process must be part of the user's system. This will enable the user to connect not only to the braille information center but to any system or terminal not specially configured to communicate in braille. Such an automatic translation system will enable the user to shop for his own communication needs in a more general (and more competitive) market place. It is essential for the automatic translation process to be bi-directional and to operate in real time. If the above two objectives are met, the braille terminal will be identical to the conventional data entry machines and the blind user will be able to access standard computerized printed material (like compositor tapes).

Not essential, but highly desirable, is to translate printed material to and from Grade II braille format. Since Grade II braille is the most common form of braille today, it will be highly advantageous for the blind user to be able to enter data and read in the language which is normally used. Grade II braille can also be read faster and requires less storage capacity.

CONCLUSIONS

The evolution of braille technology in the past decade has been very dramatic. Braille users throughout the word can consider future braille writing and reading as a convenient, fast, and portable process. The remaining obstacles for braille users are two: the slow and expensive dissemination process and the need for a manual transcribing process performed by a sighted individual. A Braille Communication System, when implemented, will remove the obstacles, and the existing communication gap between braille and printed material will thus be eliminated. More development efforts by research and educational institutions, manufacturers, and national organizations should be directed toward the establishment of the system concept and toward the definition of communication standards.

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7. J. Snelders and H. Spanjersberg, "Braille From a Wordprocessing System," Dept. of Electrical Engineering, Delf University of Technology, The Netherlands.

BRAILLE PROOFREADER FOR BLIND TYPEWRITER OPERATORS*

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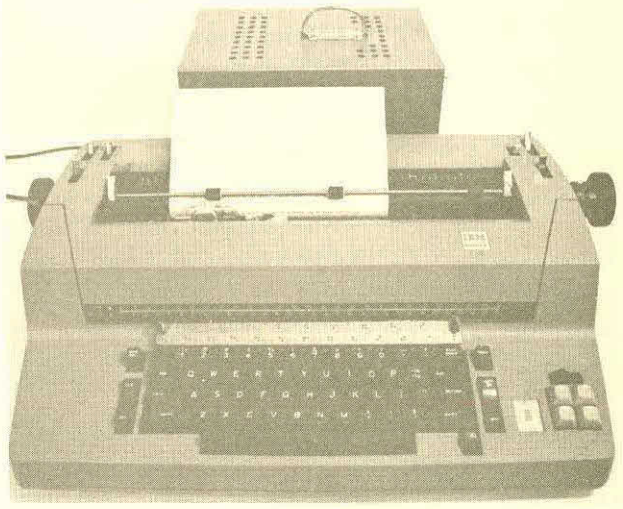
ABSTRACT

The Electronic Braille proofreader enables a blind person using conventional electric typewriters to type and store a page of typed materials in an electronic memory and display them in modified Braille code for random or sequential proofreading and correction without assistance from the sighted.

The device, if perfected, can greatly enhance the blind person's independence, self-confidence, personal privacy, and open more job opportunities for the blind.



A. INTRODUCTION

Blind persons using conventional typewriters key punches, etc., have great difficulties in proofreading their typed materials, printouts, etc., unless they get help from the sighted. To overcome this obstacle, Neou, etc., developed a Braille Verifier (1) in which mechanical coding bars were used to selectively energize the solenoids which raised the pins to form Braille cells on an endless belt driven by the carriage. As the Braille pin formations had to be erased to make room for the next line, the machine was only good for checking one line at a time.

The Braille proofreader described in this paper is a second generation Braille Verifier (2). It makes use of the latest electronic sensing and coding technology to encode and store an entire page of typed materials in an electronic memory which can be called up by the blind operator to show up in a Modified Braille Code at the Braille Readout for random or sequential checking and error correction. It can be designed to interface with any electric typewriter, and if perfected and mass produced, may cost about $300 in addition to the cost of the typewriter.

It should be remarked that the task of

* The work was sponsored by the Rehabilitation Service Administration, Office of Human Development, Department of Health, Education and Welfare under RSA Grant No. 14-P-57359/3.

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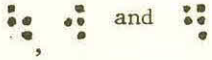
†† Now at Dynalectron Corp., Pueblo, CO.

proofreading of typed materials may as well be accomplished by using the recently developed Kurzweil Reading Machine (3) and the Optacon (4). They are, unfortunately, too expensive to be within the reach of the average blind person and his employer.

B. MODIFIED BRAILLE CODE

To avoid having double cell representations for numbers, capital letters, contractions, etc., in conventional Braille code, a Modified Braille Code is proposed and used in the experimental model. The new code has two more square-shaped dots atop of the six conventional Braille dots: the top left dot represents a small letter sign, the top right dot (slightly larger) is a capital letter sign, and the simultaneous appearance of both dots signifies a number. For example, the Braille cells for "d", "D", and "4" will appear respectively as

 and

An opinion survey of the blind revealed that they have no trouble in adapting themselves to the new simplified one-cell representation for numbers and capital letters.

C. SYSTEM DESIGN

The Braille proofreader consists of three subsystems. The following is a brief description of the organization and functions of their major components.

(I) Mechanical Movements to Electric Signal Encoders

The IBM Selectric II typewriter used in the experimental design has under its keyboard six interposers and one shift for rotating and tilting the typing head or element to position a desired character (letter, number, or punctuation, etc.) to face the paper. The seven bars are now employed as coding mechanisms to convert the motion of a typing bar into binary coded electric pulses. For example, when key "d" is depressed, the interposers on the coding bar 1,3,4 and 6 are moved to close their respective circuits and registers in the electronic memory as 1011010. Various switches and printed circuits are also installed in the typewriter to translate the motions of space bar, backspace and return keys, escapement, pitch selectors, etc., into electric signals to be processed.

The location or address (line and letter space positions) of a character on the paper can be sensed and converted into electronic pulses by means of photo-electric sensors and counters. The passage of a light beam to reach the two photo-transistors through a narrow opening in a disc turned by the platen produces an electric pulse to be added to or subtracted from, the line memory counter depending on the direction of turning of the platen. For example, the 44th

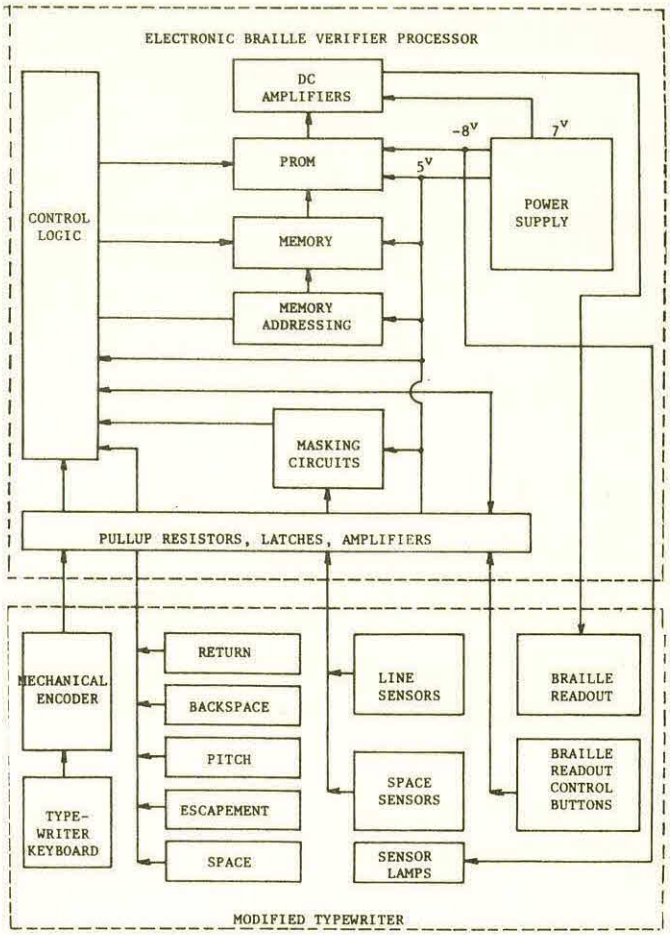


FIG. 2 SYSTEM BLOCK DIAGRAM

line on the paper will register as 101100 in the line address memory. Similarly, photo-electric sensors mounted on the carrier sense the number of passages of the light beam through the openings spaced one letter space apart in a bar fixed on the back of the typewriter. For example, the 23rd space in a line will register as 0010111 in its space memory counter. As each character stored in the memory has a unique line and a unique space addresses, any character can be recalled if its line and space positions are known.

(II) Central Processing Unit (CPU)

The CPU behind the typewriter in Fig. 1 receives the processes the seven bit signals from the encoder and converts them into eight-bit pulses in modified Braille Code to drive the display unit. The major components in CPU are (Fig. 2):

(1) Character Memory Boards. To store an entire page of typed materials in seven bit code (assuming 100 characters per line and 64 lines per page), its memory boards have 56 Intel 2012 1K x 16 RAM'S, providing a storage capacity of 44800 bits.

(2) Memory Address Counters. The line and letter space signals produced by the photoelectric sensors are amplified before they enter the two sets of binary counters (94 L 193): one for the six-bit line address, and the other for the seven-bit space address.

(3) Programmable Read Only Memory (PROM). An Intel 8702A PROM chip is programmed to convert the seven bit code in the memory into eight bit modified Braille code.

(4) D.C. Amplifier. The 8 bit signals from PROM are amplified before they go to drive the solenoids at the Braille readout.

(5) Control Logic. The control circuits make decisions on where and when the pulses should go and be processed: it channels the counting pulses to appropriate memory address counters, decodes the request from the readout controllers for selecting reading modes, commands the memory to output its stored contents and selectively energizes the solenoids, etc.

(6) Power Supplies. It converts the 110 VAC into DC on three voltage levels: a 5 VDC at 3 amp for TTL logic and memory, a -8 VDC at 200 ma for PROM and the light sources for photoelectric sensors, and a 7 VDC at 1.6 amp for driving the solenoids.

(III) Braille Readout and Readout Control.

The eight-pin Braille readout or display, four orange-colored readout control buttons, and a rocker switch are located at the right side of the keyboard. The Braille Readout has eight miniature solenoids (Electromechanisms SP-18 6 VDC). The depressing of one of the four Readout control buttons sends a message to the CPU and commands it to selectively output signals to drive the solenoids. The energized solenoids raise their respective steel pins (1/16 inch in diameter and spaced at 3/32 inch apart) 3/64 inch above the surface of the Braille Readout and form a pin formation in modified Braille code.

The four readout control buttons give the blind typist a choice of mode of reading: reading forward (F), backward (B), starting from the page top (T) and going to the next line at any point (L). By pushing one of the four buttons one at a time, the Braille pins fall and rise again to form new Braille cells one at a time at the operator's finger tip.

To inform the blind person of the whereabouts on the paper of a character to be checked, Braille displays are provided at the line and space counters. The mechanical line counter near the right platen knob turns with the platen and its unit and tenth counting wheels have 10 Braille markings on the surfaces and can be reset to 01 to coincide with the first line typed. The location of a character in a line on the paper can be found by reading the Braille numbered markings on a space counter opposite the position indicator of the pitch scale. The Braille space counter can be reset to read 01 at the left margin stop.

D. OPERATION

The blind typist can operate the typewriter in a conventional way. However, to use the Braille proofreading system, the following steps must be taken before typing.

(1) Return the carrier to the left margin stop and reset the line counting wheels to read 01 in Braille, corresponding to the first typed line on the paper.

(2) Set the space counter to start at the left margin stop.

(3) Depress the left side (for read-in) of the rocker switch to link the typewriter to the Braille Proofreader system.

(4) Press simultaneously all the four Readout Control buttons to erase the previously stored contents from the memories.

The operator can now proceed to type. If she/he wants to make a spot or random check on a certain character in the middle or at the end of typing, say the character occupying the 5th letter space in the 43rd line, she/he

(1) depresses the right side (readout or playback) of the rocker switch.

(2) turns the platen until the mechanical line counter reads the number 43 in Braille.

(3) taps the space bar until the "position indicator" sits opposite to the number 5 in Braille on the space counter.

(4) pushes the readout control button F (forward) to command the CPU to output the content in the memory and show up at the Braille Readout.

If the character is found to be a mistake, the operator can erase it in a conventional way and after depressing the left side (read-in) of the rocker switch, type in the correct character and after the rocker switch is put back in the readout mode, check it out on the Braille Readout. For proofreading the typed materials from the beginning of the page, she/he depresses the Braille Reader control button T (page top) and then buttons F or B successively.

E. CONCLUSIONS

The prototype of the Braille proofreader has successfully demonstrated its feasibility and usefulness as an aid to facilitate the proofreading and error-correcting of the typed materials by the blind users without the assistance of the sighted. The technology can readily be applied to many other keyboard operated machines for the benefit of the blind computer programmers, clerks, writers, etc. The corrected materials stored in the memory may also be used to drive an electric Braillewriter or a Braille composing machine to simplify and speed up the production and reproduction of Braille reading materials.

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A BRAILLE TSPS TELEPHONE OPERATOR'S CONSOLE

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Abstract

Microprocessor-controlled braille displays have been added to telephone operator consoles. These displays have enabled two blind operators to work successfully in a TSPS facility with a minimum of accommodation. The project is being extended by American Telephone and Telegraph and several operating companies.

INTRODUCTION

Braille displays have been installed on two Southwestern Bell telephone operator's consoles, providing vocational opportunities for blind individuals as TSPS telephone operators. A TSPS (Traffic Service Position System) operator handles most operator-assisted calls (pay phone, collect, hotel, credit card, etc.), but not directory assistance calls.

The TSPS console has approximately 80 lights or lighted keys (push button switches) and a 12 (soon to be upgraded to 14) digit numeric display for calling and called telephone numbers, time and charges for pay phone calls, and the time of day. The braille display makes all the necessary information immediately available to the blind operator. The information is grouped into a series of messages, either in English or accepted telephone industry mnemonics. The message displayed at any given time contains the information that most recently changed. The operator can override the automatic message selection and request a display of any information desired. An electronic "scratch pad" allows the operator to take notes and store any information which may be needed later.

Two blind operators in Little Rock, Arkansas, are using these braille displays, developed and built at the Massachusetts Institute of Technology. Both operators received rehabilitation training at the Arkansas Enterprises for the Blind and are now employed by Southwestern Bell Telephone Company (SWB). They received on-site training as TSPS operators from regular SWB training personnel.

The performance of one of these operators is rated as outstanding, while the other is rated as satisfactory. Very little extra accommodation at the working site has been necessary for either operator.

ORIGIN OF THE PROJECT

At the 1974 Conference on Engineering Devices in Rehabilitation, held in Boston, Massachusetts, a paper on "An Electromechanical Numeric Braille Display: A Familiar Tactile Representation of Electrically Encoded Digital Signals" was presented by an engineer from M.I.T.(1). It reported on BRLCOM(2), an electronic multimeter with braille output, which used a solenoid-driven braille cell as the output display. These cells were modular and used the standard braille spacing so that they could be stacked to form braille lines of any length. These cells had only four dots. Initially, it appeared to be simpler to make a four dot cell, since braille numbers use only the upper four dots of a six dot braille cell. The braille cells developed for use in BRLCOM were developed under a grant from the E. Matilda Ziegler Foundation.

An engineer from New England Telephone (NET) who also presented a paper at this conference(3) asked if this technology could be used to provide a braille display of numbers for blind TSPS operators. As a result of this discussion, an engineer from the Massachusetts Commission for the Blind and an engineer from M.I.T. visited a TSPS site in Quincy, Massachusetts on May 28, 1974(4). Further discussions over the summer led to an informal proposal and a request to NET by M.I.T. to embark on a joint project to equip a TSPS site for blind operators. Unfortunately, this request came at a time when NET had a surplus of operators and they were unable to justify such a project.

SELECTION OF SOUTHWESTERN BELL AS THE INITIAL INSTALLATION SITE

Subsequently, a training specialist at Arkansas Enterprises for the Blind (AEB) had interested Southwestern Bell Telephone Company (SWB) in a project to employ blind TSPS operators. When he received a commitment from SWB that a blind operator would be hired and trained, he began the construction of a simulated TSPS console. The various lamps would be sensed by a light probe. Since this training specialist was aware of BRLCOM, he felt it would solve the problem of providing TSPS numeric data to a blind TSPS

operator(5). At AEB's request, an M.I.T. engineer visited both AEB and SWB in September, 1975, to obtain technical information necessary to design the interface between the BRLCOM numeric display and the TSPS console and to develop a way to transfer the lamp data to a remote display.

In the time intervening between the 1974 conference and the first visit to AEB, the M.I.T. engineer had redesigned the four-dot braille cell into a simpler six dot full braille cell module and was life testing several modules for use in a paperless braille computer terminal. Twenty of these cells were constructed under a second grant from the E. Matilda Ziegler Foundation(6).

INITIAL AGREEMENT

SWB, AEB and M.I.T. began negotiations to reach an agreement on patents, liability and other factors. The first cut of the agreement was unsatisfactory and during the period that followed there were several communications among the principal parties and a delay of several months while differences were discussed. Even before an agreement was signed, it was decided that the construction would begin with support from several sources. The braille modules would be those previously built with support from the E. Matilda Ziegler Foundation; additional components would be funded by AEB; engineering and technical effort would be provided by the Harvard-M.I.T. Rehabilitation Engineering Center, supported by the Rehabilitation Services Administration; and SWB would assist with information and the commitment to provide employment for a blind TSPS operator.

DESIGN OF THE BRAILLE DISPLAY

While the various problems of the agreement were being considered, two engineers at M.I.T. approached the adaptation of the TSPS console for use by the blind in two different ways. First, they considered the problem as it was presented to them, namely the addition of a braille numeric display only. Secondly, the entire scope of the problem was considered. At that time the development of microcomputers had become sufficiently advanced to make their use economically and technologically feasible and the decision was made to use them, even if only in the employment of a numeric display. However, the power of the microcomputer is such that it didn't make sense to do half a job, and the engineers therefore proceeded to work on the complete task of providing almost all information from the TSPS console to a blind operator.

After studying some training material sent from SWB, one of the engineers from M.I.T. went to Little Rock in June, 1976 to obtain a better understanding of the operational aspects of the system. It was a real introduction and included operating a live board within one hour of first arriving at the site, leaving the engineer emotionally drained. At the time of this second visit the hardware interface design was reviewed with SWB engineering personnel. The information gained in these two trips to Little Rock, supplemented by later telephone conversations with SWB engineers and operators, provided the specifications for the system. During the time the hardware was being designed and constructed, an M.I.T. undergraduate student, working on a part-time basis, wrote the control program for the microprocessor(7).

INSTALLATION

The first unit was completed before the final written agreement was worked out. On February 9, 1977, the three groups involved held an "implementation" meeting, where the roles of each party in the installation, training and ongoing use of the equipment were determined. After the meeting, the formal written agreement was speedily agreed upon and signed.

The initial installation of equipment took place during the week of March 14-18, 1977. The system worked, although the numeric display data was not properly received by the microprocessor. The necessary changes were made, and one week later the system was up and running. Shortly after this, the first blind operator was transferred to the site from the "official board" (the SWB PBX where she had been working), to begin training.

TRAINING PROCEDURES

There are two major differences between training provided for blind and sighted operators. Since no training materials were available in braille, it was decided that training for blind operators would be done on a one-to-one basis, with the instructor reading the material while the blind trainee made braille notes for later study. The normal ratio is one to six trainees per instructor. Instead of practicing on a simulated console with paper tape and magnetic tape inputs, the blind trainees used a "directionalized" live console under close supervision. A directionalized board allows the calls to be selected so that the trainee services the kinds of calls recently studied. The live board training was found to be very effective and the time involved for this phase of the training was essentially the same as that for sighted trainees.

Most of the increased training time for blind operators was due to the need to make accommodations for the Register. The Register is an ink-print file at each operator position containing emergency numbers, rate and routing information, etc. Each operator brailled her own Register. This facilitates its use, since they are responsible for its contents and organizational pattern. The braille Register is contained in a 5000-card Rolodex file.

OPERATION

During the early months of operation, Banks or Perkins braille writers were used for taking notes and remembering data. It finally occurred

to the M.I.T. developers that a braille "scratch pad" could be built into the system, using the existing braille display, by adding a keyboard and some computer memory and by modifying the microprocessor's control program.

In early January of 1978 a second braille display was installed and the first unit was modified to include the braille "scratch pad". In addition, the microprocessor computer operating program was updated to reflect features which had been added to the TSPS system since the installation of the first braille display. When installed, the second system did not work initially because the TSPS console to which it was connected had much more electrical noise than the console for the first braille display. The necessary design changes were worked out and installed in both displays during the first week of February, 1978.

The first blind operator hired at SWB was employed by them until May, 1978, at which time she moved to another city. She later returned to Little Rock and was rehired in May, 1979. While working as a TSPS operator, her working time (the average time spent per call), a measure of productivity, has been consistently much better than average(8). In February, 1979, a second blind operator was hired. Her performance has been at or near the average level for sighted operators. Both operators are currently employed.

THE PACIFIC BELL SYSTEM

At about the time the first M.I.T. system was installed at SWB, another console for blind TSPS operators was installed at a Pacific Bell (PB) TSPS site in Mountain View, California(9). This system, developed by Telesensory Systems, Inc. (TSI) under support from the Sensory Aids Foundation, both in Palo Alto, California, used voice output to provide the TSPS status information and numbers. Two such systems were in use for about one year.

TECHNOLOGY TRANSFER

The success of the M.I.T./SWB program and the similar program at Pacific Bell (PB) prompted AT&T to consider a follow-up program that would result in a commercially available system and allow any of the AT&T operating companies to employ blind TSPS operators. A meeting was held in Palo Alto in May, 1978, where personnel from AT&T and Bell Laboratories reviewed the results of the M.I.T./SWB and TSI/PB programs with all parties concerned. At that time a course of action leading to a commercially produced display for blind TSPS operators was determined.

First, Bell Laboratories developed a TSPS/TIPS (TSPS Information Processing System) interface which could serve as a standard interface, meeting all Bell System practices. Then AT&T contracted with TSI to develop a braille/audio display system, incorporating speech output similar to that used previously by TSI for the "Talking Calculator" and the "Talking Optacon". The braille output display will be the same as that in their soon-to-be-released Versabraille. An M.I.T. engineer participated in TSI's initial design work, providing memorandums describing the requirements for such a system based on his experience with the Little Rock Installation. The M.I.T. engineer also outlined the message structure used in the Little Rock system and proposed a message structure based on the longer, 20-cell braille display to be used with the TSPS/TIPS display.

The present contract from AT&T was for three units. The first will be delivered to Bell Laboratories for testing and development of operating procedures. The other two will be installed at an operating company for evaluation under conditions of actual use.

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AN AUDITORY TECHNIQUE FOR VISUALIZATION OF LINE-DRAWINGS BY THE BLIND

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Mapping linear (i.e. line-like) contours from a two dimensional visual space into an auditory space might be accomplished by letting the trajectory of a moving sound source trace the contour. Motivation for exploring this possibility is provided by the needs of the blind: while the printed word is gradually becoming accessible to them via spoken-output reading devices, no efficient method exists for the auditory (or, for that matter, tactual) presentation of even the simplest graphic images. Yet a rapid mode of access to graphs and plots would be of great value to blind scientists, engineers, technicians etc. In exploratory experiments subjects listened to auditory versions of oscilloscopic displays of triangular and sinusoidal pulses. The traces were swept slowly, about 3-5 seconds per sweep. Horizontal displacement was conveyed by variations in the relative intensities of the dichotic stimuli. Vertical displacement was translated, after exponentiation, into frequency variations of a voltage-controlled oscillator. Time epochs (e.g. onset and termination of the displayed pulse) were adjusted quite accurately by our blind subject, and he could also distinguish fairly well between triangular and sinusoidal pulses.

BACKGROUND

The differences between the ways our eyes and our ears perceive the world have been the subject of numerous studies. A good recent review is that by Julesz and Hirsh [1]. Perhaps the most striking contrast is, as they point out, that whereas the basic currency of the visual experience is objects, that of audition is events. In other words, our eyes explore the space around us, which may be motionless, while our ears respond only to changes over time (of instantaneous sound pressure, power spectral distribution, etc.).

The motivation for bridging the gap between these two modes of perception is greatest when we consider those whom misfortune has deprived of one: the blind and the deaf. Attempts to make the deaf "see" speech, undoubtedly the most important acoustic signal for humans, have so far not succeeded. Visual inspection of a sound spectrogram conveys much information to the trained eye, but not enough to allow "reading" it in real time.

Transformation in the reverse direction has been more fruitful: mobility and reading aids for the blind, based on acoustic codes, have been in use for some time [2]. The main difficulty lies in conveying a sense of space, a notion particularly difficult to grasp for the congenitally blind. Three strategems have been commonly employed. First, time has been given the role of a pseudo-spatial dimension. Thus progression say from left to right was translated, via uniform motion, into variation with time. Second, pitch has been allocated the role of height, owing to its common association with physical height. Third, stereophony has been employed for the creation of images in space.

In order to clarify our terminology, we shall now briefly characterize the various auditory spaces that can be created and experienced.

ON AUDITORY SPACES

Space is something we have no direct experience of; only its boundaries, or objects found within, can be perceived. Its dimension is determined by the minimum number of independent parameters required to specify the coordinates of a point in the space.

We shall be using the term "auditory space" somewhat as a metaphor, for one of its "dimensions" (pitch) has no metric equivalent in our physical space. Also, unlike in physical space where motionless "objects" are experientially meaningful, all events in the acoustic space require the inclusion of time as a dimension.

In the next subsections we shall review three kinds of auditory space; the dimension of pitch will be discussed in the section afterwards.

The "Real" External Space

This denotes the space in which actual physical sources of sound are located, i.e., the space of our everyday experience. To map up this space we would require, ideally, dimensionless "point sources", which can only be approximated.

The auditory localization of sound sources in the external space has been extensively studied over several decades; a good review is given by Mills [3]. It is well established that interaural time and intensity differences provide the main clues to the determination of azimuth, while range (distance) is inferred from spectral characteristics and the relations between direct and reflected

sound components. Elevation determination requires some head movement, while moving sound sources are detected, in addition to their changing static parameters, by the Doppler effect they produce.

The "Illusory" External Space

The term "illusory", borrowed from Chowning [4], should actually qualify the apparent sound source, not the space. It refers to an experimental arrangement in which four (or more) loudspeakers, generating sound under computer control, create the illusion of a well-focused sound source being "out there". The listener, seated at the point of intersection of the lines connecting the diagonally opposed speaker pairs, has the distinct sensation of the source, especially when it is moving. Elaborate curves can be described by this phantom "tracer", even including some elevation effects. The curves are originally traced out by a light pen; the program introduces artificial intensity differences, reverberations and Doppler effects [5].

The Internal (Intracranial) Space

It was for long believed that a sound originating from a true external source or simulated by loudspeakers is localized by the listener "out-of-head" (OHL), while dichotic stimulation via earphones inevitably results in in-head localization (IHL). The latter is the subjective sensation that the source of the sound is "inside one's head", and is frequently distinguished by the use of the term "lateralization". The space in which such internally localized images exist is thus inside the cranium.

Recent work [6,7,8] indicate that both IHL and OHL can be achieved by either earphone or loudspeaker stimulation, and that the criteria determining whether sources are localized internally or externally (or both at the same time), are complex. Schemes have also been published [9], for "externalizing" the sound of conventional stereophonic recordings when listened to via earphones. Some of the older literature on the localization/lateralization dichotomy must thus be treated with caution.

Lateralization appears to be influenced mainly by interaural phase differences below 1500 Hz, while at higher frequencies intensity differences take over. However, time differences in slowly varying envelopes of high frequency two-tone complexes can be significant in lateralization.

THE ROLE OF PITCH

Although a universal "pitch scale" does not exist, the "height" aspect of pitch has been agreed upon well enough for sensory aids to exploit it. A variety of reading devices have been reported during the past 60 years or so, in which the vertical coordinate of an alphanumeric character was mapped into the pitch of a component tone in a chord. These are known as the Optophone [10], the Visotoner [11], the Lexiphone [12], the Stereotoner [13], etc. The above devices were all essentially monaural, or at least monophonic devices. (In the Stereotoner higher pitched components are heard louder in one ear while lower pitched ones in the other ear. Whatever stereophonic effect is created bears no relation to the horizontal scale on the paper.) In all these direct translation devices users must move a probe from left to right along the line, so that the visual image of a character becomes translated to a time-varying pattern of chords. With the gestalt of the character thus destroyed, readers have difficulty in associating the code pattern with the character. These pitch-based machines are rapidly being displaced by the tactual Optacon and, eventually, the speech output systems by Kurzweil Computer Products Inc. and Telesensory Systems Inc.

Pitch and Internal Space

Three examples will now be given in which the dimension of pitch was combined with the (horizontal) dimension of the internal auditory space in order to create 2-dimensional effects.

Julesz and Levitt [14] tested the discriminability of "spatial chords" – 5-tone chords presented dichotically, with the internal location of the component tones independently controlled by interaural time delays. Chords were presented in pairs lasting 500 ms each, and one or more of the component tones were shifted in subjective location between members of the pair. Though none of the listeners were able to detect the individual component and its changed location, they succeeded well in discerning the changes on the basis of some overall audible change in the chord. The "global properties" of the chord were thus altered via a change in its "geometric layout", even though its spectral content was invariant.

The dimension of time was added to pitch and lateral position by Fish [15 and 16], who employed a flying-spot scanner or television camera to raster-scan two-dimensional images. The scanner output turned on and off a tone oscillator, whose frequency was a (logarithmic) function of the height of the scan line. The horizontal location of the scanning spot was made to correspond, via (logarithmic) amplitude differences, to the lateral position of the tone. Fairly good discrimination of simple geometric patterns (up to six solid shapes in a field) were reported.

Most relevant to our proposal is the acoustic equivalent of the T.V. "Pong" game, named "Paddleball", and developed as part of a game package for the blind by Telesensory Systems, Inc. [17]. In this game both "paddles" and "ball" are created by tones whose pitch corresponds to height. Left-right motion of the ball is accomplished by interaural intensity differences. Like in its video counterpart, the ball moves in straight lines back and forth, up and down; to "hit" it, the player must move his paddle (easily identified by its chopped tone) to the correct "pitch position" to meet the arriving ball. The successful player must thus aurally extrapolate the trajectory of the ball.

While in the "spatial chords" of Julesz and Levitt [14] the potential for employing pitch as a

vertical dimension was only implied, the technique of Fish and the Paddleball game do so explicitly. In a subsequent section we describe our preliminary experiments in "drawing" traces on the laterality-pitch plane.

Finally, an important feature of pitch discrimination should be noted. The hearing mechanism, in common with physical instruments, cannot reliably resolve small differences in frequency unless they are observed for a sufficiently long time. This is reflected both in absolute pitch judgments for short tonal bursts and in difference limens for their frequency; comparable effects with tone glides can no doubt be found.

Experimental evidence obtained by Sekey [18] was interpreted as indicating that when the rate of such temporal changes varies, the ear swiftly adjusts its "analyzing bandwidth" to the stimulus it receives, or is expecting to receive. This may account for the coexistence of very small JND's for fundamental frequency and formants on the one hand, and acute perception of temporal events (e.g., formant transitions) on the other, albeit not at the same instant of time.

RATIONALE FOR THE PROPOSED TECHNIQUE

Work previously described has shown that the illusion of a moving "point-like" sound source can be created artificially via synthesized binaural signals. The trajectory "traversed" by such a moving image will be referred to as a "sound path". The sound path is thus a trace in an intra- or extracranial auditory host space.

We hypothesize that by temporal integration of binaural stimuli listeners could retain in short-term memory entire sound paths, and perceive them as gestalten. If so, a mapping from a two or three dimensional (visual) space into the auditory space could be accomplished by a moving sound source traversing the contour to be mapped. Closed linear (i.e., "line-like") contours may be mapped via a continuous, repeated passage by the source, while open ones (e.g., oscilloscopic traces) require a perceived or suppressed retrace path. For brevity, the term "trace" will be used to designate both closed contours and open traces.

The characteristics of the auditory host space require further exploration: it may have a measurable extent, a lower limit of resolvable positional differences, and perhaps also some characteristic "curvature" or "shape". Its coordinates, aximuth, range and elevation (or altitude), may be substantially "warped". Altitude, whose perception is expected to be limited, will be augmented by the parameter of pitch. Additional dimensions (e.g., modulation, timbre) would be introduced if they were found to enhance the rate of information transfer.

Our technique is still largely experimental, combining passive listening with active control of sound image movement by subjects. Some preliminary experiments, aimed at exploring our working hypothesis, are described in the next section.

EXPERIMENTAL RESULTS

We assembled the signal generating system shown in Figure 1. in order to study the phenomenon of tracing curves in the pitch – latitude plane. The sweep output of the Wavetek 144 was set at a conveniently low rate, about 3-5 seconds per sweep. The sweep was converted in subsequent stages of the analog computer into gain functions for each ear. Both linear and exponential variations [16] were possible. Notwithstanding the logarithmic sensitivity of the ear to volume, we found the linear variation to yield a more uniform subjective velocity from ear to ear.

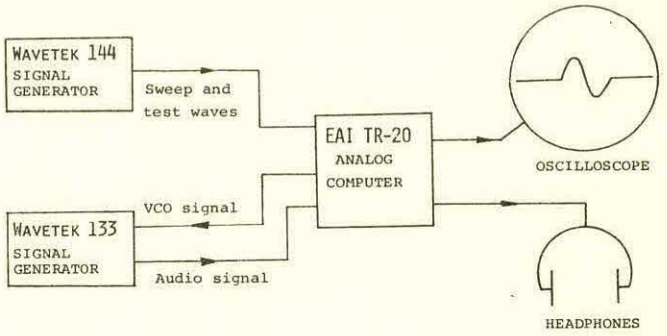

Figure 1. Schematic diagram of the signal generating system

At a set point of each sweep, under potentiometer control, a triggered signal generator produced a single pulse of adjustable duration. The pulse shapes employed were triangular and sinusoidal, including skewing and arbitrary phase shifts. These pulses were exponentiated so that the logarithm of the VCO control voltage was made proportional to the pulse as seen on the screen. The exponentiated pulses caused a proportional variation in the frequency of the audio generator. Typical values for a symmetric pulse were:

$$f_{min} = 133 \text{ Hz}; \; f_o = 200 \text{ Hz}; \; f_{max} = 300 \text{ Hz}.$$

The frequency was thus deviated by a factor of 1.5, or a Pythagorean fifth. This was kept fairly constant, since the present instrumentation necessitates lengthy calibration to ensure (logarithmically) symmetric frequency deviations.

Informal experiments were performed on one blind subject, J. R. The following types of task were given to him:

(1) Describe your impression of the waveform on the screen.
(2) Shift the waveform so that it begins at a specified point.
(3) Alter the width of the waveform so that it extends to a given proportion of the sweep.
(4) Determine if the wave displayed is a sinusoidal or triangular pulse.
(5) Adjust the skewed triangular wave until it becomes symmetrical.
(6) Adjust the phase lock control (of a Hewlett-

Packard 3300A Oscillator) so that the single sinusoidal pulse begins and terminates at any one of 9 preassigned levels. (The frequency equivalent of the levels ranged a musical fifth above and below the center frequency.)

While these pilot experiments were intended only to provide insight and suggest more formal tests, the results were instructive and encouraging. The following observations were made:

i) In tests involving horizontal coordinates, our subject relied more on his sense of time than on interaural space; however, his impressively accurate performance (0-5% of the screen width) deteriorated when one of the earphones was removed.
ii) Discrimination between triangular and sinusoidal pulses was almost faultless after the first couple of trials.
iii) Standard deviation in adjustments involving the vertical direction (i.e., pitch) was 6-8% of the total range. This is fairly poor, but was accomplished without training.
iv) Subject preferred a scan time of 3-4 seconds; he reported a sensation of the image "moving around the head band of the earphones".

CONCLUSIONS

The significance of the proposed work lies in the exploration, perhaps for the first time, of the internal sound space as a host space for the stereophonic, time-varying presentation of graphic images to the blind. (Stereophonic hearing has been previously exploited for mobility aids [2]). At present only two experimental projects are known to us offering limited solutions to this type of problem. One is an as yet unreported scheme by workers of the Smith-Kettlewell Institute in which blind operators can explore a waveform displayed on an oscilloscope by moving a sliding contact back and forth; the pitch of a (monophonic) tone corresponds to the amplitude at each point. Good results have been achieved in the discrimination amongst 8 different waveshapes. The other is a tactual device named "sensory quill" [19], which produces raised line drawings on stiff paper under manual or automatic control.

Neither of these devices has the speed desirable for efficient interaction with an oscilloscope. Also, the former does not lend itself to the creation of a spatial image, while the latter is slow, bulky and uses costly paper.

We believe that our proposed technique could thus represent a new departure in making an important class of graphic images, namely traces and contours, accessible to the blind. Moreover, our exploratory experiments will be sufficiently general to open the way to other applications, for non-blind users, such as audio diagnostic tools, hearing tests, learning aids, etc.

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CLINICAL APPLICATION OF A DESKTOP COMPUTER-AIDED DIGITIZER SYSTEM FOR THE ANALYSIS OF SPINAL DEFORMITIES

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ABSTRACT

Analysis of spinal deformities using routine e clinical methods is severely restricted in accuracy and reproducibility. Determination of the true three-dimensional geometry of an unstable and deforming spine is not possible. The absence of an accurate assessment of the actual spine geometry and the lack of suitable criteriae for the prophylactic management of a deforming spine often results in a delayed decision on the orthopaedic treatment of the disorder. The problem as it applies to non-ambulatory children with Duchenne Muscular Dystrophy is being solved through the use of a desktop computer-aided digitizer system. The system performs a complete analysis of the three-dimensional geometry of the spine from data digitized off routine spine radiographic films.

THE PROBLEM

Clinical assessment and quantification of spinal deformities (scoliosis) from routine radiographic films (x-rays) is frought with limitations of accuracy and reproducibility. This is particularly so in diseases which manifest themselves in a severely unstable spine with deformation taking place in several planes simultaneously. Such deformities are impossible to determine from the single spine radiograph usually obtained for the assessment of scoliosis.

Shortcomings like these, seriously compromise the ability of the clinician to accurately assess and diagnose the problem presented by a spine which is unstable and deforming. Delaying clinical decision on the course of treatment of an unstable spine either by conservative means (external bracing) or by surgical stabilization may have serious consequences on the success of the clinical management of the disorder.

A SOLUTION

The foregoing statements apply to the clinical problem of scoliosis in general. This paper will present a solution as it is being applied to a particular example of spine instability, namely the non-ambulatory victim of the Duchenne type of Muscular Dystrophy (M.D.).

A portable desktop computer-aided digitizer (Figure 1) is situated in the M.D. Clinic at The Hospital For Sick Children. Antero-posterior and Lateral x-ray films are placed, one at a time, on the digitizer platen and a cursor is placed in turn over twenty anatomical points on the spine and pelvis on each x-ray view. The coordinates of each of these points are sent to the computer and stored for later use. The coordinates are then used to establish a three-dimensional geometric representattion of the spine. Three views (P-A, Lateral and Apical) are then presented to the clinician together with a complete quantitative analysis of the spine and pelvis in both two and three-dimensional form (Figure 2). The whole process takes a few minutes while the patient is waiting to be seen.

RESULTS

Maximum curve

Among the quantitative results is an indication of the maximum spine curvature and the location of the plane of maximum curvature in relation to the P-A view. As an option, the P-A view may be rotated at ten degree intervals through the plane of maximum curvature and into the lateral plane (Figure 3). For each ten degree interval the curvature, curve limits and apex for that view are calculated and labelled on the graphical presentation.

Review of stored data

Another option which is available retrieves digitized coordinates from x-rays taken during previous clinic visits. Graphical presentation of the previous spine curvatures may be superimposed on each other (Figure 4) for direct comparison. Any number of previous examinations may be retrieved for review. At the end of the review, a summary report on the quantitative results of each examination chosen for review is presented along with a graphical representation of some of the main parameters of clinical interest. (Figure 5)

Perspective plots

Additional options for graphical presentation of a patient's file include perspective plotting of P-A and Lateral views or Maximum and Minimum planes of curvature either separately or simultaneously on the same perspective plot.(Figures 6 and 7)

Statistical analysis

Quantitative results for the latest analysis for each patient are stored in a string data file. Variables stored in this file may be analyzed

statistically for the entire patient population on file or for any sub-group of patients selected by any number or combination of criteria defined in terms of the variables being stored. Data points for any two of the variables analyzed may be presented on a scatter plot and a polynomial regression analysis may be performed on the data set (Figure 8).

CONCLUDING REMARKS

The foregoing is an extremely brief description of a comprehensive analytical approach to the problem of spine and pelvic instability in non-ambulatory children with Duchenne M.D. The approach was originally conceived in 1973, and implemented as a research tool in 1975 for the M.D. Clinic at The Hospital For Sick Children. In April 1979, the system was implemented in the Clinic and at the same time, any M.D. clinic in Canada (sponsored by the Muscular Dystrophy Association of Canada) was invited to participate by submitting data on their patients and sharing in the data base already established at The Hospital For Sick Children. At the time of this writing (February, 1980) the data base consists of 225 patients with orthopaedic followup ranging from one to seven years.

Selected examples of the clinical significance of the approach just described will be presented and discussed.

ACKNOWLEDGEMENT

This work is funded by the Muscular Dystrophy Association of Canada.

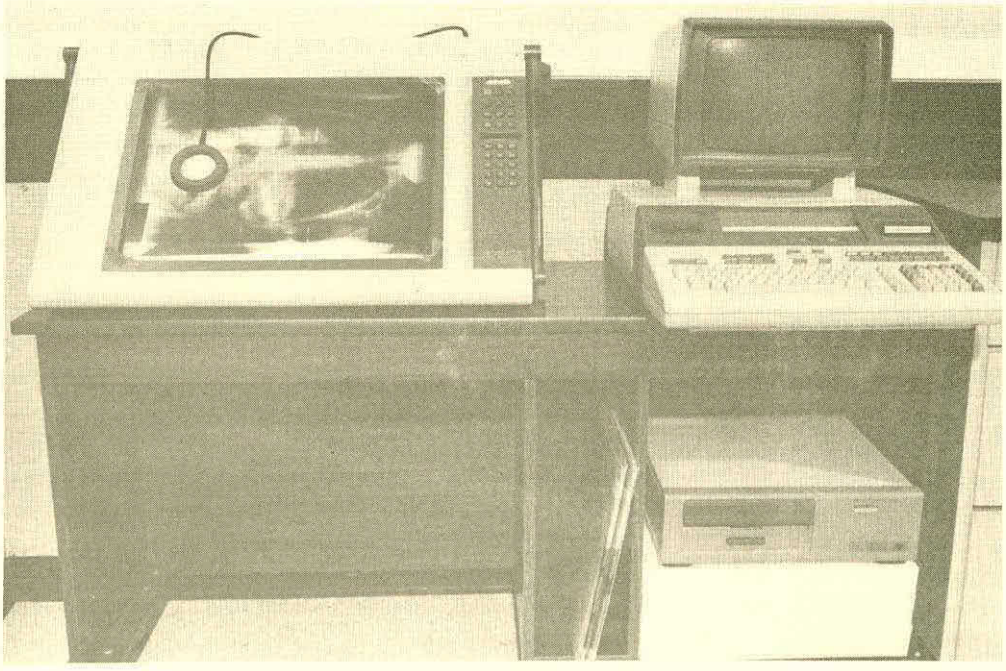

Figure 1: The portable desktop computer-aided digitizer system.

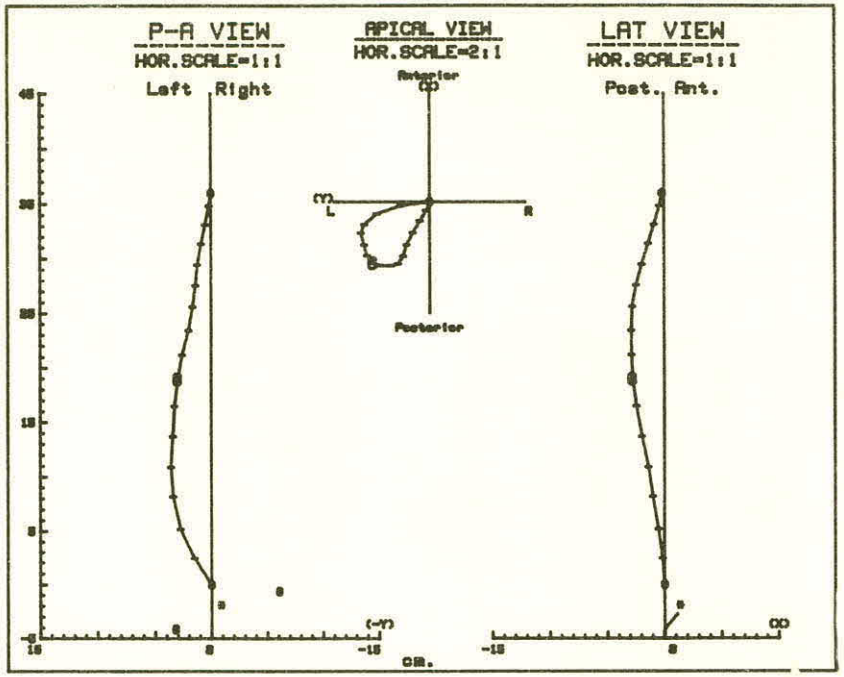

Figure 2: Graphical representation of the three spine views.

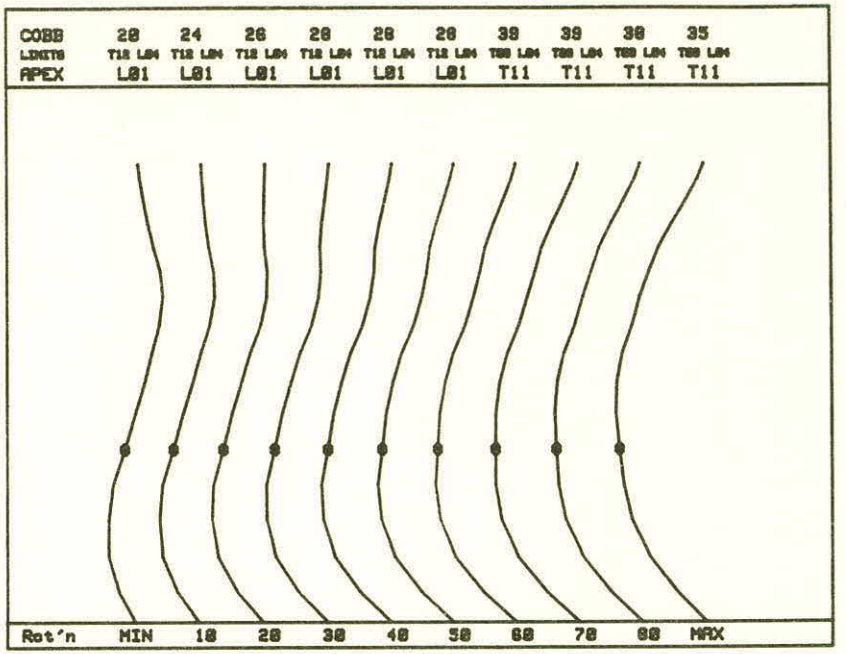

Figure 3: Graphical rotation of spine view about the vertical axis

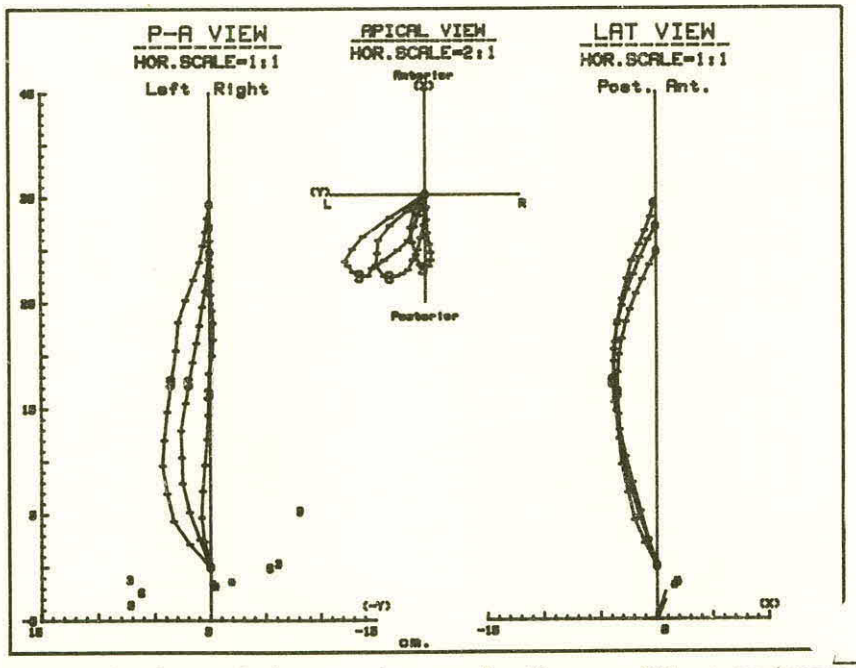

Figure 4: Plots of three review examinations on file superimposed on the same graphical presentation.

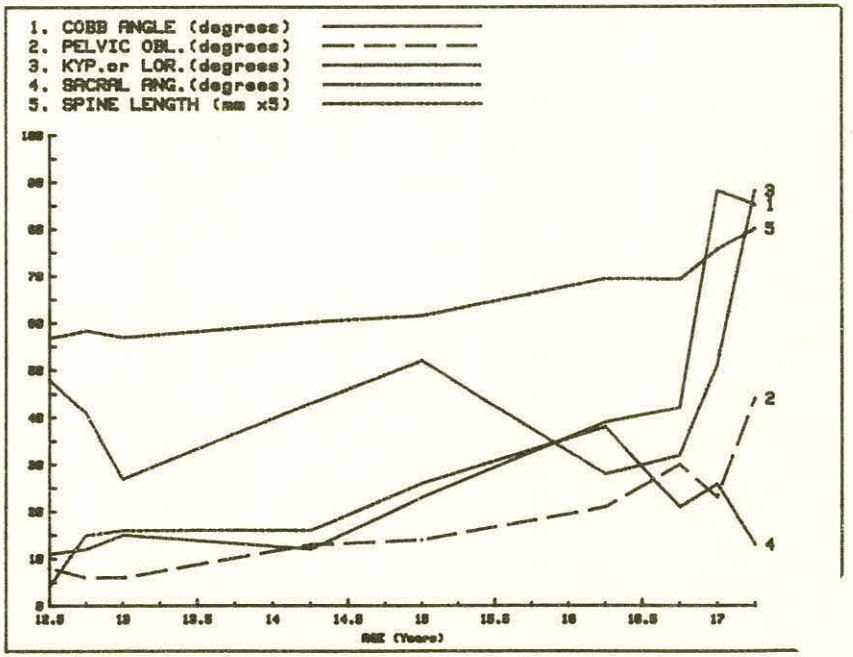

Figure 5: Review summary graphical presentation for a single patient of some clinical spine and pelvis orthopaedic parameters plotted as a function of age.

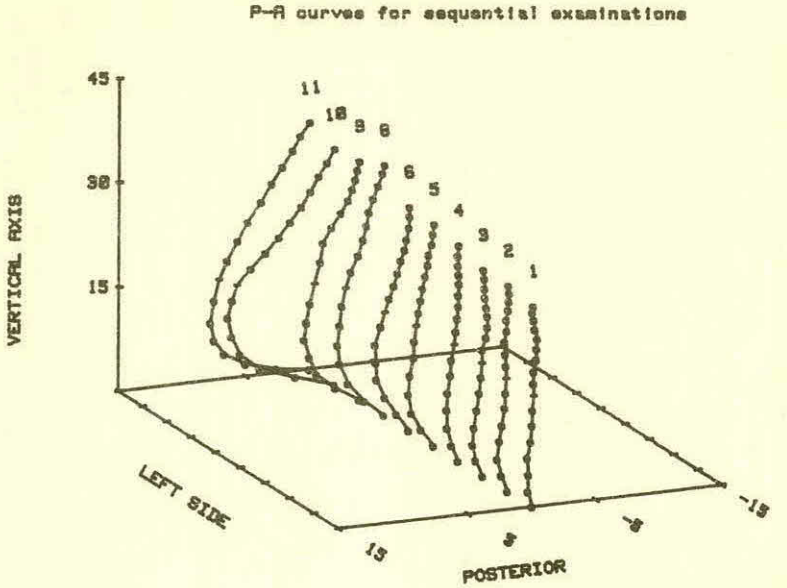

Figure 6: Perspective spine plots of P-A view curves for one patient.

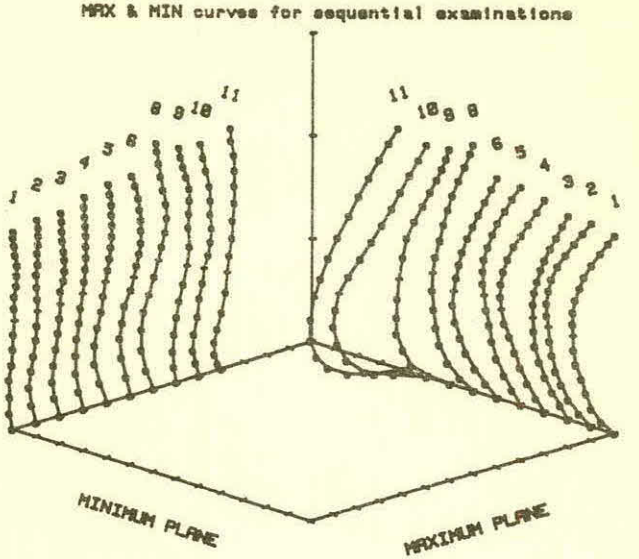

Figure 7: Perspective plots of Maximum and Minimum curves for the same patient.

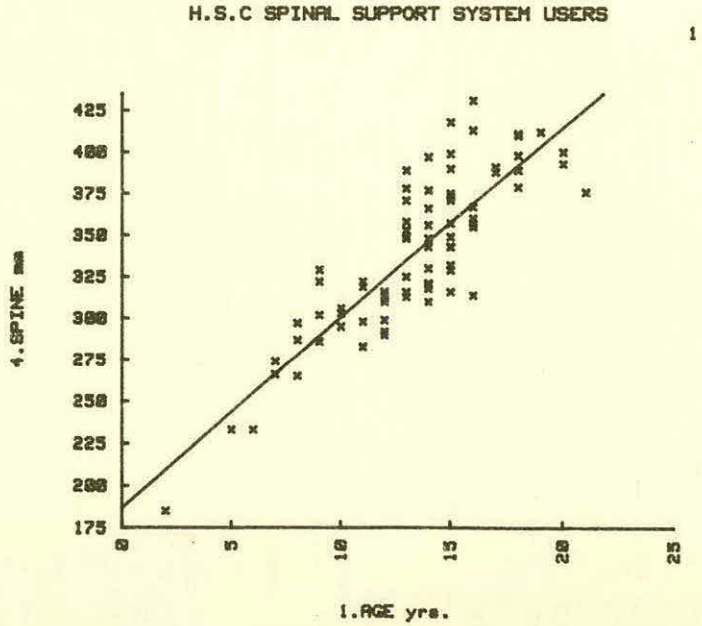

Figure 8: Example of a scatter plot and regression analysis of two variables (in this case Age vs. Spine length).

BIOMECHANICAL REQUIREMENTS FOR A BETTER SITTING POSTURE FOR FRIEDREICH ATAXIA PATIENTS: PRELIMINARY REPORT

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ABSTRACT

This paper presents a one year study on the evolution of scoliosis in Friedreich ataxia patients. Fourteen boys and seven girls who had no surgery at the time of this study are followed by means of standardized spinal radiographs taken in the Scoliosis Chariot or the Throne. Several indices have been investigated such as the location of the apex of the deformity, the length of the spine and the kypho-scoliosis index, and analysed in an attempt to define the biomechanical requirements for better sitting posture. Preliminary results seem to indicate that Friedreich ataxia would benefit from, a posterior lumbar support, low thoracic lateral supports and a slight inclination of the seating system.

INTRODUCTION

The management of scoliosis affecting Friedreich ataxia patient has been so far limited to body braces and back surgery (Geoffroy et al.1976). The sparsity of this type of neurological disorder makes it difficult to carry out an adequate clinical and analytical study; thus, little is known about the disease itself as well as its management. Recent studies by Barbeau et al. (1976,1979) have shed new light mostly on the biochemical aspects of the etiology of this pathology. Concurrently with Barbeau's studies, a research programme on the etiology and management of scoliosis in Friedreich ataxia was initiated at Ste Justine's Hospital where patients are regularly seen in the Neuro-Muscular Clinics. In an attempt to determine the requirements for a better sitting posture, the need was felt to investigate the progression of scoliosis from the early onset when the patients are still ambulantory to the time when they are wheelchair-bound with severe structural deformation of the spine.

The report presents the preliminary results of the first year study on the evolution of scoliosis in fourteen boys and seven girls. Thirteen scoliosis-related parameters are extracted from forty-seven standardized spinal radiographs of patients diagnosed as having Friedreich ataxia and having not undergone any back surgery yet.

METHOD

An important consideration in the study of scoliosis by means of radiographs is the accuracy of the measurement method. Traditionally, the antero-posterior spinal radiographs are taken with the patient either in standing or in supine position. If additional lateral radiographs are taken, they are viewed separately. The physician and the chiropractician quantify a tri-dimensional or spatial deformity of the spine from its projection in a single plane, namely that of the radiograph. It has been shown (McNeice and Dawson, 1976) that the currently accepted technique of obtaining antero-posterior and lateral radiographs leads to inconsistent measurement in scoliosis.

In this study, standardized spinal radiographies were obtained by means of the Scoliosis Chariot shown in Fig. 1 and the Throne shown in Fig.2 for ambulant and non-ambulant patients respectively.

The Chariot, designed by McNeice and Dawson (1976), consists of a booth positioned in such a way that when the patient stands in it, he is in a true lateral position to the X-ray tube. The first spinal radiograph is taken; subsequently, the Chariot is rotated through 90° and positioned so that the patient now stands relatively to the X-ray tube in a true antero-posterior position in which the second radiograph is taken.

The Throne, designed by Koreska et al. (1978a), consists of a seat on which the patient is placed with his back resting against two referenced plexiglass rods and with the side of his pelvis leaning against the lateral support of the Throne. Antero-posterior and lateral radiographs are taken by positioning the Throne accordingly against the X-ray cassette.

Utilising these radiographs, the centroid of each vertebra between the 7th cervical and the 1st sacral as well as the reference scales are located and then traced on an overlay transparent sheet of paper. These tracings are sent to the Depart-

ment of Civil Engineering, University of Waterloo Ontario, where they are digitized. The information is fed into a computer programme which generates a graphical display of the spine and computes several tri-dimensional indices (Koreska et al. 1978 b). This information is mailed to Montreal for clinical analysis.

PRELIMINARY RESULTS AND DISCUSSION

Although thirteen scoliosis-related parameters are included in the computer analysis of the standardized radiographs, four of them, namely, the spinal deformity angle, the location of the apex of the scoliosis, the length of the spine and the kypho-scoliosis index, pertain particularly to the seating definition. The remaining ones provide mostly useful information on the etiology of the scoliosis.

The spinal deformity angle is measured by means of the Cobb method (Cobb, 1948). Table I presents the average values of spinal deformities classified in four age groups consisting each of a five year interval. These results seem to show the existence of a transition region between the age groups I and II where the Cobb angle increases by 108% and 167% for girls and boys respectively. Between the age groups II and III a less marked second transition is noticed, both in the girls and the boys where the Cobb angle increases respectively by 15% and 50%. A slight decrease in the spinal deformity in the following age group may be explained by the small number of observations for that age group. None of these increases observed in the deformity is related to a substantial growth spurt as evidenced by the absence abrupt changes in the true length of the lumbar thoracic and thoracic lumbar spine shown in Fig.3.

The apex of the scoliosis shown in Table II, seems to be shifting towards the low region of the thoracic spine as the spinal deformity increases. For the girls, the location of the apex changes from T8 to L3 with a transition from T8 to T10 at approximately the age of ten, followed by a stable period between the ages of 10 to 20, where the apex is situated at the T10 level.

For the boys, the apex changes from T6 to T8-T9 where it seems to stabilize. The most noticeable transition of T6 to T9 occurs again between groups I and II.

To restrain the progression of scoliosis in non-ambulant patients, lateral support at the T9 level should be useful.

For this study, the lumbar kypho-scoliosis index (KSI) (Raso, 1977), is a measure of the lordosis where as the thoracic KSI emphazizes the degree of kyphosis. As shown in Fig. 4, both indices increase between the age of 6 and 8, and then remain fairly stable until the age of 17.

After, the lumbar SKI value decreases by 75% while the thoracic KSI remains approximately constant. This represents a loss in the lumbar lordosis towards a more kyphotic posture, probably as a result of the sitting posture assumed by the non-ambulant patient.

Since the scoliosis encountered in Friedreich's ataxia is paralytic, the evolution of lumbar lordosis in a kyphosis is to be expected, resulting in an increased scoliosis and permanent structural deformation of the spine.

The use of a posterior lumbar support in an inclined seat such as the Hospital for Sick Children Spinal Support System (Koreska et al. 1975), may delay the progression of the scoliosis.

SUMMARY

From the preliminary results an attempt to define the sitting requirements of Friedreich ataxia patients could be summarized as follows: (i) a posterior lumbar support (ii) lateral support at the low throacic level, approximately at T9 and (iii) a slight inclination of the seating system.

ACKNOWLEDGEMENTS

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TABLE I

Average Spinal Deformities

(Cobb angle)

Age Group	Boys	Girls	Total
I (5-10)	9	13	11
II (10-15)	24	27	25
III (15-20)	36	31	34
IV (20-25)	30	22	28

TABLE II

Average Location of the Apex of the Spinal Deformity

Age group	Boys	Girls	Total
I (5-10)	T6	T8	T7
II (10-15)	T9	T10	T9
III (15-20)	T9	T10	T9
IV (20-25)	T8	L3	T10

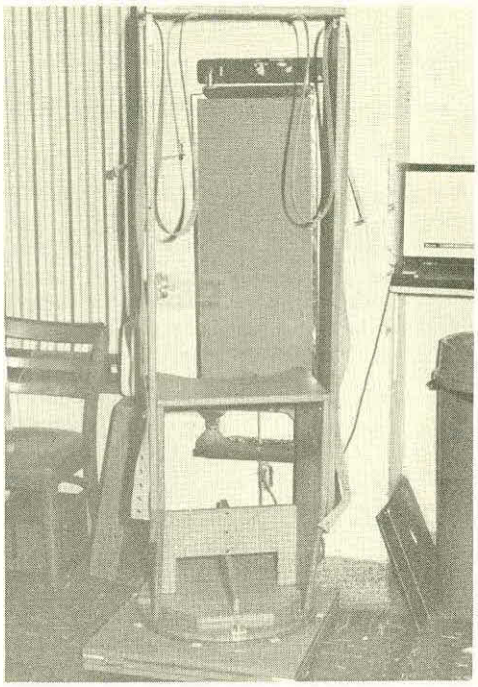

Fig. 1 The "Scoliosis Chariot"

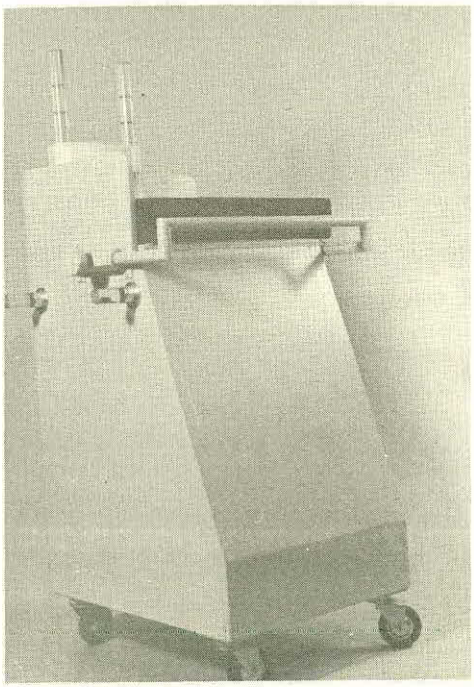

Fig. 2 The "Throne"

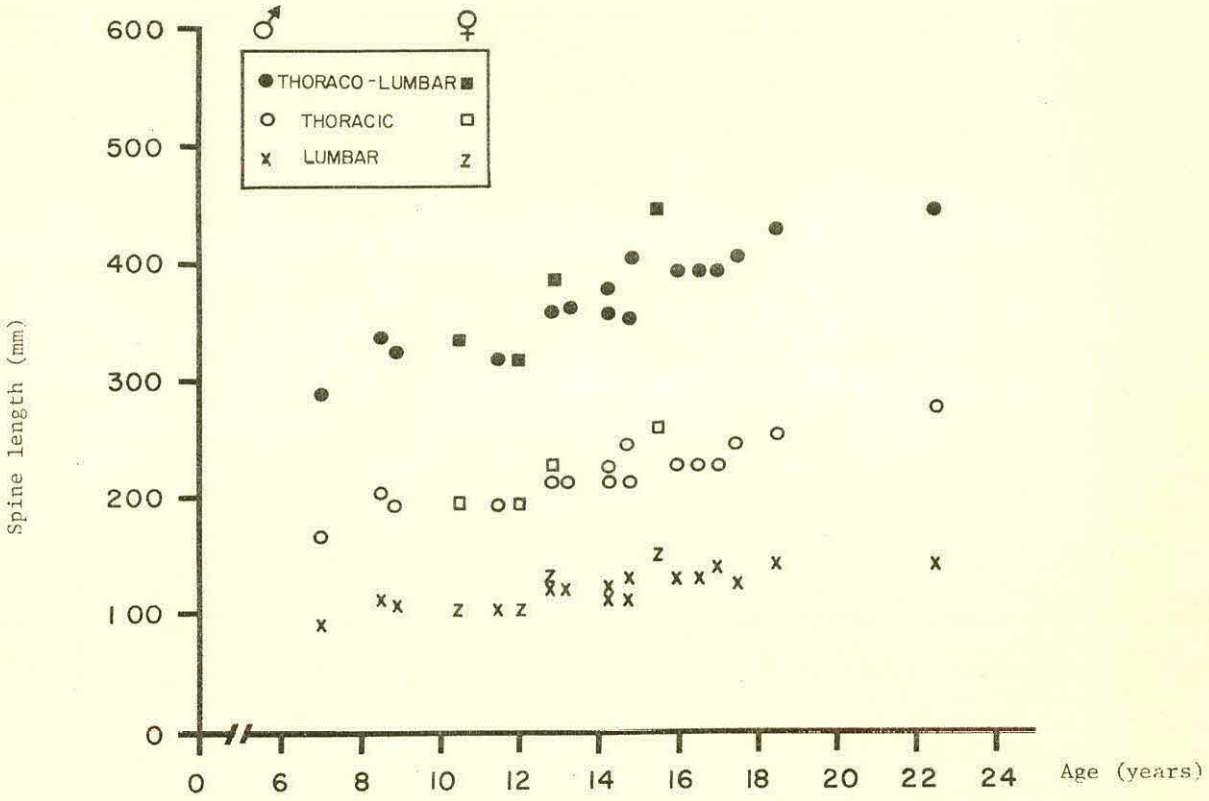

Fig. 3 Computerized length of the thoracic, lumbar and thoraco-lumbar spine

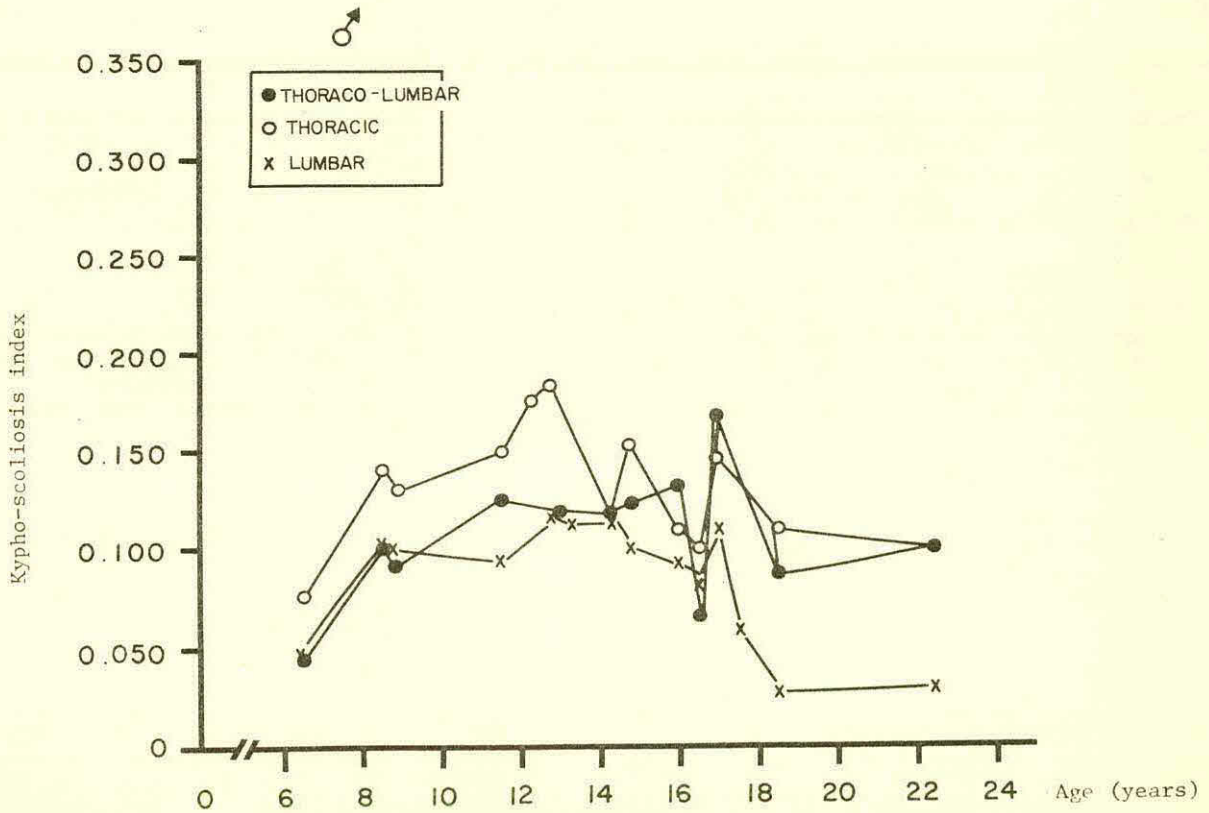

Fig. 4. Kypho-scoliosis index for the thoracic lumbar and thoraco-lumbar spine

A NEW CONCEPT FOR THE CORRECTION OF A LATERALLY DEFORMED SPINAL COLUMN AND FOR ITS FIXATION IN THE ADJUSTED STATE

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1. ABSTRACT

The present paper deals with the correction of lateral deformation (or bending) of the spine (referred to as Scoliosis) from an engineering perspective.

First we present a short description of the conventional method for the treatment of severe scoliosis. In the process of adjustment of the deformed spinal column, it gives rise to, on the one hand, a high flexure stress, and, on the other hand, a residual deformation (or curvature) that cannot be eliminated. Our mechanical analysis points out the cause of these shortcomings.

Then, from this mechanical analysis, a concept is developed for an adjustment (or a distraction of the spine) process which will enable a complete correction (of the deformation) with only a slight flexural stress on the spinal column.

Aspects of construction of a device or prototype development (to some extent theoretical and yet quite rigorous and detailed) based on the concepts developed in the paper are briefly introduced. These designs could, if subjected to further systematic development, eventually lead to the replacement of the conventional adjustment procedure and instrumentation system.

2. INTRODUCTION

Concerning the stabilization of the adjusted spinal column by the conventional mode of treatment, the main points of criticism are the lack of mobility in the sagital plane (in bending the spine forward), inability of the spine to rotate about its axis, and the excessive size of the implanted metal parts. Therefore, in this paper, we will first outline the shortcomings of the conventional method by means of mechanical analysis and from this we will develop a concept for adjustment and complete correction of a deformed spinal column.

3. DESCRIPTION OF THE CONVENTIONAL TREATMENT METHOD

The following review is in a way based on the papers of P.R. Harrington (1-2).

In contrast to Kyphosis (posterior spinal deformation) and Lordosis (spinal deformation in an anterior direction), traction devices, massage of the musculature of the back, and other medicomechanical methods, surgical intervention is necessary in the case of Scoliosis when a certain degree of severity is reached. The degree of a scoliosis is determined by the Harrington Factor, which is the ratio of the angle of deformation (in degrees) and the number of vertebrae in the deformed area (equal to 67/7 or 11 in Figure 1). Surgical intervention becomes necessary when the Harrington Factor is in excess of the value 5.

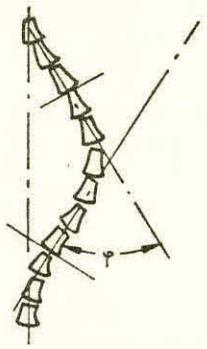

Fig. 1: Determination of the Harrington Factor
Example: Angle of deformation: $\varphi = 67°$
Number of affected vertrebrae: 6
Resulting Harrington Factor: 67/6=11

The following description of the steps of the surgical operation will concern the mechanical aspects of adjustment (distraction) and stabilization of the spinal column.

First the spine is exposed (laid open) from the posterior side, over the entire deformed area.

Then the transverse processes of the uppermost and the lowermost vertebrae affected by the deformation are sufficiently exposed (or laid open, i.e., freed of overlaying tissue) on the concave side of the deformation, so that the clamp jaw (or the hook clamp, clamp jaw) of the "distraction force device" can be placed around these processes. The distraction force application device essentially consists of a threaded rod and two gripping arms. These gripping arms, one stationary and the other adjustable, accept the hook clamps.

When the hook clamps are connected to the distraction device, and placed around the transverse processes, the adjustment nut on the rod is turned, thereby moving the adjustable clamp jaw and pulling apart the two jaws of the clamp until the spinal column is largely corrected. (Fig. 2). With that, the adjustment procedure is complete.

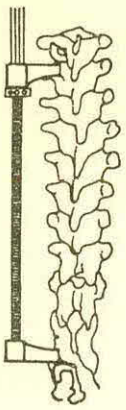

Fig. 2: Spine corrected by application of the distraction force device.

The now essentially adjusted spine remains to be stabilized. This is carried out in four steps:

First step: The compression system is applied to the convex side of the deformation. This system is a highly flexible threaded rod with a number of hooks. The hooks are placed around the transverse processes, in the manner shown in Figure 3, and tightened against the transverse processes with the nuts on the threaded rod. This makes possible a slight additional correction.

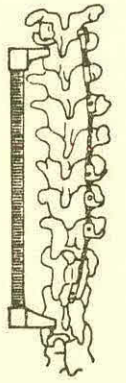

Fig. 3: Compression system, consisting of a flexible threaded rod with hooks attached to it; the hooks are tightened against the transverse processes with the help of the nuts of the threaded flexible rod.

Second Step: Circular vertebral segments of 12.7mm in diameter are then cut out from the vertebrae. This is done with a special oscillating hole saw. The holes thus produced in the body of the vertebrae are then filled with pieces from the pelvic bone, such that a lateral rotation of the vertebrae in relation to each other is prevented by the inserted pieces of bone.

Third Step: The distraction force device is replaced by the distraction rod. The distraction rod is a rod on the upper end of which are lathed helical threads, which taper in a conical fashion. In one of these threads and at the lower end of the distraction rod are hooks (or clamps) which are placed around the transverse processes.

Fourth Step: In the last step of stabilization, long, thin strips (Ea. 2x5 mm in diameter) are peeled from the ilium of the patient and laid on the spine to fuse with it. The interspaces between the ilium strips and the spine are filled with fusion mass.

The mechanical phase of the operation is finished at this stage, and the incision is then closed in this state.

3. ANALYSIS OF THE MECHANICS OF THE CONVENTIONAL METHOD OF TREATMENT, AND THE REQUIREMENTS FOR AN IMPROVED METHOD

The following points in the conventional method warrant criticism:

1. A complete adjustment is not possible with the distraction force. The residual angle of deformation with the conventional method amounts to approximately 15° to 35°.
2. When adjusting small angles of deformation the stress on the spine is very high.
3. Freedom of spine motion in all directions is almost completely prevented with the stabilized spine.
4. The patient must wear a cast and a corset for a very long period (almost a year).
5. The implanted metal components (the distraction rod and compression system) take up too much space.

An improved method must meet the following requirements:

1. For the adjustment and correction procedure, an apparatus must be constructed which achieves maximal correction, while subjecting the spine to minimal stress.
2. For the stabilization, a solution (technique/device) must be found which:
   a. Completely or at least maximally corrects (adjusts) the scoliosis while guaranteeing almost complete mobility of the spine.
   b. is completely functional upon healing of the operation incision, thereby making a cast or corset unnecessary.
   c. entails small sized components of all the stabilizing metal parts.

A concept for such an improved solution to the problem is developed step by step, from the associated mechanical analysis, in sections (3.3) and (3.4).

3.1 A Mechanical Analog Model of the Spine

If one wishes to describe and analyze real-life (i.e., problems that exist in real terms, in this case in the body) problems of skeletal disorders in mechanical terms, it is necessary to replace or simulate the individual components and phases of the real problem by means of basic mechanics elements, in such a manner that the original essential relationships are completely retained. With the laterally deformed spine, this is done by (i) considering the individual vertebrae as short rigid beams and (ii) the ligaments and the discs between the vertebrae as torsion springs, which are assumed to be in an untrained state in the deformed state of the spine.

Figure 4 juxtaposes the real problem and its mechanical replacement model.

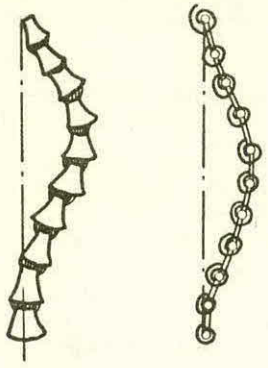

Fig. 4: Mechanical substitute model of the laterally deformed spine. The deformation forces are represented by torsional springs.

3.2 Description of the mechanics of the conventional adjustment procedure

In the description of the mechanics of the adjustment with the conventional distraction force technique and later with the improved concept, the following case data (or study) will be employed for examplifiying the description: (i) the deformed portion consists of 7 vertebrae of equal length (l), (ii) all of the vertebrae are in their deformed or pretreatment configuration, inclined to one another at the (curvature) α_i, (iii), the elasticity of the ligaments simulating torsion springs, will be assumed to follow the Hooke's law, with a rotational stiffness factor D.

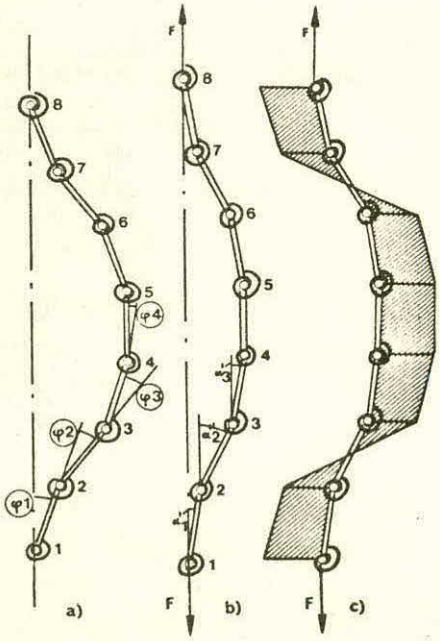

Fig. 5: Deformed spinal column
a) before adjustment
b) upon application of the distraction force
c) progression of bending factor caused by the forces F (qualitative).

Now when the deformed spinal column is pulled apart with the distraction force (F) this corresponds, in the mechanical simulation model, to the tension force F applied at both its ends. Figure 5 (a-c) shows the pretreatment configuration in part a, the adjusted configuration in part b, and the quantitative variation of the bending factors upon application of the distraction force. For clarity, the torsion springs representing the ligamentous and disc connections of the vertebral joints are numbered consecutively in Figure 5 a-c.

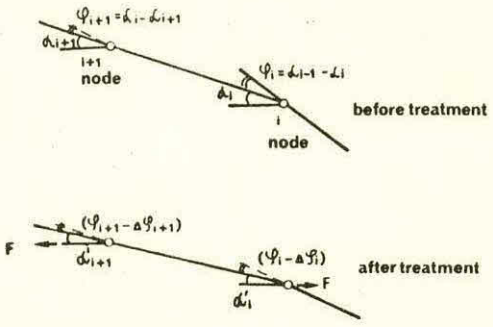

Fig. 6: Pretreatment and distracted configurations of a typical segment joining notes i and i+1.

Figure 6 defines the pretreatment and distracted configurations of a typical segment joining nodes i an i + 1. From geometry consideration, we have (from Figs. 6)

$$\alpha'_i = \alpha'_{i-1} - (\varphi_i - \Delta\varphi_i) = \alpha'_{i-1} - (\alpha_{i-1} - \alpha_i) + \Delta\varphi_i \quad (1)$$

$$\alpha'_{i+1} = \alpha'_i - (\varphi_{i+1} - \Delta\varphi_{i+1}) = \alpha'_i - (\alpha_i - \alpha_{i+1}) + \Delta\varphi_{i+1} \quad (2)$$

From equations (1) and (2), we have

$$(\alpha'_{i+1} - \alpha'_{i-1}) + (\alpha_{i-1} - \alpha_{i+1}) = (\Delta\varphi_{i+1} + \Delta\varphi_i) \quad (3)$$

From equilibrium consideration, we have

$$F \cdot l \cdot \sin\alpha'_i = D(\Delta\varphi_{i+1} + \Delta\varphi_i) \quad (4)$$

$$F = \frac{D(\Delta\varphi_{i+1} + \Delta\varphi_i)}{l \cdot \sin\alpha'_i} \quad (5)$$

It is seen from equations (5) and (3) that
(i) if corrections are small (i.e., if $\Delta\varphi_i$, $\Delta\varphi_{i+1}$ are small) and the distracted configuration (represented by α'_i) is not to be appreciably straightened (i.e., if α'_i is large), then a small value of the distraction force (F) is needed,
(ii) if we want straightening up of the spine (i.e. α'_i small) and even the distracted spinal curvature ($\alpha_{i-1} - \alpha_{i+1}$) small, and if the initial curvature ($\alpha_{i-1} - \alpha_{i+1}$) was appreciable, then a large value of the distraction force (F) is required (according to equations 3 and 5),
(iii) if the spine is not too curved to begin with (i.e., if the initial curvature $\alpha_{i+1} - \alpha_i$ is small), then a small distraction force (F) is required to decrease the curvature of to straighten the spine; this is because the numerator of the term in equation (5), namely $(\alpha'_{i+1} - \alpha'_{i-1}) + (\alpha_{i-1} - \alpha_{i+1})$ will be small relative to the denominator α'_i

The above analysis shows that the distraction technique requires a high value of the distraction force of the initial curvature is high, i.e. of the scoliotic spine is grossly deformed. Hence since the spine can only stand a limited value of F, a complete correction can never be achieved by this technique.

3.3 Description of the mechanics analog of an alternative improved concept for the correction procedure

The essential difference in the improved concept (to be presented here), in comparison with the distraction force concept, is that the forces applied to the spine are no longer limited to the direction of the axis of the body, but are applied transversly to it as well. In the direction of the body axis, a guided force is applied to each end of the deformed spinal segment that the ends can only move in the direction of the body's axis; additionally, as shown schematically in Figure 7 a-c) transverse forces are also applied. Together, the guided axial force and the transverse forces are intended to produce a near uniform rotation of all the vertebral joints.

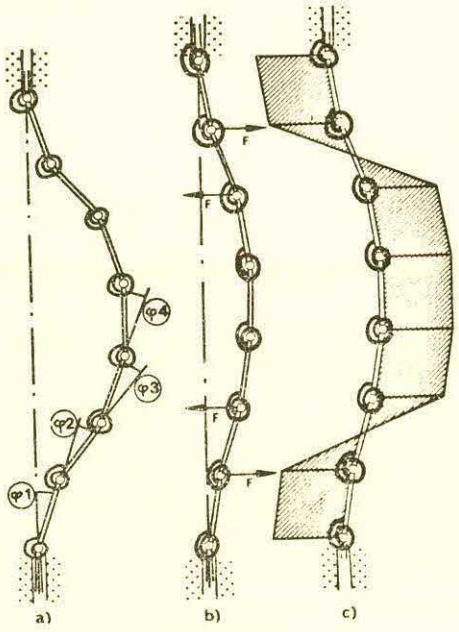

Fig. 7: Deformed spine
a) before adjustment
b) under the stress of the lateral forces F
c) progression of bending factor caused by the forces F (qualitative).

For explanation, reference is again made to the example depicted in Figures (5 and 6). Here too, the starting position (pretreatment configuration) is a deformed spine with the curvature angle φ_i between the individual vertebrae. In this case, a complete adjustment is achieved at the very point when the following relation holds; for the applied transverse force F,

$$F \ell \cos \alpha_i' = D(\Delta\varphi_{i+1} + \Delta\varphi_i) \qquad (6)$$

It is seen, from equations (6) and (3), that it is possible to achieve a major correction of the spine without applying a large force.

Figure 8 illustrates the variation of the non-dimensional force per unit initial curvature $F\ell/D(\alpha_{i-1} - \alpha_{i+1})$ with the final configuration parameter angle α_i'; it is seen that with the new system the magnitude of the applied F can be maintained finitely small even for small α_i'.

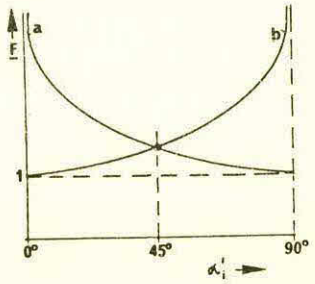

Fig. 8: Variation of the non-dimensional force per unit initial curvature over the final configuration parameter angle
a) Longitudinal forces (Harrington system)
b) Lateral forces (new system)

3.4 Description of the mechanics of the device for the improved concept for stabilization of the corrected spine

We assume that the inter-vertebral disc and ligaments are represented by linear bilateral springs (shown in Figure 9), each at a distance a from the center of the vertebra. Then if C denotes the elasticity of each of the two linear springs, we can relate the equivalent point rotational stiffness D to the composite disc-ligament elasticity C (as depicted in Figure 9):

$$D\varphi_i = C(a\varphi_i)(2a) \qquad (7)$$

$$D = 2Ca^2 \qquad (8)$$

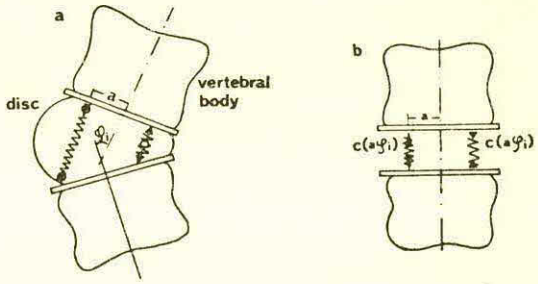

Fig. 9: Representation of the inter-vertebral disc and ligaments by linear bilateral springs in the deformed (a) and corrected states(b)
a) Deformed scoliotic state, where the spring forces are zero.
b) The effect of the spring forces is to produce a deviatory moment on each vertebral body tending to return to the scoliotic state.

This relation expresses the equivalent torsional stiffness of the inter-vertebral joint in terms of the disc elasticity. The equivalent torsional moment $D \cdot \varphi_i$ has a tendency to make the spine return to its scoliotic state.
Hence, the need for spinal stabilization (in the corrected configuration) is to connect the two vertebrae in such a way that the corrected state is maintained. For this purpose, bilateral helical springs (with the spring stiffness λ) are fastened to each of the transverse processes in such a way that their strain (resulting from the prestrain in the disc) will, upon their insertion, infact provide the restoring moment ($D \cdot \varphi_i$, equation 7) to retain the spine in the corrected configuration. (Figure 10).

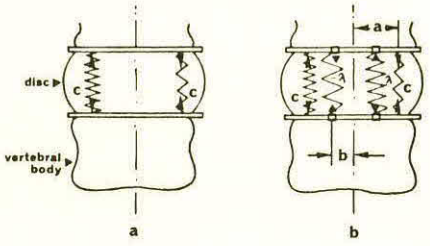

Fig. 10: Schematics of the adjusted state (a) and the spring balanced final state after fixation (b).
(a) Adjustment causes a deviatory moment on the vertebrae (due to the spring resistance force) of amoung $D \cdot \varphi_i$
(b) The helical springs (of the device) mounted at interval b from the central axis compensate for the deviatory moment $D \cdot \varphi_i$ due to the disc spring deformation (shown in part a).

If the undeformed length of the inserted helical spring equals its length in the corrected spinal configuration, then the restoring moment (M_r) exerted by the inserted helical spring

$$M_r = (2b)[\lambda(a\,\varphi_i)]$$

equals the deviatory moment $D \cdot \varphi_i$ caused by the deformation of the disc springs due to the correction, so that

$$D \cdot \varphi_i = 2b\lambda a\,\varphi_i = 2aC a\,\varphi_i \qquad (9)$$

yielding

$$\lambda = C\,\frac{a}{b} \qquad (10)$$

wherein (i) a = disc radius/2, (ii) b is based on the transverse process size.

From equation (10) it can be seen that theoretically an appropriate spring stiffness can be selected. However, additional criteria need be invoked, such as:

1. small construction size
2. kink-proof springs
3. good stabilization of the spine in the adjusted position (even in cases of subsequent change in the ligature resilience)

In comparison to the conventional rigid stabilization (by means of the distraction rod), well dimensioned flexible-dynamic stabilization afforded by the hinged springs possesses the following advantages:

1. nearly complete freedom of motion of forward inclination of the upper body (since a hinged spring can easily be twisted)
2. limited but still sufficient lateral freedom of motion of the upper body
3. almost unlimited rotatability of the upper body on its axis
4. good absorption of shocks (e.g., when jumping or stepping hard on the ground)

4. CONSTRUCTION

According to the mechanical analysis for an improvement of the spinal adjustment procedure, and for its stabilization all necessary mechanical apparatus and devices were developed, including:

1. Adjustment/correction apparatus to be used during the surgical procedure.
2. The spring-loaded stabilization device for individual vertebrae as an implant.

The detailed designs cannot be developed here and, therefore, only a photograph of the modular prototype stabilization device is shown from a lateral view (Figure 11).

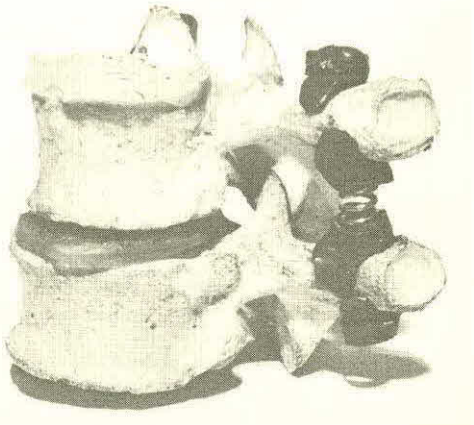

Fig. 11: Lateral view of the prototype modular fixation device, depicting the springs and the fixation mode.

Literature

1) Harrington, P.R. Spine Instrumentation and Fusion Technique Publication of Zimmer Corp., Warsaw, Indiana, U.S.A.

2) Harrington, P.R. Instrumentation in the Treatment for Pathological States of the Human Spine Publication of Zimmer Corp., Warsaw, Indiana, U.S.A.

MULTI-PATIENT ENVIRONMENTAL CONTROL SYSTEM IN A SPINAL INJURIES UNIT: A PRELIMINARY EVALUATION

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A multi-patient environmental control system (ECS) has been developed and installed in a new 30-bed Spinal Injuries Unit. The system provides inbuilt continuous monitoring and control over 14 bedheads from a microprocessor based central control unit. Two interface systems were required to cater for acute and non-acute phases of a tetraplegic's stay in hospital. A review of the first 10 months of clinical operation indicates that tetraplegics needing the system derive very definite benefits from it but that total numbers are well below the maximum ECS capacity.

INTRODUCTION

Electronic environmental control systems are well established in their role of providing the severely physically disabled with a means to control certain functions in their immediate environment (1). The advent of microprocessors has further stimulated the development of more powerful and versatile environmental control equipment (2). However most of these systems have been designed specifically for the home or work situation and are consequently less than ideal when implemented in a hospital environment where a different set of requirements exist.

The building of a new Spinal Injuries Unit provided an excellent opportunity to overcome the inherent disadvantages which would otherwise be met with the continued use of commercially available modular environmental control units. A multi-patient environmental control system (ECS) was developed and in-built as an integral part of the overall Spinal Injuries Unit (3,4).

The $1.5 million 30-bed Unit was commissioned in April 1979, and caters for the southern half of New Zealand's population. It has full facilities for accepting and treating spinal injury patients from the acute phase following their accident through to the medical, surgical, and longterm rehabilitation phases of their stay in hospital.

SYSTEM DESCRIPTION

System Outline

The ECS has been designed around a modified standard communication system, providing each patient with the facilities: nursecall (including intercomm to central nurse station), emergency nursecall, light, radio/television channel and volume, and power outlets. All functions can be operated either directly from the bedhead panel and its handset controller or from one of a variety of possible single switch tetraplegic inputs used in conjunction with an overhead scanning/selection display panel (Fig.1).

Of a total 31 bedhead units, including one in Bio-engineering Laboratory, 15 have been set up for both tetraplegic and paraplegic control. The remainder have been allocated for paraplegic manual control only, but have in-built capacity for easy conversion at some later date if found desirable.

All automated tetraplegic bedheads are monitored and controlled from a single central control system based around a National SC/MP Microprocessor System. The link between this and the bedheads is provided through a multi-wire bus cable with suitable interface electronics at both

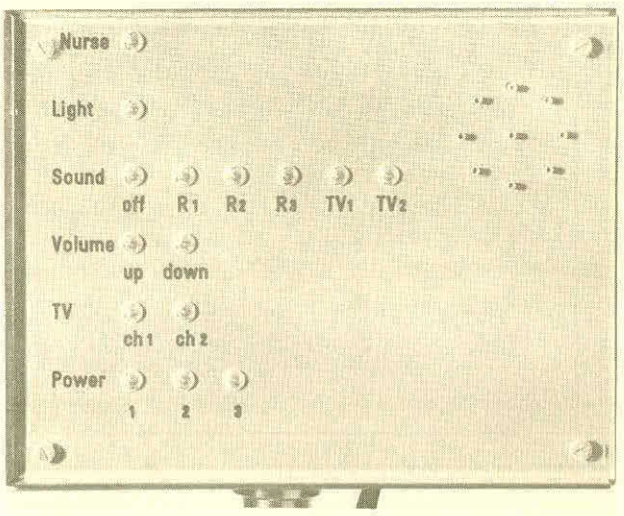

Fig.1. Overhead display panel.

ends. As well as being able to remotely control all bedhead functions, the central control system continually senses all switches on bedhead panels and handsets to ensure that the correct states of patient facilities are indicated on overhead displays.

All tetraplegic bedhead panels have an additional switch allowing 'manual operation only' to be selected which allows remote microprocessor control of facilities to be explicitly disconnected in the event of a control system fault. This system safeguard is further enhanced by a "watchdog" in each bedhead unit which in the event of the microprocessor shutting down for any one of a variety of reasons reverts the bedhead to manual operation. This automatically switches the tetraplegic switch into the nursecall circuit.

The overall system is conveniently subdivided into three sections: patient interfaces, system hardware, system software. These have been described in detail elsewhere (3), with the following descriptions highlighting only the main features.

Patient Interfaces and Operation

Experimentation with a variety of approaches has led to the belief that two types of general interface systems are necessary to optimally provide for most patient/nursing requirement. The two systems correspond closely with the acute and non-acute phases of a patient's medical treatment programme rather than with the actual level of cervical spinal cord lesion.

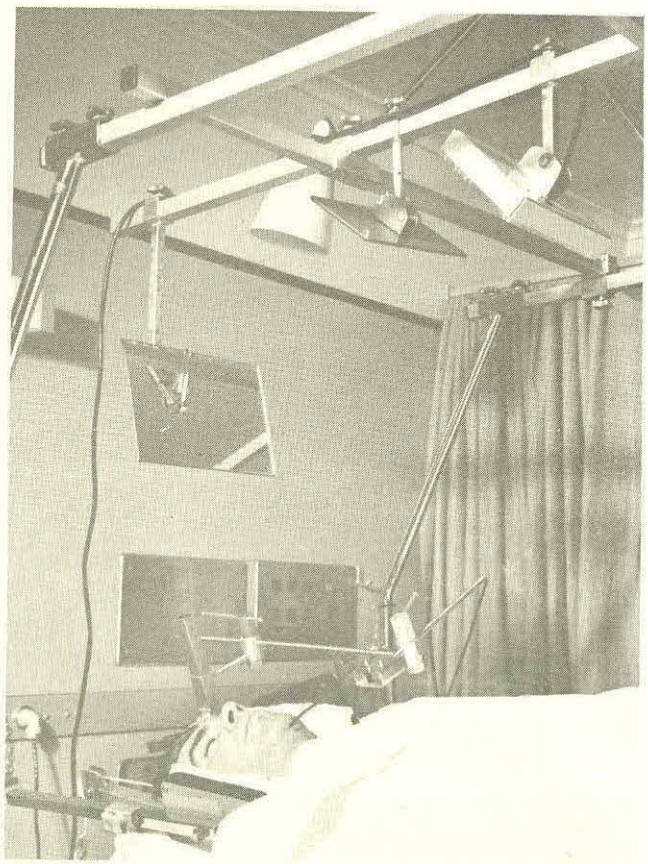

Fig. 2. Chin-switch and overhead sliding support gantry for overhead display and mirrors.

Acute phase - The high level tetraplegic is immobilized by paralysis in all limbs to varying degrees, and by cervical traction for six weeks. During this phase a chin-operated switch has proved the most effective tetraplegic interface (Fig. 2). The assembly mounts upon the turning frame of an electric turning bed and therefore requires little if any adjustment during two hourly turns. It can be quickly swung up out of the patient's way for meals, physiotherapy, etc. An orthopaedic frame and moving gantry is set up above the bed for the fixing of the overhead visual display as well as a double mirror system.

Non-acute phase - The acute phase generally ends after the six weeks of cervical traction is completed. The increased mobility and increasing return of residual function means that most of the previous apparatus is superfluous and a simple pressure switch proves sufficient for most tetraplegics at this stage.

Once the appropriate patient interface is selected and fitted to patient, a two wire cable from the interface microswitch is plugged into a miniature phone socket on the handset controller itself. All services which are available to paraplegics, and some lower level tetraplegics, via the handset and/or bedhead panel then become available to the high level tetraplegic.

Selection of functions specified on the overhead display panel is accomplished by operating the interface switch permitting automatic stepping through functions at a rate of one per 1.6 s. Stepping is indicated by the lighting up of the respective LED and an accompanying audio beep emitted from a small loudspeaker mounted behind the display panel. The stepping mode used is that of starting at the top left corner and moving along each horizontal row in turn. On reaching the desired function, the switch is released and a confirmation period of 1.2 s. is given within which a further activation of the switch will turn the function 'On' if 'Off' and 'Off' if 'On'. If a wrong selection is made the absence of further action will cause the selection system to reset. All functions in the On-state are indicated by their respective LED's remaining on.

An exception to the above is necessary for volume control. After patient has selected Volume Up or Volume Down, he activates switch again, holding it until desired sound level is reached. A second exception is required for activation of the emergency nursecall. Normal activation of the nursecall function results in a 1.5 s long tone burst, and the lighting up of room indicator lights at several annunciator panels strategically located around the Spinal Unit. In this condition the handset loudspeaker/microphone is automatically connected into the nurse-patient intercomm circuit allowing a nurse to converse directly with the patient from the main nurse station console. Where urgent attention is required

required, the emergency nursecall can be activated by effectively calling the nurse twice in quick succession. The microprocessor on detecting this sequence of events proceeds to alternately activate and cancel the nursecall alarm system at a rate of 0.4 Hz. The resulting series of tone bursts are of course very effective in achieving the required urgent attention.

System Hardware

A schematic of the electrical hardware used to implement the multipatient ECS is shown in Fig.3. At the heart of the system is a National SC/MP II Microprocessor. This achieves direct control over all functions at each bedhead by means of a 150 m screened 11-wire bus running under the floor of the Spinal Unit wards with feeders to each of the bedheads.

The 9-bit Ward Address Bus can be subdivided into two sections:

(i) 6-bit Bedhead Address Bus allowing up to 64 bedheads to be selected.

(ii) 2-bit Bedhead Function Bus allowing the 4 functional units of each bedhead to be selected - Bedhead Sense, Bedhead Action, Overhead Display, Volume Control.

The microprocessor selects the relevant bedhead unit and function requiring servicing by having the corresponding 8-bit Ward Address number on the 8-bit data bus and enabling this into the Bus Interface Buffer Latch by placing 1FFF on its own 16-bit address bus. This activates the desired Bedhead Receiver permitting serial data to be read out to (SOUT) or read in from (SIN) Bedhead Controller registers as desired. All serial data is transmitted at a rate of 6700 baud by means of a software generated clock (SC/MP Flag FO set to 1 and 0 in successive operations). This clock rate is maximal in terms of allowing sufficient time for successive SIN or SOUT bits to settle on the ward bus to achieve a zero transmission error rate.

Each complete bedhead controller physically comprises a bedhead and an adjacent bedhead interface unit. The interface unit contains the bus Bedhead Receiver and Volume Control boards while the bedhead unit contains the Bedhead Sense and Bedhead Action functional units as well as electrical interfacing to standard bedhead facilities.

System Software

The microprocessor program is based around a ward service loop in which each of the 15 automated bedheads is monitored and/or serviced in turn at a rate of approximately 6.7 times per second. As each bedhead is scanned, data is read from its Sense Register and compared with the stored previous version to determine whether any manual switch changes, including that from the patient interface, have occurred. If no change is detected, the microprocessor pulses the bedhead's watchdog and moves onto the next bedhead. If a change is detected, various subroutines are called to initiate appropriate action including updating of the overhead visual display.

CLINICAL EVALUATION

During the first ten months of clinical operation in the Spinal Injuries Unit, the superior features of the inbuilt multi-patient ECS have been well proven when compared with the prevously used single-patient modular alternative. These features are: virtual elimination of a proliferation of various appliances which can cause an unsafe and unhygienic nurse/patient environment; elimination of difficulties foreseen in interfacing to standard manually operated hospital communication systems; minimisation of setting up requirements.

The tetraplegics who made use of the ECS facilities were found to benefit in several ways:

(i) Reduction of helplessness-induced anxiety by being able to operate the nursecall system.

(ii) Regaining of some independence and confidence by being able to control a variety of functions from a single switch input.

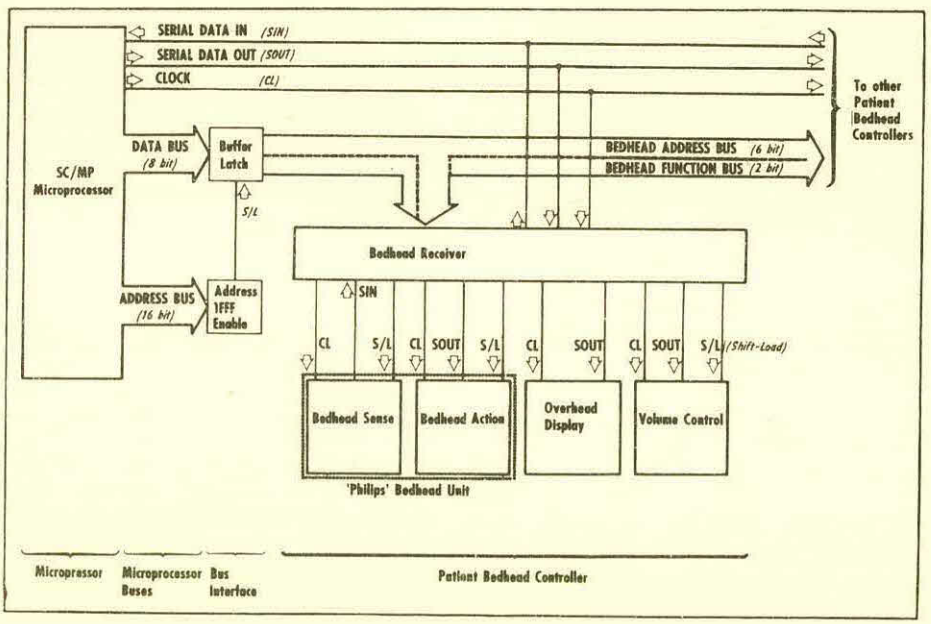

Fig.3 Schematic of ECS hardware.

(iii) Increase in motivation for self-rehabilitation with the realization that there are other ways in which certain activites can be carried out.

(iv) Relief from the boredom of absolutely nothing to do by being able to practice at optimal control of the ECS.

A total of 17 acute tetraplegics were admitted to the Unit during the initial clinical period of mid-April 1979 to mid-February 1980 (Table 1). This group consisted of 15 males and 2 females, a mean age of 23 years (range = 18-45 years), and an etiological distribution of 5 rugby, 5 fall, 4 motor-cycle, 2 motor-vehicle, and 1 diving accident. Of the total 17 acute tetraplegic admissions, 9 had the ECS facilities set up for them including 3 with sufficient arm function in the acute stage to bypass a need for the more elaborate chin-switch interface. Further use of the ECS was also made by 3 high-level tetraplegics re-admitted for routine checkups.

Eight of the 17 acute tetraplegic admissions did not utilize the ECS. They had definite upper limb involvement but their lesions were either incomplete or sufficiently low level that they were capable of operating the standard handset controller without having to resort to a special interface.

Overall several general conclusions have been reached regarding patient usage of the ECS:

1. The benefits gained by tetraplegics from the ECS are most evident but are largely limited to the acute phase of treatment. Once out of cervical traction, usually about six weeks after admission, the need for ECS facilities drops significantly. This is primarily due to increased mobility in terms of upright position in bed or wheel-chair which helps effective arm function through a greater enabling of gravity return. The frequent gradual return of an extra spinal level of function following resolution of oedema in the spinal cord superior to the lesion also contributes to the decreased need for ECS.
2. It is considered that as a general rule if a tetraplegic has good C6 function - that is, good wrist extension - he should have no eventual need for ECS facilities, either in the Unit or after discharge. However they may benefit from the ECS if restricted to bed for any reason and if they have difficulty with forearm extension.
3. From a nursing point of view, the ECS does have potential as a time saver where staff shortages are evident. It is recognized however that the workings of the chin-switch interface are not fully appreciated by many of the nurzing staff which often leads to less than optimal use of system.
4. Overall the ECS is far from being utilized to its maximum capacity but most of this can be attributed to a significant drop in recent numbers of high level spinal injury admissions. This gratifying situation appears to be at least partly due to a substantial decline in serious road accidents in New Zealand over the last year or so. Increased enforcement of the 80 kph open-road speed limit and anti-drinking driver campaigns are obviously significant factors leading to this trend.

ECS Usage Category	Orthopaedic Injury[1]	Patient Number	Sub-Total
Chin & Pressure Switches-	C4/5	1	
	C5 #	1	
	C5&6 #	1	
	C5/6	2	
	C6/7	1	6
Pressure switch only-	C5/6	1	
	C6/7	1	
	CCC	1	3
Not required-	C5 #	1	
	C6 #	2	
	C6&7 #	1	
	C7 #	1	
	CCC	3	8
Total acute tetraplegic admissions =			17

TABLE 1.
Analysis of acute tetraplegic admissions and ECS usage during first 10 months of operation in Spinal Injuries Unit.

(C = cervical; / = dislocation; # = fracture; CCC = cervical central cord)

[1] Although the orthopaedic diagnosis is usually quite definitive, it can only be used as a general guide to the actual acute and long-term levels of neurological function.

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MICRODEC: A MICROPROCESSOR-BASED ENVIRONMENTAL CONTROL SYSTEM

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ABSTRACT

A microprocessor-based environmental control system for severely disabled people has been developed which can be operated by two switches and can control up to sixteen devices and a single telephone. Coded signals are transmitted over the power mains to receiver modules at each peripheral device. Commercially available receivers are utilized to simplify installation, increase portability, and decrease total system cost. Microprocessor design was incorporated to enhance flexibility and to provide complete user control over all functions of the system.

INTRODUCTION

This paper describes a microprocessor-based environmental control system for severely disabled people that has been developed in our laboratory. The system can be operated with any arrangement of two switches, touch, pressure or pneumatic, and can control up to sixteen peripheral devices and a single telephone. Remote operation via a radio link is also possible.

Device control is effected by transmitting coded signals over the power mains to receiver modules located at each peripheral device. The receiver modules decode the signals and selectively control peripheral operation. The concept of "remote" operation by transmission of signals over the power mains is not new, having been used in intercom systems for some years. We first became interested in this concept of peripheral control several years ago (Doubler et al [1]), describing it as "distributed" control. Based on this concept, we have selected the acronym "MicroDEC" (Microprocessor-based Distributed Environmental Control) to specify the system described here.

DESIGN CONCEPTS

The present form of the MicroDEC system is the result of several fundamental design criteria. From our previous work with environmental control systems, we knew that direct connection to peripheral devices by wires had many disadvantages. These include (1) difficulty of installation, (2) unsightly, cluttered appearance and (3) decreased portability and flexibility. Distributed control eliminates wires, thus simplifying installation and making movement of a peripheral device from one location to another as simple as unplugging it and its receiver from one wall socket and plugging them into another. Because of its modular nature, distributed control also has the potential to be more cost effective since users could purchase just the right number of receiver modules to meet their requirements and install them without assistance. If desired, they could start with just a few, reducing initial system cost, and add more later as their needs increase or as funds become available.

It was felt that the utilization of microprocessor technology held the best promise for the development of a practical system that effectively meets the needs of a large group of disabled people. Choosing a microprocessor as the basic control element in the system simplified hardware design and reduced expense by placing many of the system functions under software control. This approach also makes the system more resistant to hardware obsolesence. Implementing software control allowed us to include several user programmable features which enable individuals to easily configure the system to meet their specific and changing needs.

Finally, in designing the MicroDEC system we tried to use commercially available components whenever possible. Because the market for assistive devices for severely disabled people is relatively small, it is difficult to attract larger manufacturers who possess the facilities and technical expertise to develop, mass produce and service low-cost, reliable devices. Thus the construction and distribution of assistive devices is generally performed by small specialized companies. Utilizing components that are already in distribution significantly reduces manufacturing costs and simplifies construction.

SYSTEM HARDWARE

The basic MicroDEC system consists of the MicroDEC control unit with appropriate switch inputs (touch, pressure, pneumatic, etc.) and several receiver modules. The receiver modules used are commercially available products, part of the BSR X-10 Home Control System. These components are shown in Fig. 1 together with the BSR system command console.

The core of the MicroDEC control unit is a

Fig. 1 MicroDEC control unit, BSR receiver modules and BSR system command console.

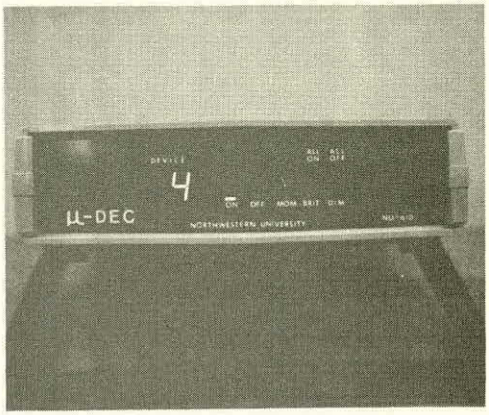

Fig. 2 Close-up of MicroDEC control unit display.

custom designed single-board microcomputer based on the Motorola M6802 microprocessor. The M6802 is a version of the 6800 microprocessor having 128 bytes of on-board RAM and an on-board clock. The program controlling system operation is stored on a single 2K EPROM. Two parallel ports are used, one for transmitting information to an eight character, seven-segment LED display, and the other for user input and device and telephone control output. A commercially available telephone coupler (FCC approved) is utilized in the system, allowing the use of any telephone with a standard connector. Accommodations have been made for the use of a commercially available radio transmitter/receiver system when remote operation is desired (e.g. from a wheelchair or bed). The receiver is mounted within the control unit.

A custom designed circuit was developed for implementing device control by transmission of coded signals over the power mains. This circuit and the software driver provide digitally modulated 120kHz control signals that are directly compatible with the BSR X-10 Home Control System receiver modules. First available at the end of 1978, the BSR system itself consists of a push-button activated command console and three different types of receiver modules: an appliance module for controlling 110 volt electrical appliances (e.g. radio, fan) a lamp module providing on/off and dimmer control of incandescent lamps and a wall switch module providing similar lighting control. We have developed a switching unit that makes it possible to control self-powered devices, such as a page turner, and a control unit that allows operation of multi-input devices, such as a hospital bed or TV. Both of these units are used in conjunction with a BSR appliance module.

The MicroDEC control unit is designed to augment the BSR system without interfering with it in any way. No modification of any BSR component is necessary; the MicroDEC control unit acts solely as an interface for a disabled person. It does not restrict the use of controlled devices by other people, who can simply use the BSR command console. The BSR command console is not necessary for MicroDEC operation although it is recommended for use (by attendants, etc.) in conjunction with the MicroDEC unit.

SYSTEM OPERATION

The MicroDEC system utilizes a scanning method of operation requiring two switch closure input signals, one for advancing the device scan and the other for selecting or activating the specified control function. There are three basic functional modes of operation: (1) device control, (2) telephone control and (3) user programming.

A picture of the system display by which the user monitors system operation is shown in Fig. 2. Each peripheral device being controlled is assigned a number that corresponds to the thumbwheel switch setting of its receiver module. The numbers are displayed sequentially in the left-most position of the system display as the user scans through the possible selections. When the user selects the number of the peripheral to be activated, a cursor appears that can be advanced from left to right across the display, pointing to the possible control functions. Selecting a control function causes the appropriate control signal to be transmitted, activating the desired peripheral advice. Dimmer control of lamps and an all-on/all-off feature (capabilities included in the BSR system) are also provided.

In addition to the peripheral device numbers, selections are presented to the user in the basic scanning sequence that allow branching to the telephone control and user programming modes. These selections appear on the display as the seven-segment equivalents of the words "CALL" and "DIAL" (telephone control) and "CODE" (user programming). The telephone control selections, however, are presented in the basic scanning sequence only if the user has programmed their inclusion. When specified for display, the

telephone selections are always presented after device number 2 in the basic scanning sequence.

By selecting CALL, the user can enter and save in system memory a telephone number containing a maximum of 13 digits. Only one telephone number can be saved at a time. When CALL is first selected, the previously stored number is displayed, shifted right to left across the display. If a new number is desired, activating the appropriate switch clears the old number and begins an automatic scanning of digits in the right-most position of the display. New selections are stored in system memory and displayed. The user can halt the number building at any time, at which point the entire new number is displayed momentarily before the system returns to the basic scanning sequence.

To dial the stored number or answer an incoming call, the user selects DIAL, which is presented after CALL in the scanning sequence. Selecting DIAL opens the telephone line. By activating the proper switch, the stored number can then be dialed or the user can return to the device control mode with the telephone line closed (on-hook) or open (off-hook).

The user programming mode of operation can be entered by activating the CODE selection, which is presented after the final device number in the scanning sequence. In the user programming mode, five options coded as letters of the alphabet (A through E) are presented in scanning sequence in the left-most position of the display. Option A allows an audio tone accompanying device and function selections to be turned on or off. Option B allows the user to set the automatic scan when building telephone numbers. With option C the number of devices included in the basic scanning sequence can be set to any number from 1 through 16. Option D allows the addition or removal of the telephone selections from the scanning sequence. Finally, option E determines the mode of operation of each peripheral device in the scanning sequence (i.e. momentary or on/off control).

CONCLUSIONS

We are presently finalizing the MicroDEC system hardware. Present plans call for the development of a system that can be controlled using a single switch input. No hardware changes will be needed to implement this new system, only new software.

Three prototype systems have been placed in the homes of severely disabled persons and several more systems are being constructed for field evaluation. We hope that the MicroDEC system will become a commercially available system in the near future. We feel it has the potential to be a practical and cost effective means of significantly improving the ability of disabled people to control devices and appliances in their environment.

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C2E2: A MICRO-COMPUTER SYSTEM TO AID THE SEVERELY PHYSICALLY DISABLED IN ACTIVITIES OF DAILY LIVING

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This paper describes the conceptualization, design and development of a micro-processor based device to be used by the severely physically disabled for purposes of environmental control, communications, education and entertainment. The "C2E2 SYSTEM," developed by the Bio-Medical Engineering Group at the University of Alabama in Birmingham's Medical Rehabilitation Research and Training Center, (RT-19), is a cost-effective alternative to the majority of commercially available environmental control systems for the physically handicapped which are, without exception, of limited flexibility and considerable expense.

Due to improved medical-rehabilitation practices, there has been an ironic increase in the number of persons with severe disabilities in recent years; more people are surviving accidents and disease. Consequently, an ever-increasing number of persons with functional disabilities can look forward to "near normal" life spans, but must face abounding obstacles to daily living.

The ability to perform various activities of daily living, and the associated quality of life experienced by persons with little or no muscular control, is enhanced, somewhat, by the availability of assistive devices which enable them to "interface" with their physical and social environment. To an extent, some of the specific needs of the physically disabled have been met by devices such as electrically powered wheelchairs, page turners, "hands-free" telephones, specialized eating utensils and writing equipment, some of which may be activated and/or controlled by movement of the head, teeth, eyes and respired air. Along similar lines, a few coordinated interface systems have been developed, but they have generally been single or limited purpose units (such as communication substitutes or environmental control systems) which enable the severely physically disabled party to operate a limited number of household appliances or in some applications simple business machines. Experience has shown, however, that such equipment is expensive, limited in capability and utility as well as frequently being unreliable (1,2,3,4,5,6,7,8).

In recent years, rapid strides achieved in the area of computer technology have resulted in significant changes occurring in this emerging field which are now playing an increasingly important role in the lives of many handicapped individuals (6,9).

To fully understand the nature of the approach we have employed, a distinction between "small computer" hardware terminology must be made. The two major terms are mini-computer and micro-computer.

A "mini-computer" is a device/system of relatively small size. "Minis" are frequently found in research laboratories and dedicated to a specific task, quite often being supported by a staff of one or two personnel. Mini-computing systems range in price from $10,000 to $100,000. Numerous attempts to utilize mini-computers as aids to the physically disabled have been characterized by the necessary development of sophisticated and costly custom-built interface mechanisms.

A "micro-computer" is appreciably smaller in size than the mini-computer counterpart. Typically, a micro-computer will occupy a single printed circuit board and is sometimes identified by the term "SBC," meaning Single Board Computer. These devices are highly flexible in their applications, quite portable and moderately priced, costing from several hundred to a few thousand dollars.

Unlike mini-computer software, micro-computer software is easily supported by a single individual. In addition, the extraordinary popularity of the micro-computer has resulted in a highly competitive market situation which has witnessed (1) improvements in the quality and capability of commercially available units, (2) significant reduction in hardware and software costs and (3) the development of a national network of service/maintenance organizations.

The advantages of the micro-computer - for application in the realm of the physically disabled - is readily apparent. Low cost, high quality "off-the-shelf" hardware and software are quickly gaining popularity over one-of-a-kind, custom-built devices.

Finally, a description of micro-computer systems would be partially incomplete without specification of its most basic component - the micro-processor - which is a single "chip" controlling software and executing instructions dictated by that software.

C2E2:

In 1979, the Bio-Medical Engineering Group of the Medical Rehabilitation Research and Training Center (RT-19) at the University of Alabama

in Birmingham received a modest grant enabling them to pursue a micro-computer project with the following objectives:

(1) To develop and demonstrate the usefulness and economic feasibility of micro-computer system applications for purposes of communication, environmental control, computer aided instruction (CAI) and entertainment for persons with severely disabling conditions.

(2) To disseminate system information and otherwise promote, encourage and stimulate further applications and expanded utility via the development of additional software and hardware.

(3) To solicit, evaluate and disseminate software and hardware designs created expressly for use by the severely physically disabled.

The first objective and associated tasks addressed the design, construction and demonstration of a low-cost multi-purpose micro-computer based system to be used by a severely physically disabled population.

Achievement of the objective required an accurate assessment of target population needs, interests and desires followed by the development of an appropriate system capable of meeting these requisites.

Examples of applications we identified included (1) the composition and production of written communication, (2) the storage and retrieval of information, (3) the operation of common household electrical appliances and (4) the provision of computer assisted instruction for personal and vocational purposes.

The second objective was designed to assure comprehensive dissemination of information about the project. The strategy we selected required activity at two levels: (1) the rehabilitation community, potential users, independent micro-computer hobbyists and (2) manufacturers of micro-computers and peripheral equipment. An intensive effort has been made to demonstrate and report on the model system at both levels. Obviously, the ultimate value of the research endeavor is dependent, in large part, upon the degree of awareness and interest the developers are able to create in the user and manufacturing communities.

The third objective was designed to stimulate enhancement and refinement of the model system by providing an effective and efficient medium of information exchange among a widely diverse group of users, hobbyists and manufacturers. In part, our interest in participating in this Conference is reflected in Objectives 2 and 3.

Suffice it to say, after defining the overall objectives, careful consideration was given to establishing specific criteria for the C2E2 System. Ultimately, the following specifications were set out:

(1) The system must be software programmable in a widely used computer language to accomplish multiple functions without attendant hardware changes;

(2) The system must be composed of readily available plug compatible components and backed by a dependable and easily accessible service organization. Custom hardware must be held to a strict minimum;

(3) The system must be capable of accepting varied inputs, including voice commands, switch closure, and joystick for system control.

The current system utilizes an Apple II Computer, which satisfied the array of performance criteria we established. The only piece of custom hardware is a "black box" containing a small logic-card and relays to control the television and telephone system. For convenience the headset, joystick and switch inputs were also relocated in the "black box" which is of low cost and design simple enough to be well within the capabilities of the novice electrophile (Figure 1).

The present system performs the following tasks: (1) answers the telephone; (2) dials the telephone from a stored directory; (3) functionally operates a television receiver; (4) turns appliances on/off; (5) prepares typewritten material using an associated video display for editing and correcting prior to hard copying; (6) provides entertainment through challenging game programs; (7) educates and instructs the user on a variety of topics such as math, vocabulary, etc.; (8) enables the user to compose and generate; and (9) provides access to other computer systems via the switched-telephone network (Figure 2).

System control is achieved through a number of selection menus. The master control menu is "brought up" from a cold start situation without operator intervention. Menu selection is accomplished using operator directed voice input. If the user is non-vocal, command sequences may be initiated by switch closures at an appropriate time, word-board-style, by locating a "cursor" or command point with the joystick controller.

Video output is "composite video" allowing placement of additional monitors to check system activities if desired. A "hard wired" television screen provides continuous monitoring of system status and operation; an additional remote control color television set operates independently of the system monitor for entertainment or provision of color displays and graphics at the discretion of the operator. Audible system alarms and the generation of computer-synthesized music is provided through a speaker. A more advanced music generation system is being developed using a commercially available synthesizer which will plug into one of the expansion slots.

The present model system operates in a fast but somewhat limited "integer" BASIC language. An advanced system is in preparation which will use a more powerful "floating point" BASIC. Sophisticated languages such as FORTRAN, PASCAL and COBOL are available, but BASIC is the easiest language for "new users."

The Apple II was selected for the C2E2 for a variety of reasons: (1) it is in national distribution because of its popularity with computer hobbyists; (2) it is backed by a reliable service organization; (3) it is programmable in BASIC; (4) it has excellent software capabilities from many varied sources; (5) it is readily expandable

"on-board" both in memory size and I/O capability; (6) it possesses numerous built-in features which are normally available only as "add-ons" in other systems we evaluated and considered; (7) it is well designed from a human engineering standpoint coupled with being attractive, lightweight and rugged; and (8) it is reasonably priced considering its many standard features. There are other micro-computer systems which would serve equally well, however, our group selected the Apple II as the "best-buy" on a "dollar-for-dollar" basis.

When first proposed, this project was criticized as "being grossly underestimated in scale and feasibility" by reviewers. Happily, we can report this was not the case. In fact, the initial model system was developed at considerably less expense than our original, modest projections. We were, as a result of the cost-savings realized, able to markedly enhance and improve the system without additional grant support from the funding agency.

To summarize, the commitment of stable manufacturing firms to the production of home microcomputers, along with interest and support from computer hobbyists, indicates there will be a considerable amount of independent effort to further expand the possible applications of C2E2 along with development of imaginative software designed specifically for use by physically disabled persons.

C2E2 will enable severely physically handicapped persons to address many of their physical needs and perform numerous activities of daily living with minimal assistance from family members or attendants. Having control over one's environment, even in a limited manner, promotes a sense of independence, self-esteem and well-being. Thus, utilization of this and similar devices for environmental control, communication, entertainment and education will address numerous physical and psychological needs of severely disabled individuals.

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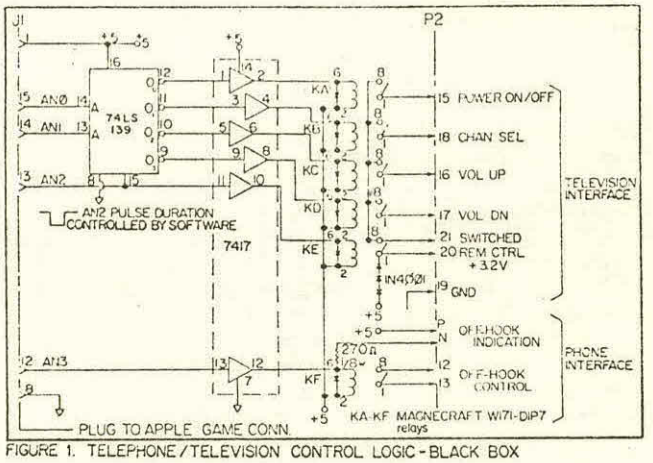

FIGURE 1. TELEPHONE/TELEVISION CONTROL LOGIC - BLACK BOX

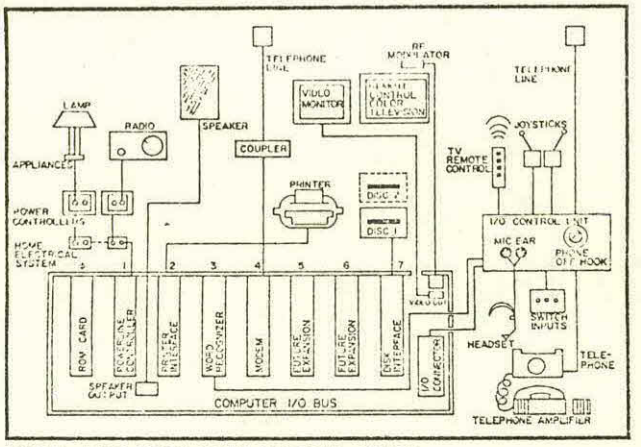

FIGURE 2. BLOCK DIAGRAM OF THE C2E2 MODEL SYSTEM

TOTAL ARM PROSTHESIS DRIVEN BY 12 MICRO-MOTORS, POCKETABLE MICROCOMPUTER AND VOICE AND LOOK-SIGHT MICROCOMMANDING SYSTEM

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Abstract A control system of electrical prosthesis of total arm has developed for victims of a shoulder amputation or Thalidomide tragedy. We introduced a microcomputer for co-operative controlled electrical prosthesis of total arm and the development of practical electrical prosthesis of total arm was made miniaturizing the control system to pocketable size.
As a result, this electrical prosthesis involves 12 motors, which enable prearrange degrees of freedom required by handicapped persons to make easily daily movement correctly and safely. As the command devices which reduce operative work of handicapped persons, we utilized a voice controller with the vocal humming signal and a look-sight controller which makes the automatical movements of prosthesis to the visual direction.

Introduction

Our research group, supported financially by The Science and Technical Agency and the Ministry of Health, has been working for 12 years to develop a microcomputer controlled total arm prosthesis system that will permit persons who have lost their arms through disease or accident to perform basic daily movements, controlling the arm by humming. We have finished the technical development and have begun field tests with handicapped persons.

The results so far achieved in this research are;

i) The design and machining of all mechanical parts has been completed.
ii) The testing of all mechanical parts has been completed.
iii) Based on a movement analysis of the human arm, a general numerical formula has been developed, which allows a microcomputer to control the 12 motors in the artificial arm to simulate the movements of human arm.
iv) A new pocketable microcomputer, voice command device and look-sight command device have been designed and fabricated.
v) To permit digital servo-control of the arm by the microcomputer, a micromotor unit (composed of an optical shaft encoder, a micro-coreless motor, and a reduction gear box) was developed.
vi) The formulas (above iii) were implemented in software to control arm movements.

Our research group includes both engineering and medical personel. The engineering part includes researchers from 5 universities of engineering faculties and about 30 industrial companies, while the medical researchers come from 4 universities medical departments and 5 rehabilitation centers. In all, about 100 researchers are participating in this project.

1. The Electrical Prosthesis of Total Arm with 12 Motors.

The electrical prosthesis of total arm was developed following these principal conditions:

A. 12 degrees of freedom utilized to the electrical prosthesis.

3 in the shoulder as horizontal adduction -abduction, flexion-extension and external-internal rotation, 2 in the elbow as flexion-extension, forearm pro-supination, 2 in the wrist extension-flexion, radial--flexion and ulnar-extension and 5 in all fingers to obtain the degrees of freedom required by the prosthesis to handle essential daily tasks.

B. Weight reduction.

This was accomplished by using carbon fiber chiefly as the parts of structure and duralumin in the torque transmission parts.
The proportion of motor and gear-box was designed to be less than 60 %---the total weight of the prosthesis.
The interface circuits were lightened by hybrid formation using microcomputer in order to manage the controls of the 12 motors.
As a result, the physically handicapped persons became able to carry on their equipments.

C. Reduction of power consumption.

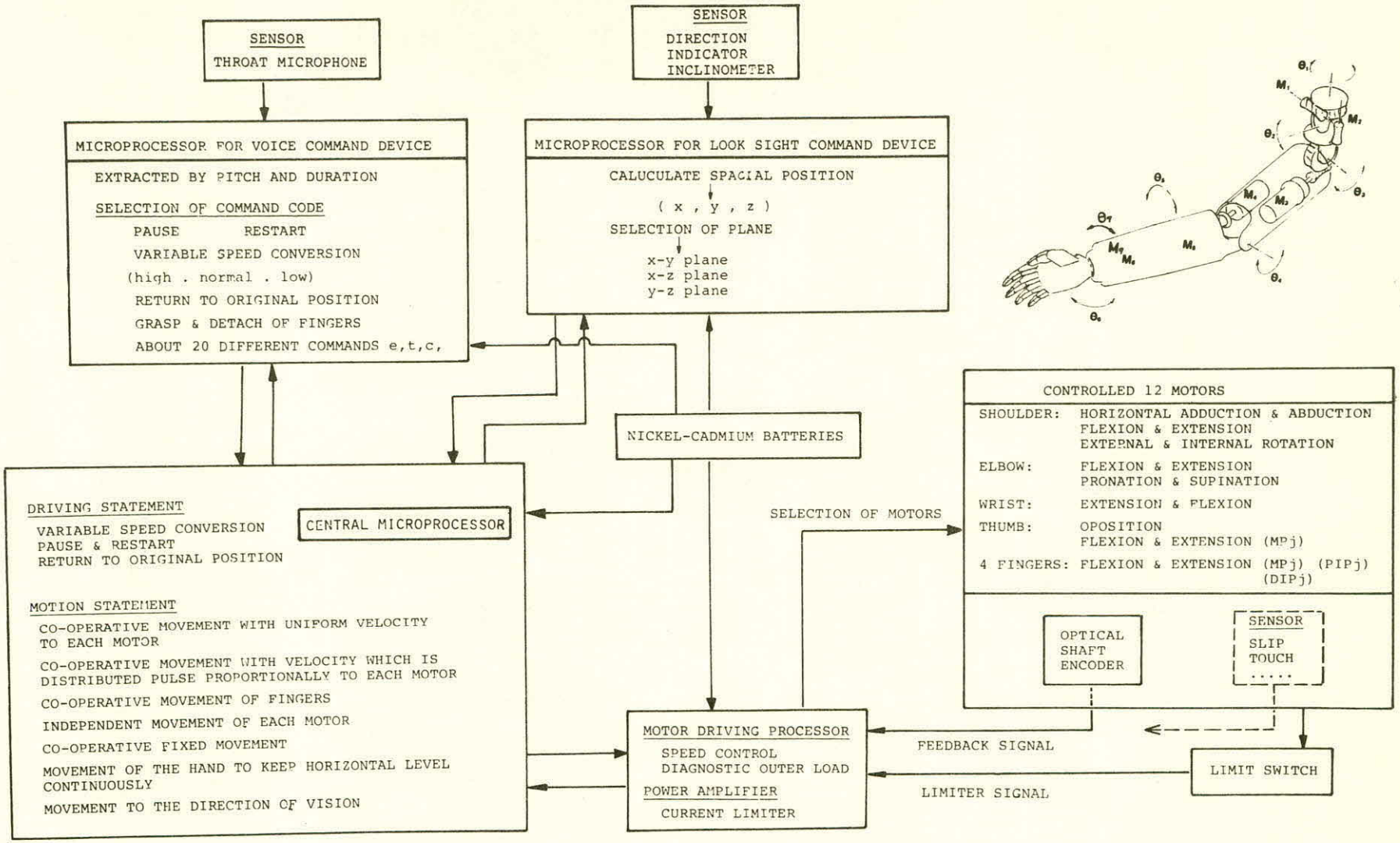

Fig. 1. Block diagramme of the system which enables to transmit the signals to operate the electrical total arm prosthesis.

i) The driving motor power source.

The reduction of power consumption and the continuation of motor's life were insured by introducing pulse drive system.

ii) Computer and interface circuits.

Interface circuits are composed of C-MOS IC 8085 family of microcomputer was used for computer hardware. However, the power source will be reduced by changing to the microcomputer of C-MOS type. Ni-Cd batteries are used as the power sources.

D. Simple operation.

One of the difficult problems in designing and operating multi-freedom prosthesis is to avoid imposing a burden on the persons using it. Therefore, the operation mode uses both the voice controller and look-sight controller which act according to the movements of visual direction.

E. Safety and reliability.

Since the multi-freedom prosthesis has a movement zone simulating the natural arm, in some cases, the prosthesis can become harmful to its operater. From this point,

i) A limit switch for the limitation of movements of each freedom was provided.
ii) A current limiter in power amplifier was provided.
iii) A function which stops directly before the limit of movement was added to the software programming of movement.
iv) Also should the output pulses from shaft-encoder be not feedback by outer load on exceeding the threshold the total arm returns to its original position.
v) The improvement of electrical reliability by hybrid formation of computer and interface circuits.
vi) The controller of prosthesis can be used in another ways. When the handicaped persons are regenerated, they can use it by their microcomputers as a controller of the machine tool if they work to handle the numerical controlled machine tool. And then, the man-machine system between the electrical prostheisis and numerical controlled machine tool can be formed.

2. Total System.

Fig. 1 shows the block diagram of the system which enables the signal to be transmitted to total system. In addition, a microcomputer was developed as the control system based on the conditions of the facility of multi-freedom control as well as system reliability and system conservation. There are three classes of instruments for the items and measuring the microcomputer system as table 1.

Table 1

I. Driving of micromotor; limit of movement zone, limit of allowable load in the hand artificial sensing of grasping movement are indluded. And position, speed current limiter, position sensor, pressure sensor slip sensor and touch sensor are setted.

II. Controlling of movements; As a command devices, voice and look-sight controller and mechanical switch are equipped. Controlling items by voice controller are; start, stop, pause, restart, right, left, forward, back, under the left, under the right, over the left, over the right, central position, return to original position, keeping hand horizontaly, fixed movement (eating, drinking etc. as programmed), grasp, 3 stages change of speed so on.
Controlling by looksight controller; directions and angles.
Controlling by mechanical switch; Manual position control.

III. Operation of central microprocessor.

Manual control:	transfer to space lattice
Motion control:	co-oprative movement depending on the pulse distributed proportionally to the movement.
Position control: (trajectory)	movements on 3 axes movement to the direction of vision co-operative programmed movements movement of the hand to keep horizontal level continuously
Position control:	Independent movement of each motor return to original position
Interruption:	command devices can interrupt overtravel limit overflow pulse to overload initial reset (power supply) initial reset (I/O port)
Speed control:	slow-up, slow-down, 10 kinds of valiable velocities accepted

3. Specification of Total Arm Prosthesis.

length 673mm
upper arm 273mm forearm 209mm
hand 191mm

Movement of each articulation:

Shoulder:	horizontal adduction	-150°
	" abduction	90°
	flexion	110°
	extension	-15°
	external rotation	90°
	internal rotation	-110°
Elbow:	flexion-extension	0--135°
Forearm:	pronation	120°
	supination	-120°

Wrist:	radial-flexion	90°
	ulnar-flexion	-90°
Thumb:	oposition	0--90°
	flexion-extension (MPj)	0--70°
4 fingers:	flexion-extension (MPj,PIPj,DIPj)	0--80°

Weight:

Prosthesis	2.2kg
Socket (millwaukee type)	0.9kg
Dummy arm (include batteries and amplifiers)	2.3kg
Control device	0.5kg

(Note) The original hand part used in this arm prosthesis was made by the Centre d'Etude et de Recherche des Anciens Combattants de Paris in France. We have improved this original hand to a practical level by several improvements on the mechanisms with the collaboration of Dr Cassagne, Director of the Center.

4. Conclusion.

The field tests are being performed for two amelias by thalidomide tragedy (man 18 age, woman 20 age), the amputator of shoulders (man 50 age) who engaged in power electronics and suffered from a disaster, and the amputator of upper limbs (50 age). As the common ADL, the movements of eating and drinking are chiefly examined. Consequently, we obtained the following results:

i) The standard size of prosthesis was designed.
ii) The design and machining of mechanical parts of prosthesis have been completed.
iii) The system can be organized according to the object for forearm or upper arm.
iv) The fixed memory can memorize new movements according to the demand, excute partial changes and adapt itself to environment of handicaped persons.

Presently, further research is being conducted to develop the arm to a more practical stage making some minor improvements and developing related software in accordance with the result of tests by handicaped persons over the next few years.

Acknowledgements

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The authors wish also thank all the members participating on this special project research.

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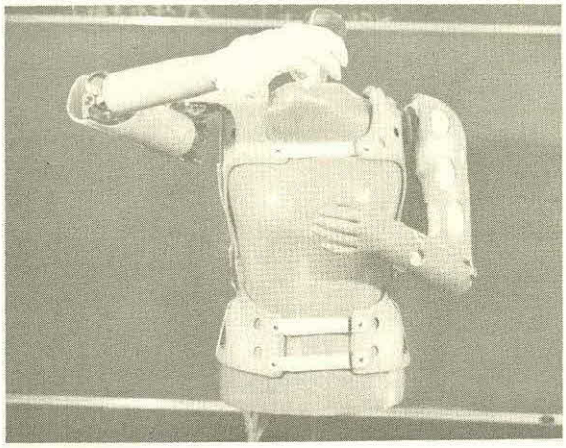
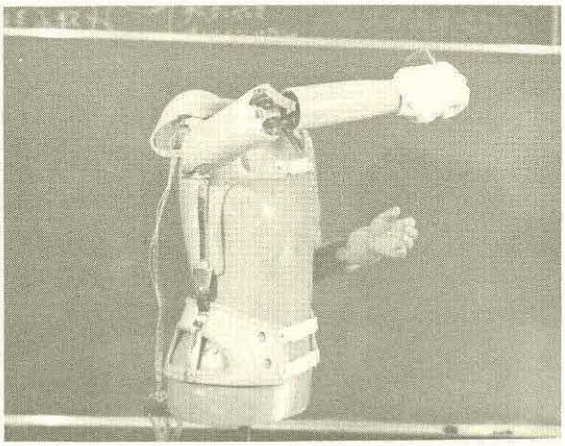


(a)



(b)

Fig.2 Electrical total arm prosthesis.

A NEW MODULAR MYOELECTRIC CONTROL AND SENSORY FEEDBACK SYSTEM

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A new family of modular myoelectic control and sensory feedback elements for use in upper extremity prostheses is described. The three-state control units feature control of two functions, such as hand opening and closing, from a single muscle control site. The sensory feedback units use electrocutaneous stimulation to indicate prehensile force on the hand prosthesis. The system is designed to complement features available in commercial systems, and is fully compatible with terminal devices available from Otto Bock and Systemteknik. All elements described in this paper are available to selected clinics in limited quantities from the developer. The basic system elements will be available commercially within about six months.

INTRODUCTION

Design and development of myoelectric control systems in this Institute began in 1962, and a number of identifiable "models" have emerged since the first clinical trial in 1965. The most recent of these was a 12 volt 3-state system designed in 1975 [1]. Compatible sensory feedback systems have been under development in the Institute since 1971, clinical evaluation of the most recent system, [2], having begun in 1978. In this paper the next phase of development is described.

DESIGN OBJECTIVES

Based upon clinical experience with previous UNB systems, the concept of 3-state control was retained as being particularly beneficial to those amputees who have relatively few control sites available. Because the largest supplier of electric hands, Otto Bock, is converting from 12 to 6 volt devices, and because Systemteknik also has standardized on 6 volt hands, it was apparent that a 6 volt control system was needed. Further, the options of using either an externally removeable or an internal battery were desired. A modular design was indicated to provide, as well, the option of using sensory feedback and the choice of two different signal processors--the traditional "analog" processor or a new "digital" processor. With respect to previous UNB systems, a reduction in size and weight was desired to make the system suitable for short below-elbow amputees as young as 8-10 years of age. Ease of fabrication using conventional methods, and ease of repair, were further objectives.

PACKAGING METHOD AND OPTIONS

To provide essential mechanical protection without the added weight of solid encapsulation, and to facilitate repair or replacement of modules, a vacuum-molded two-part case of 1 mm Kydex acrylic-PVC alloy was selected. All leads used in normal operation of the system are attached permanently, and a multi-pin connector is provided for adjustment and testing. Rectangular circuit boards are used for most efficient circuit layout, and inter-board connections are minimized for convenience and reliability. The resulting family of packages is shown in Figure 1.

A simple alphanumeric code is used to identify features in the modular system:

first two digits - model year
subsequent letters - module features
A = analog control unit
B = internal battery
D = digital control unit
F = feedback unit

Thus 79ABF is a 1979 design with analog control, internal battery and feedback, while 79F is a feedback module for use with a non-myoelectric prosthesis.

Approximate overall dimensions and weights are as follows:

Model	Length (mm)	Width (mm)	Height (mm)	Weight (g)
A or F	63	43	20	41
AB	63	43	41	122
ABF or DBF	63	43	49	140

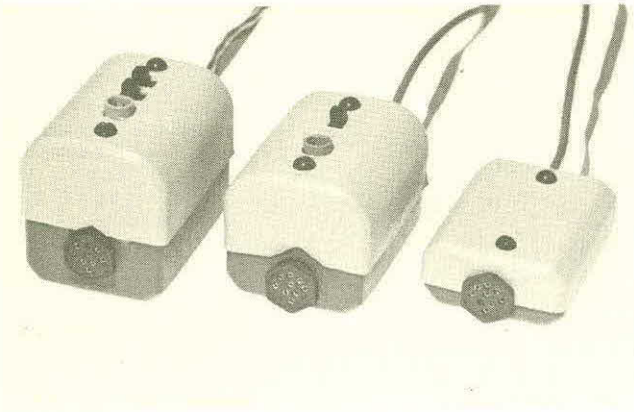

Figure 1 Packaging Options
Left to Right: 79ABF or 79DBF,
79AB, 79A or 79F

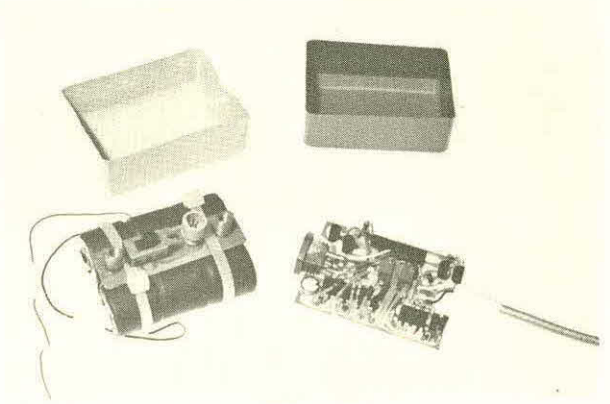

Figure 2 79AB System Diassembled,
Showing Case, Battery Pack and
Analog Control Board

3-STATE ANALOG CONTROL UNIT

The analog control unit is shown in Figure 2, and a block diagram in Figure 3. Myoelectric signal is picked up by stainless steel surface electrodes and fed to amplifier A, which has a differential input, a common mode rejection ratio of 85 dB, and a gain of 2500. Processor P is a low-pass filter (50 ms time constant), with a voltage gain of 2. The output of the processor is compared with adjustable switching levels S1 and S2 in comparators C1 and C2, where S1 < S2. If the signal is < S1 both Dr1 and Dr2 output lines are at zero potential, and the prosthesis is not activated. If the signal is > S1 but < S2, Dr1 is high, corresponding to one function of the prosthesis (e.g. hand closing). If the signal is > S2, Dr1 is at zero and Dr2 high, yielding the other function (e.g. hand opening). A time delay permits transitions between "off" and Dr2 without activating Dr1. The logic signals at Dr1 and Dr2 are compatible with the power switching modules in 6 V Otto Bock and Systemteknik hands.

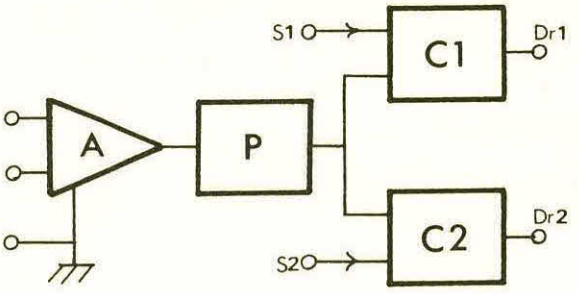

Figure 3 Block Diagram Of
3-State Analog Control Unit

An operational amplifier not shown in Figure 3 is used to establish a virtual ground at the midpoint of the 6V battery, eliminating the need for a centre-tapped supply. The internal battery contains 6 225 mAh NiCad cells, giving longer life and faster operation than the 5-cell unit supplied by Otto Bock (No. 757B8). The battery charger operates at a nominal 25 mA rate, and has positive indication of charging current.

TRAINER

To apply a 3-state analog myoelectric control system it is desireable to adjust S1 and S2 to optimum levels for the patient. These settings depend upon the dynamic range of myoelectic signals available. Measurement of this and of S1 and S2 is facilitated by use of the 79T Trainer, a simple meter arranged to connect directly to the analog control module. The trainer also may be used for training patients in the use of myoelectric control, using either the meter, indicator lights (corresponding to Dr1 and Dr2) or a prosthesis as the display. The 79T Trainer is illustrated in Figure 4.

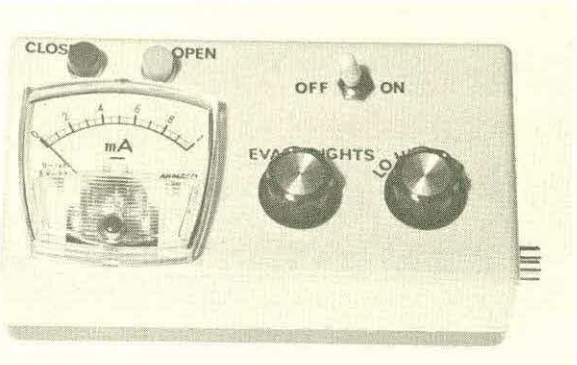

Figure 4 79T Trainer

SENSORY FEEDBACK (ANALOG SYSTEM)

In this system electrocutaneous stimulation is used to indicate the prehensile force applied by an artificial hand. The force is detected by a pair of semiconductor strain gages mounted on the index finger, using a conventional bridge configuration. The amplified output of the bridge is fed to a voltage controlled oscillator which modulates a 3 kHz square wave generator. The output of the square wave generator is amplified and transformer coupled to a self-shielding coaxial surface electrode.

The resulting stimulus is a series of bursts or pulses of a 3 kHz square wave, with the pulse rate proportional to force, over a range of Ø-6Ø pulses per second corresponding to a force range of Ø-1ØØ N. Some advantage in information rate is achieved by having the modulator operate at a constant duty cycle at pulse rates over 2Ø pulses per second.

In addition to the use of a shielded stimulating electrode, it has been found useful to gate the myoelectric preamplifier off during the stimulus bursts in order to minimize interference from the feedback signal into the control system. With these precautions no problems have been observed in relation to electrode placement, although it is customary to place the stimulating electrode remote from pickup electrodes to the extent that this is feasible within the socket.

3-STATE DIGITAL CONTROL UNIT

Development of a digital myoelectric control by Prof. V. Dunfield of this Institute in collaboration with Dr. E. Shwedyk of the University of Manitoba has been under way for about four years, and clinical trials are to begin early in 1980. Although this new approach has been designated "digital" because of the use of digital circuitry, a more appropriate name might have been a threshold processor. The technique is related to one used by Childress [3] who first employed the "time greater than a threshold" as a myoelectric control parameter. Anticipated advantages relative to the analog system include better noise immunity, simplicity of field adjustment, and adaptability to more complex processing techniques.

With reference to Figure 5, the myoelectric signal is amplified (A) and passed to a window comparator (WC) which gives an output whenever the absolute value of the signal exceeds a preset threshold. This output is sampled at the clock frequency and the resulting "1" or "Ø" stored in the shift register (SR). This method stores, in sequence, a representation of the signal over the averaging period T. Between samples the contents of the shift register are recycled and accumulated in the counter (CR). Each time, the oldest bit of information is discarded. This gives a new estimate of the signal each time a sample is taken. Each sample is weighted equally, giving an output equivalent to an analog finite time integrator.

Thus the output of the counter (CR) represents the signal "strength", and may be monitored easily to determine which of three states is desired. This is done by using two retriggerable monostables (DM). The first gives an output when the S1 switching level is crossed. The second gives an output when S2 is crossed and shuts down the first. A short delay is added to the first output, as in the analog system, to permit transitions between off and the higher level without activating the intermediate level. The outputs Dr1 and Dr2 are compatible with the Otto Bock and Systemteknik hands.

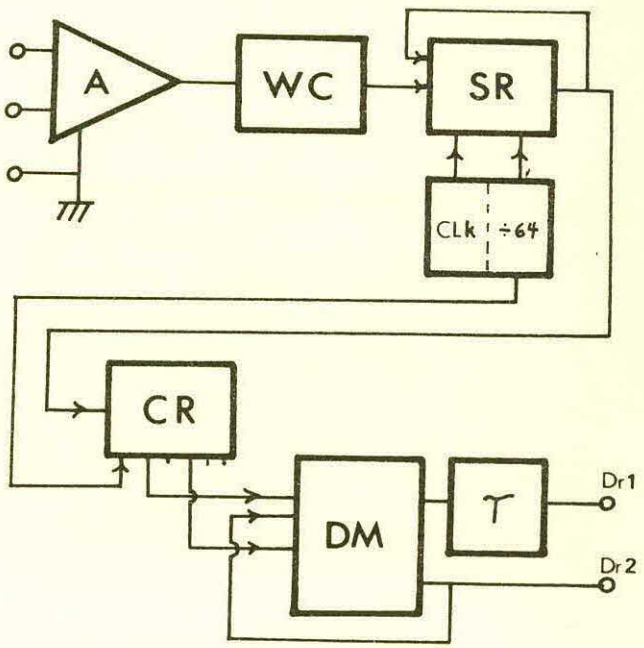

Figure 5 Block Diagram of 3-State Digital Control Unit

Further developments planned include a matching sensory feedback unit, sharing some circuitry with the digital control. Also, a control system which responds differently to transient and steady state signals is being studied.

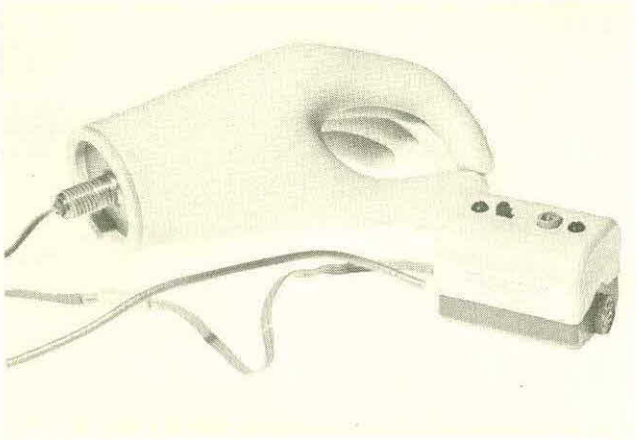

Figure 6 79AB Control with Internal Battery and Otto Bock 8E14 Hand

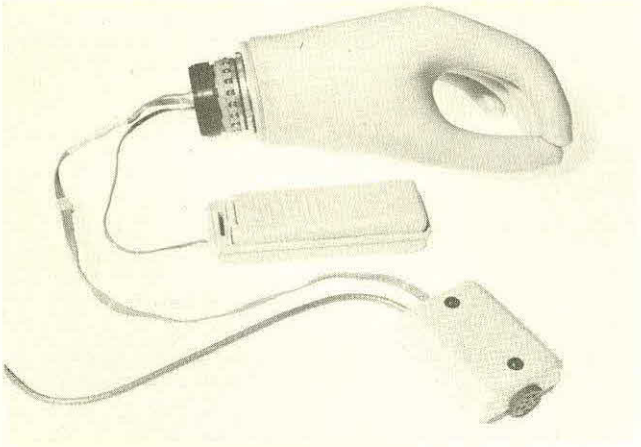

Figure 7 79A Control with Otto Bock 8E17 Hand and 757B8 Battery

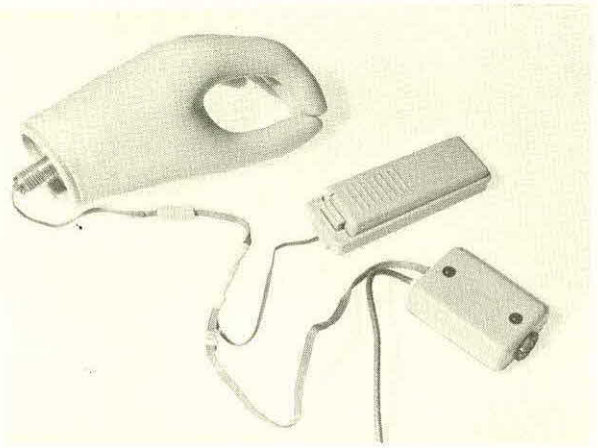

Figure 8 79A Control with Otto Bock 8E14 Hand and 757B8 Battery

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EXTENDED PHYSIOLOGICAL PROPRIOCEPTION IN A MICROPROCESSOR CONTROLLED PROSTHESIS

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ABSTRACT

In providing proprioceptive feedback for multifunctional prostheses one method termed " extended physiological proprioception " (e.p.p.) has achieved a good degree of success. This method is criticized, however, on the grounds that the control algorithm does not follow the neuromuscular synergies of normal movement. A further problem with conventional e.p.p. systems is that they incorporate fixed linkages between actuators and this limits the functions that can be performed by the prosthesis. A new e.p.p. design is presented in this paper, which overcomes the problem of fixed linkages by using a microprocessor system with selectable programs to control the input-output relationships. The advantages of this system should outweight any disadvantages caused by breaking natural neuromuscular synergies.

SENSORY FEEDBACK

The normal arm receives vast amounts of sensory information encompassing such modalities as force, position, movement, and active touch (i.e. the ability to feel shape, texture, etc). Ideally, a prosthesis should be equipped with the means of supplying all of these feedback modalities. However, active touch would be an extremely difficult modality to supply (4). Thus it will not be discussed here. This paper then, examines various methods for providing an amputee with feedback of force, position, and movement.

Force feedback has received the greatest degree of attention because it should be possible to relate it to any amplitude-varying stimulus. Some researchers believe that, in the case of EMG controlled prostheses, force feedback can be imparted via afferent signals originating at the prosthesis control site (8). This form of feedback can be augmented, if necessary, by negative feedback of the force on the prosthesis which is subtracted from the EMG control signal. Other systems use electrocutaneous or vibrotactile stimulation of the amputee's stump, the intensity of the stimulation being proportional to the force generated by the prosthesis (10,11).

When designing an externally powered prosthesis for an above-elbow or higher level amputee, feedback of movement and position are of paramount importance. Without this type of feedback, the amputee will have to make excessive use of his visual system and automatic movements will be impossible. Among those who accept the need for some form of movement and position feedback, there are two widely differing approaches. The first approach proposes the use of the phenomenon of phantom sensation to inform the amputee of the position of his prosthesis (4). Prostheses which use pattern recognition of EMG signals from several muscle sites as a control modality, should be optimum for this type of position sensation. It should be realized, however, that phantom sensation is an open-loop phenomenon. This severely limits the degree of success one can expect from such a system.

A second approach to the feedback problem is one which starts off by recognizing that excellent proprioceptive feedback is obtained by an amputee using a body-powered prosthesis incorporating a Bowden cable. Force, movement, and position information are fed back to the amputee in a very natural and easily comprehendable manner. This system works very well because the movement, force and position of the prosthesis is related on a 1 : 1 basis to the movement, force and position of the Bowden cable. Dr.D.C.Simpson, of Edinburgh, Scotland, achieves a similar quality of proprioceptive information with his CO_2 powered prosthesis (12). In designing his prosthesis, Dr.Simpson introduced two important control concepts. These are termed " extended physiological proprioception " (e.p.p.) and the " unbeatable servo ", respectively. Extended physiological proprioception refers to the establishment of a direct relationship between the movement of a normally functioning joint (the input signal) and the movement of the prosthesis. The unbeatable servo feature prevents the controlling joint from moving faster or further than the prosthesis can follow. In Dr.Simpson's prosthesis, the positions of the clavicular joints are used as inputs. The prosthesis is successful and has been fitted on a large number of patients. The principle of e.p.p. is being used by other research groups in the design of electrically powered prostheses (1,13). Like Dr.Simpson, they use clavicular joint positions as input signals.

A major problem with existing multi degree of freedom prostheses using e.p.p. is that they incorporate a fixed input-output relationship which forces the movement to follow a pre-assigned pattern. This severely limits the range of activities which can be performed. Another criticism which has been leveled at e.p.p. systems is that the control signals do not follow the natural neuromuscular synergies (9). It is

claimed that the only truly feasible design for a prosthesis is one which uses a separate control site for each degree of freedom, with the control sites chosen so as to follow the synergies of normal limb movement. This strategy involves the sacrifice of position proprioception as well as giving to the amputee the job of coordinating the various degrees of freedom. Work continues on developing prostheses based on this principle (3) but, to date, no fully acceptable unit has been designed.

A relatively new approach to prosthesis design involves the use of microprocessors to analyze the EMG signals and control the actuators. Two major developments are currently underway: one by Lyman and his associates at U.C.L.A. and the other by Graupe and his associates at Colorado State University (6, 2). The U.C.L.A. system uses EMG signals from nine sites quantized into five amplitude levels. The microprocessor then uses a Nearest Neighbour Classifier (NNC) algorithm to estimate the desired movement of the prosthesis. Some work, on a very basic level, is being done to provide sensory feedback by means of an electrotactile display (7). It should be noted that this system, based as it is on mean EMG levels, uses the low frequency characteristics of the EMG signals.

A system which uses the whole EMG frequency spectrum has been developed by Graupe and his associates (2). In this system, from one to three muscle sites are used to control five limb functions. Function discrimination is achieved using an Auto-Regressive Moving Average (ARMA) indentification algorithm. The discrimination success rate is an unacceptable 85% and no sensory feedback is provided.

The two prostheses discussed above can be considered open-loop systems since the only attempt to provide feedback information is through electrotactile stimuli (The U.C.L.A. system). This form of display, while useful to some extent for force feedback, is totally inadequate as a means of supplying position information. Because of the open-loop nature of the control, neither system is likely to gain wide acceptance among amputees.

After reviewing the various options for the control of prostheses, and based on our experience in this field, it was decided to set down the following priorities for prosthesis design:

(i) Proprioceptive feedback of force, movement and position should be made available to the amputee.

(ii) An easy-to-learn means of achieving coordinated movement of the prosthesis should be implemented.

To attain these two priorities we choose a modified e.p.p. system. The principal modification involves the incorporation of pre-programmed (and therefore selectable) input-output relationships. Thus, while linked motions are retained, the system is made much more flexible by allowing many different linkages to be pre-programmed. An additional advantage stemming from the use of a microprocessor is that the dynamic characteristics of the actuators can be put under program control. In choosing this particular design, we are ignoring the synergy of normal movement. However, we believe that the gains resulting from the excellent proprioceptive feedback will far outweigh the problems caused by forcing the amputee to learn new neuromuscular synergies.

DESCRIPTION OF PROTOTYPE PROSTHESIS

In order to evaluate the design, a prototype has been built for an above-elbow amputee. Later, the design will be extended to cover higher-level amputees.

Functions

The functions envisaged are: hand prehension, wrist rotation, and elbow flexion/extension.

Control

Hand prehension will be controlled by EMG signals from biceps and triceps muscles using the system developed at the Rehabilitation Institute of Montreal (5). Sensory feedback is not as important for prehension as it is for wrist rotation and elbow flexion/extension, because prehension is not primarily a position function. Wrist rotation and elbow flexion/extension will be controlled in a coordinated manner using, as input, an under-arm goniometer which measures shoulder flexion/extension and abduction/adduction. This goniometer will be used to provide both the extended physiological proprioception and the unbeatable servo features. The choice of shoulder movement as the input signal is based on a recognition of the fact that many manipulative tasks involve a coordination of shoulder, elbow and wrist movements. Thus the use of shoulder movement as an input signal will, in many cases, be a help rather than a hindrance to the control of the prosthesis. Thus, for example, with ten selectable programs stored in the microprocessor memory, eight can deal with very specific tasks (e.g. bringing objects to the mouth, opening a door, etc.). The remaining two can be for general purpose use (e.g. one to be used while standing and the other while sitting.

To include the unbeatable servo feature in the system, the input goniometer will have to be able to continuously limit shoulder movement in accordance with the current output capabilities of the system. We hope to achieve this by providing the goniometer with a braking mechanism having a small amount of backlash. This strategy will enable the microprocessor to sense the direction of attempted movement. It can thus release the brake if movement is attempted in one direction while at the same time preventing movement in the opposite direction. The position and direction of the braking will be continuously varied by the microprocessor program. For shoulder disarticulation and forequarter amputees, clavicular positions will likely by used as the input signal. However, other alternatives will be investigated.

PROTOTYPE MICROPROCESSOR SYSTEM

The microprocessor system being used in this work is a prototyping system based on the RCA 1802 CMOS processor, which is hosted by a DEC

PDP 11/45 minicomputer. The minicomputer provides a wide range of facilities for program development. Programs are written in the assembly code of the micoprocessor and are assembled by the minicomputer. Any alterations to the text of the program can easily be made using the editor of the minicomputer. The assembled code can be down-line loaded into the microprocessor memory and the program can be run under minicomputer supervision using single step and trace facilities. The microprocessor, its support electronics, 2K RAM and the PDP II interface are housed in a 19 inch rack with a common backplane bus. This has enabled us to build a prosthesis interface simply on two cards for the rack system. The RCA 1802 has been chosen for this application because it is a CMOS processor and therefore has a low power dissipation. The low power dissipation will help us to have a self contained prosthesis, with integral power supplies, which has a sufficiently long daily life to make it practical.

A simplified block diagram of the prototype electronic hardware is shown in figure 1. The input to the microprocessor is taken from four linear potentiometers. Two potentiometers provide the amputee's shoulder flexion/extension and abduction/adduction information from the goniometer. The other two potentiometers provide information on the prosthesis wrist rotation and elbow flexion/extension. All four potentiometer signals pass into an analog multiplexer, the output of which is fed to an analog to digital converter (ADC). The multiplexer is under the complete control of the microprocessor via the Potentiometer Register. This allows for maximum flexibility in developing programs for a prototype amputee-worn prosthesis. This flexibility means that the potentiometers need not be considered in a fixed sequence at all times but that the sequence may be varied or a potentiometer omitted, for instance if shoulder abduction/adduction is not encountered in a particular movement.

The microprocessor, having received data on the shoulder position, can use this data with a look-up table to determine the required wrist and elbow positions. Then, using the wrist and elbow data input from the potentiometers, the desired motor drive can be calculated and a signal, which is basically pulse width in nature, can be applied to the wrist and elbow motors. Repeated interations of this nature provide full control for the prosthesis. To enable a different set of linked movements between shoulder and prosthesis, it is only necessary to employ an alternate look-up table for the above calculations. Many alternatives may be stored in the microprocessor system, any particular one being chosen by some external input to the system.

A bench prototype has been built and programmed. Initial tests have been performed using a goniometer previously designed for research purposes (see figure 2). A less obtrusive goniometer will be designed for fitting on amputees. The results obtained so far have been very encouraging and necessary modifications are currently being made.

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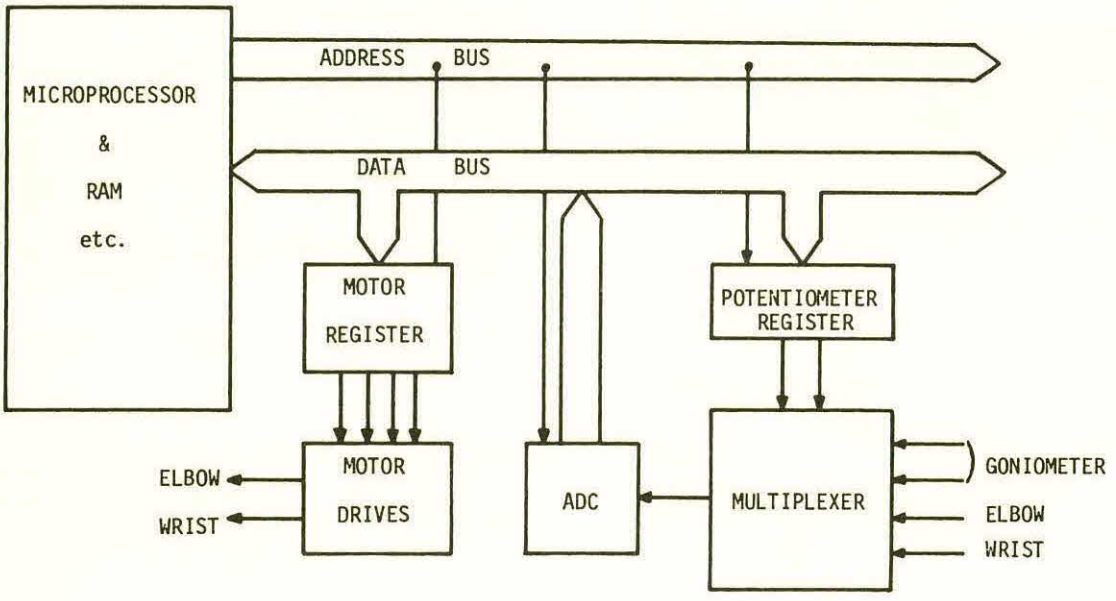

Figure 1. Simplified diagram of the prototype electronic hardware.

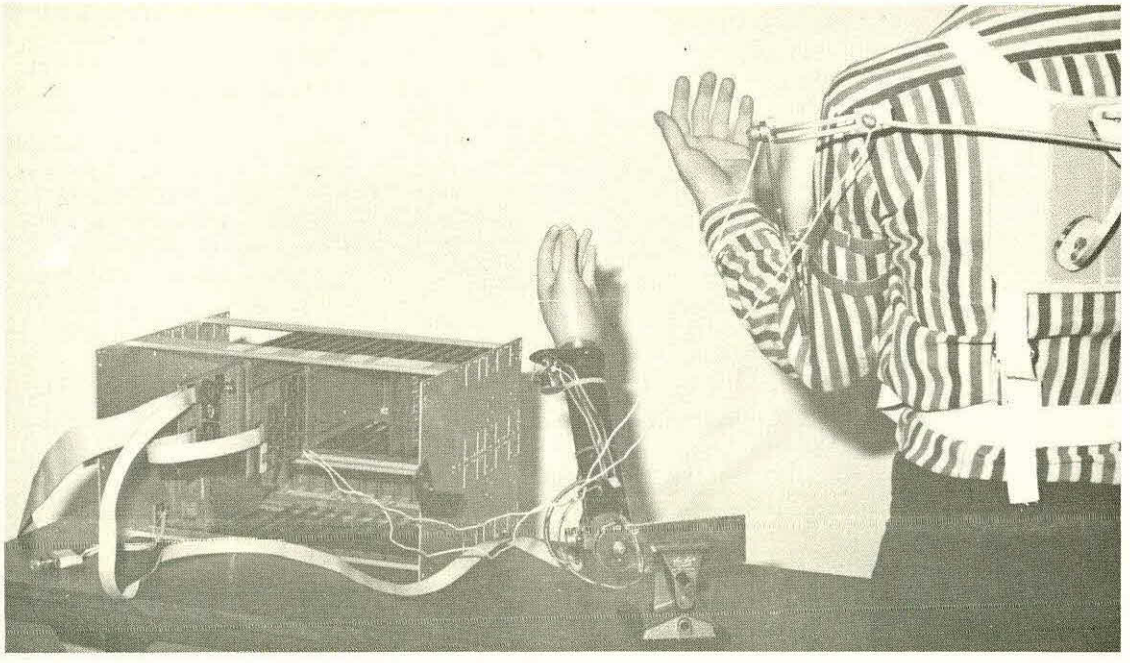

Figure 2. Bench prototype of the prosthesis.

MYOELECTRIC KNEE CONTROLLER FOR A/K PROSTHESES

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ABSTRACT

Work is presented leading to the development of an experimental, active, volitionally controlled above-knee prosthesis. The project is being implemented in four phases: 1) An investigation of myoelectric processing techniques to identify the intent of the amputee. 2) The design and modeling of a knee actuating device which is small and light enough to be contained within the prosthesis, yet is capable of meeting the torque requirements of swing phase control at various cadences. The device will also permit the amputee limited torque capability for volitional control in non-gait situations. 3) The design of a controller which, given the intent of amputee, will calculate the appropriate input for the actuator to realize that intent. 4) The construction and preliminary evaluation of an experimental, prototype prosthesis.

INTRODUCTION

In the past, knee control for above-knee (A/K) amputees has been largely limited to preprogrammed, passive devices which are not easily controlled by the amputee. The earliest attempts at control involved the introduction of constant friction into the knee joint; e.g., (1) or (2). An improvement is offered by retarding heel-rise following the initiation of knee flexion and by retarding knee extension preceding heel contact. The principle disadvantage of such devices is that the torque introduced does not vary throughout the gait cycle. Improvement can be offered by making damping a function of knee angle using an eccentric cam.

Currently the most popular damping devices are hydraulic or pneumatic. The important advantages offered are due to the property that the resistance to flow of a fluid is proportional to its velocity. Therefore, as cadence rate is increased, knee resistance is increased. The device typically takes the form of a piston and cylinder with two separate flow paths governed by one-way valves; one path controlling resistance to knee extension; the other, resistance to flexion; e.g., see (2).

The state of the art in the design of knee control for A/K prostheses is reflected in the work of Flowers, et. al. at M.I.T. (3). An electromagnetic particle brake controls knee damping as a function of the swing phase torque profile using slide potentiometers. The amputee is provided freedom to adjust the device characteristics as cadence rate and environmental conditions change.

A major deficiency in the design of all A/K prostheses exists between the amputee and the knee controller, i.e., the communications link between the amputee and his device. If the controller is not to function only in a preprogrammed manner, it becomes necessary to provide the user with a means to convey his intent.

Several control schemes for upper extremity prostheses attempt to solve the problem by employing pattern classifiers operating on spatial patterns of EMG. Each EMG sampled is rectified and low pass filtered such that the resulting signal is proportional to the energy content of the EMG measured. Exampled include the "Temple Arm" (4) and "Swedish Hand" (5).

Other approaches attempt to use more of the information contained in the EMG than simply magnitude or energy content. At the University of Utah, Jacobsen et. al. (6) are correlating shoulder torques, using linear regression analysis. Graupe et. al. (7) are developing a control scheme based on an autoregressive (AR) model of an EMG signal. The coefficients of the AR model are used to assign an unknown signal into classes of allowable joint responses. The method has been shown to work in the control of a single degree of freedom assistive device and is being investigated for use in multiple degree of freedom control.

In this paper the authors examine the feasibility of the spatial pattern recognition approach in discriminating between EMG activity which is directed at the knee from EMG activity directed at the hip. Part I. describes the analysis and results of the pattern recognition work. Part II. describes the design of the actuator, and Part III. the design of the control system. Part IV. concludes and summarizes the work.

I. PATTERN RECOGNITION

Experimental Design

It was hypothesized that spatial patterns of EMG from muscles above the site of amputation could be separated into groups representing: 1) knee flexion, 2) knee extension, and 3) hip action involving no knee activity. The following experiment was designed to test this hypothesis:

Using normal subjects, bipolar surface electrodes were placed on six muscles in the thigh and hip region of the right leg. A strap was placed around the ankle and above the knee providing attachment points for hooks and cord. An apparatus

was constructed so that known weights could be used to load the hip and knee joints through a system of pulleys. EMG was recorded as the hip and knee underwent combinations of quasi-isometric flexion and extension with the right leg in various configurations. The signals obtained were logarithmically amplified, bandpass filtered, integrated, and sampled.

Analysis

Each pattern resulting from the six EMG recording sites was considered as a vector in six-dimensional space, with all vectors from knee flexion runs belonging to one class, and all remaining vectors to a second class. Pattern recognition algorithms were then applied to judge the degree of separability between the two classes. Knee extension vectors were treated similarly.

Three algorithms were used: 1) a linear discriminate function (LDF), as used in several existing control schemes, 2) a quadratic discriminate function (QDF), and 3) a polynomial discriminate function (PDF) (8). The PDF was preceded by a feature extraction (FE) algorithm which linearly transformed each vector to a reduced dimensionality, while maximizing the expected distance of points in the first class from those in the second.

Results

Experimental results indicated that the discriminate functions, each based upon a Bayes' classifier, will yield virtually errorless separability between spatial patterns of EMG which result from knee action. Additionally, an apparent correlation between classifier output and knee torque was noticed. Figs. 1 and 2 show pattern classifier output for knee torques increasing in magnitude such that first, a progressively large flexion moment was required (Fig. 1), and secondly, such that a progressively larger extension moment was required (Fig. 2). Positive ordinate values indicate that a pattern has been accepted into the class; negative values indicate rejection. It can be seen that classification was performed without error.

II. KNEE ACTUATOR

The design of the actuator consists of a pneumatically driven, double-acting ram with a pressure reserve vessel located in the shank of the prosthesis. If the actuator is to act as an energy dissipator (i.e., the torque which must be supplied by the device opposes the direction of knee rotation), valves will be appropriately adjusted to provide air flow resistance. When the pressure on either side of the ram exceeds the pressure of the reserve tank, that side of the ram will be connected to the pressure vessel to conserve energy. When it is required that the actuator supply energy (i.e., provide torque which is in the direction of motion), the vessel will serve as the driver. Digital computer stimulation indicates that the actuator will be capable of driving the knee with torques of up to 270 in.-lbs., well within the torque demands of swing phase in normal locomotion.

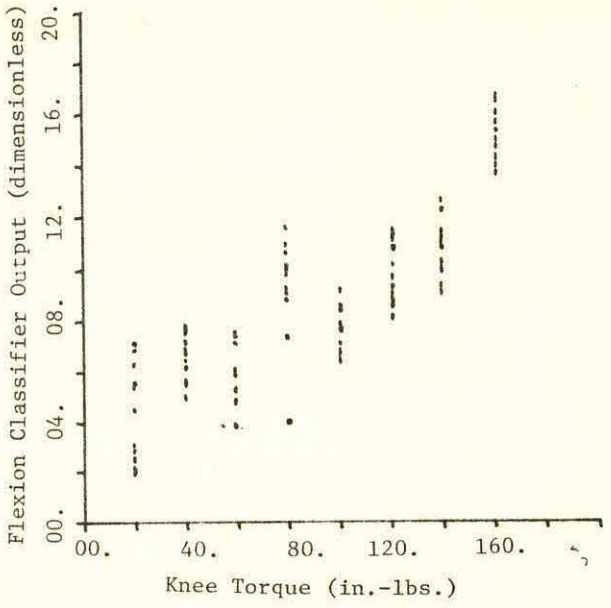

Fig. 1 Flexion Classifier Output vs. Knee Torque

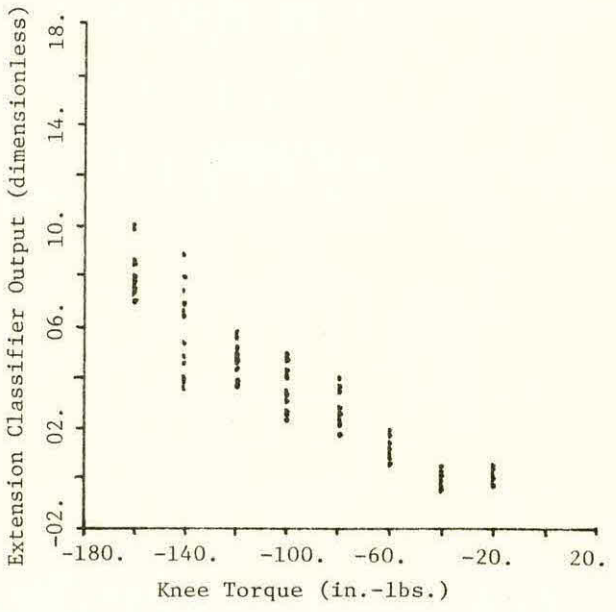

Fig. 2 Extension Classifier Output vs. Knee Torque

III. ACTUATOR CONTROL SYSTEM

Each of the equations governing the behavior of the actuator was linearized using a Taylor series expansion about a normal trajectory of the state variables of the actuator. A piece-wise linear model of the actuator was then used as the plant for which a compensator was to be designed. Fig. 3 illustrates an expanded view of the knee control device. The controller can be formulated as a linear multivariable tracking problem (10)

with input disturbance. The disturbance F_{ext} results from forces external to the actuator caused by hip angular rotations and displacements. The reference input, F_a, the desired actuator force, is determined by the difference between the output of the flexion and extension classifiers. The dashed line in Fig. 3 represents the feedback of state information from the actuator. This is accomplished, in part, by the incorporation of appropriate transducers in the design of the actuator, and, in part, by the use of a reduced order state observer.

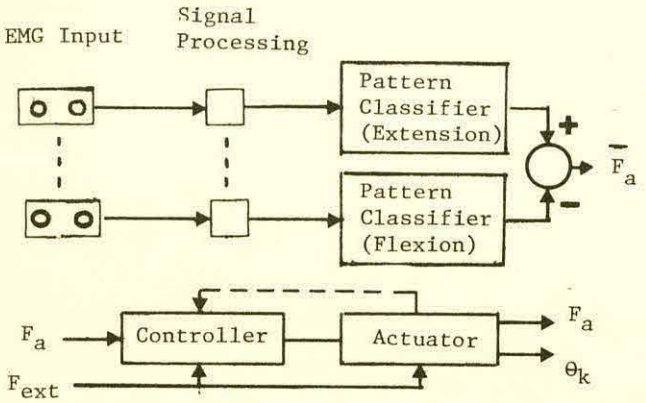

Fig. 3 Block Diagram of Device

Digital Simulation

A model of the actuator was derived, and the entire thigh-actuator-shank system was simulated on a digital computer. The system was treated as a pair of rigid links pivoted at the knee center with shank mass concentrated at its CG (see Fig. 4). The hip was subjected to displacements (X_H, Y_H) and the thigh to rotation (θ_T) as would occur during normal locomotion. The controller was programmed to provide input to the actuator such that a normal torque profile would be produced at the knee. Results of the simulation are presented in Fig. 5 where "actuator length" is as defined in Fig. 4.

IV. PROTOTYPE CONSTRUCTION

The results of the simulation have provided the design specifications which must be realized by the experimental model if reasonable performance is to be expected. A socket was constructed incorporating the electrodes necessary for EMG detection. An experimental actuator including the necessary transducers for feedback of state information to the controller was then installed. Evaluation and modification of the device is currently being undertaken to compare the performance of the device against existing, passive A/K prostheses.

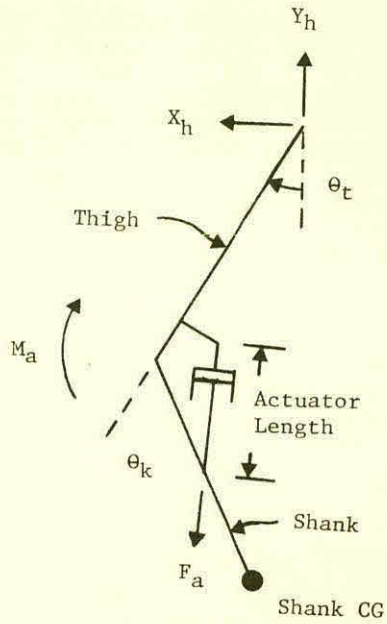

Fig. 4 Thigh-Actuator-Shank System

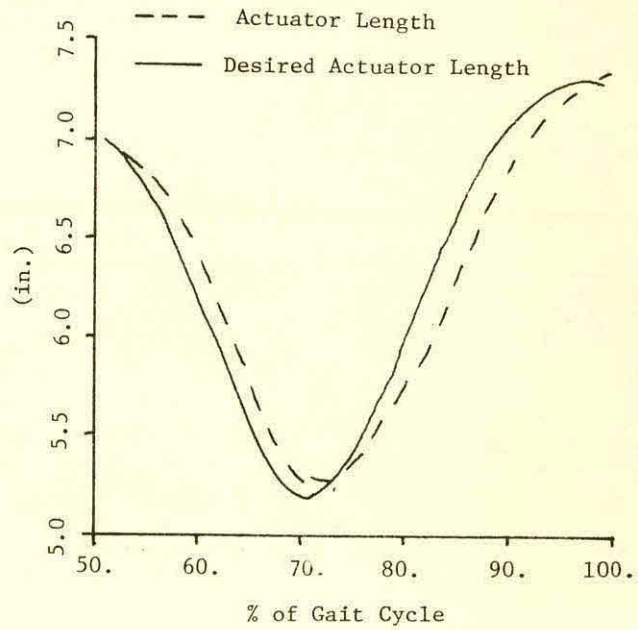

Fig. 5 Desired vs. Simulated Actuator Response

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AN EMG STARTLE RESPONSE AS A STUMBLE INDICATOR

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ABSTRACT

An EMG startle response was tested as an indicator of a stumble situation. Experiments on normal subjects verified that the startle response does occur shortly after a stumble and can therefore be used as an indicator of a stumble. The approach was employed as part of a safety system on an active experimental A/K prosthesis which is capable of many active knee functions (step over step stair climbing, etc.). The EMG stumble indicator worked very well in our active system and appears well suited for other locomotion related safety systems.

Precautions to insure safety must be included in any prosthetic device. Special attention to safety is necessary when the prosthetic device is experimental. Amputee safety can be improved by including some relatively simple precautions in a safety system. In the case of experimental lower limb prostheses two main hazards exist which must be considered and included in such safety systems.

First, even when an experimental prosthetic system is performing in the desired manner, the chances that the amputee might stumble and fall are higher than if the amputee were accustomed to the prosthesis. The amputee's inputs and reactions to a new prosthesis or prosthesis controller change as the amputee develops a mental model of how the prosthesis will behave in different situations. Even though the amputee is generally more cautions during this time, chances of a stumble are somewhat higher during this training period than after the subject has become accustomed to the prosthesis. The second hazard in experimental prosthetic systems is common to any experimental device. Experimental systems fail more frequently. Some problems cannot be anticipated during the formulation stage of a project and do not appear until experiments are performed.

An experimental prosthesis can be made safer for an amputee by including automatic safety precautions. Such a safety system can employ safety measures if a stumble or other emergency situation occurs. If a stumble does occur, it must be recognized quickly if any automatic safety system is to be effective. One automatic safety system that has proven very effective in testing an active experimental electrohydraulic prosthesis system (1) uses an EMG startle response as a stumble indicator.

Shortly after an emergency or startle situation, many skeletal muscles are tensed prior to major recovery activity. This "startle response" has been observed in humans in drop tests (2) but has not been tested in stumbles during locomotion. In order to determine whether an EMG startle response is a useful indicator of a stumble situation, a series of experiments were performed on normal subjects.

Nine normal subjects were used in deliberate tripping tests during level walking. After watching a demonstration of the experiment, each new subject was fitted with knee and elbow pads and a hockey helmet (Fig. 1). EMG surface electrodes were placed on the skin over the biceps and triceps on each arm and held under elastic wraps. Biceps and triceps were chosen because they have a low EMG level during normal locomotion. Although this approach would not be useful if the subject were actively using his arms while walking, it is a useful approach for a laboratory situation. For these experiments, a rope was tied to each ankle. A trip switch between each ankle and the rope indicated when the rope was pulled. Each subject was asked to walk on a level surface and at an unexpected time the investigator pulled on one of the ropes, thereby creating a stumble situation. The rope pull always occurred during the swing phase since a much larger disturbance would be required to cause a stumble during stance. The rope pull was not violent enough to upend the subject but just enough to interrupt the subject's gait pattern. Every subject recovered from each test without falling to the floor. Approximately 20 tests per subject were performed. All nine subjects exhibited a startle response similar to the test shown in Fig. 2. The average time between the onset of a trip and approximate onset of EMG over 93 tests was 140 ms. with a standard deviation of about 30 ms. The response is sufficiently fast so that safety precautions can be exercised before any injury can occur.

These results for normals were assumed to be similar to stumble situations for amputees for two reasons. First, the EMG startle response appears to occur universally. All subjects tested in this investigation exhibited the response.

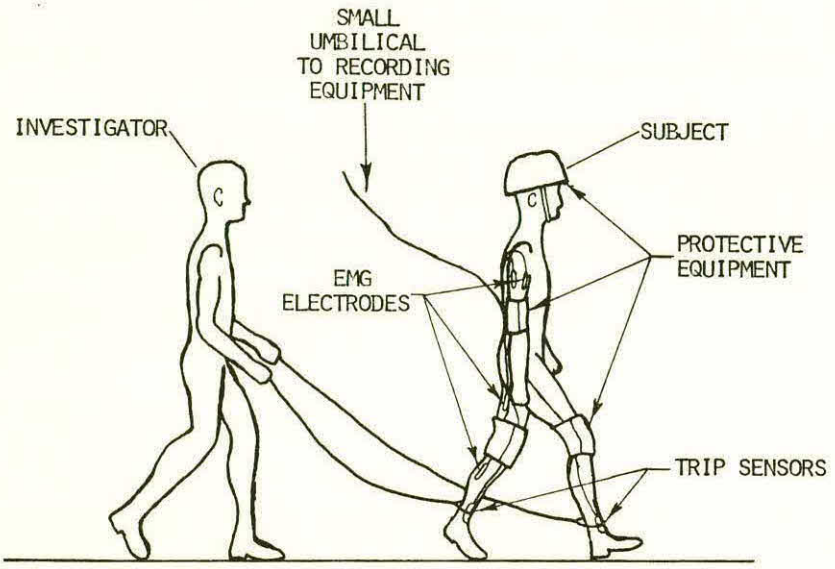


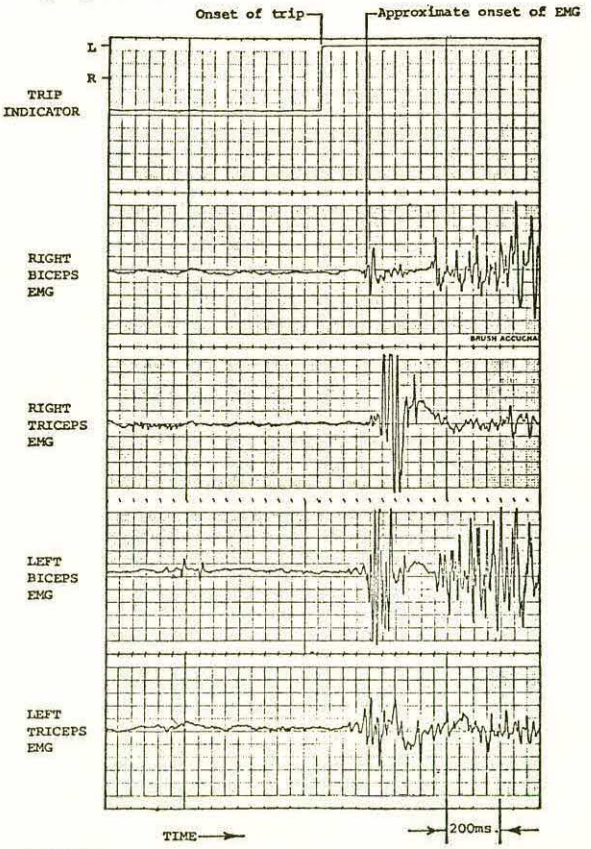
Figure 1: Startle Response Test With Normals

Figure 2: Typical EMG Patterns

Secondly, although amputees typically stumble more frequently than normals, the response does not appear to habituate away. Individual subjects were tripped many times on different days and the EMG startle response was unchanged. Although these tests were not designed to be extensive enough to determine the exact physiological mechanism for stumble startle responses, the response appears to be a functionally useful indicator of stumble situations.

For use with our active experimental prosthesis system each EMG signal is processed (rectified and low passed τ = 150 ms.). A stumble is defined when any three of the four processed EMG signals exceeds normal locomotion activity levels. If a stumble condition is recognized, an abort valve locks the prosthesis. Although different researchers suggest different courses of action (free swing, slow yield, or lock) in emergency situations, this particular hydraulic prosthesis must be locked in an emergency. Any other action requires control using transducer signals from the man/prosthesis system for feedback. If a transducer was damaged in a fall or some other condition led to an undesired control signal, a powered kick might result. The total time required to lock the prosthesis after a trip is approximately 300 ms. That time includes 140 ms. for the startle response, 60 ms. EMG processing delay, and 100 ms. for the abort valve to lock the prosthesis.

This startle response stumble indicator has proved to be a valuable safety feature during amputee locomotion trials. This safety feature was evoked on several occasions by different subjects while performing experiments. This approach has helped prevent any injury to amputee subjects due to the active capabilities of the system. These results suggest that the EMG startle response can be used as a valuable indicator of a stumble situation in many different locomotion safety systems.



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TAIPER , A TEMPORARY AIR-INFLATED PROSTHESIS FOR EARLY REHABILITATION

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Patients with peripheral vascular disease, who make up the bulk of the lower limb amputee population at Vancouver General Hospital, are often elderly and in poor general condition, and thus it is especially important to rehabilitate them as quickly as possible. A temporary prosthesis should control post-surgical oedema, promote healing, permit early ambulation and, unlike the conventional plaster cast and jig, be easily removed so that the wound can be inspected. T.A.I.P.E.R. consists of two inflatable vinyl air splints of different sizes and an aluminum frame with a SACH foot on a pylon attachment. The larger air splint provides the main support. The smaller air splint is applied first and allows control of the distal end pressure independant of the main inflation pressure. The aluminum frame is applied and the main air bag is inflated to a pressure of 30 mm Hg against the frame. Leg length can be adjusted by a telescoping fitting. The fluctuation in pressure on weight bearing removes oedema and facilitates venous and lymphatic return. T.A.I.P.E.R. is light, simple and easy to use. Clinical trials of the device are in progress.

Introduction

In 1979, thirty-one patients were admitted to the Rehabilitation Ward of Vancouver General Hospital with lower limb amputations for peripheral vascular disease. Two-thirds of them were over sixty. Surveys indicate that one-third of a group of vascular amputees will be dead within two years, and two-thirds within five years[1]. It is therefore important to speedily return these patients to their optimum level of function.

The value of the immediate post-surgical cast and jig in the early treatment and rehabilitation of lower limb amputees is well established.[2,3,4] This technique, first used by Berlemont in 1958[5] and modified by Weiss[6] involves applying a plaster-of-paris cast to the stump at the time of operation and incorporating a metal plate. An aluminum pylon and a SACH (Solid Ankle Cushion Heel) foot is slotted into the plate on the cast and the patient may begin partial weight-bearing using walking aids on the second post-operative day. The cast is left in place undisturbed for about two weeks and is then removed to view the wound, and may be reapplied for a further week. The advantages of this type of treatment can include improved healing due to controlled post-operative oedema, decreased phantom limb sensation, shorter hospital stays and a higher state of patient morale.

The immediate rigid post-surgical prosthesis does, however, have some severe shortcomings. The major disadvantage of the treatment is the inability to inspect the wound directly. In amputations due to peripheral vascular disease (P.V.D.) this is often of prime importance as the wounds are slow to heal and tissue breakdown is likely. For this reason alone, many vascular surgeons are reluctant to treat P.V.D. patients with rigid dressings. Other disadvantages of the immediate post-surgical cast and jig include its cumbersome bulk in bed, the need to limit the patient to only partial weight-bearing, and the danger of damage to the stump through localized pressures caused by a poorly applied cast, or by the stump changing shape as it matures.

The optimum rehabilitation aid would be one which controlled post-surgical oedema, permitted early ambulation and was light, safe and easy to use.

T.A.I.P.E.R.

Many of the advantages of an immediate post-surgical cast can be obtained without the disadvantages by using a double walled air splint to envelop and support the stump. The use of an air bag subjects the patient's stump to stresses of low magnitude and uniform distribution, which have been shown to reduce oedema and promote healing.[7] A temporary pneumatic prosthesis, consisting of an air bag, an aluminum frame and a SACH foot was first used by Little[8] in Australia in 1971. A second type was developed from the Australian original at Roehampton in Britain[9], and is being sold in the U.K. Since the start of the T.A.I.P.E.R project at the Vancouver General Hospital, a pneumatic prosthesis using a fibre-glass frame has become commercially available in the United States. Despite the simplicity and effectiveness of the temporary pneumatic prosthesis method of rehabilitating amputees, the technique is not widely used, largely because the device is not readily available on the commercial market.

To meet the rehabilitation needs of amputees at Vancouver General Hospital, T.A.I.P.E.R., a Temporary Air Inflated Prosthesis for Early Rehabilitation was developed by the Departments of Physical Medicine and Biomedical Engineering on a grant from the British Columbia Health Care Research Foundation. The device is composed of a double-walled air splint inflated inside an aluminum

frame, to which is attached a SACH foot. The T.A.I.P.E.R. limb allows partial weight-bearing to begin at an early stage in a patient's rehabilitation progressing to weight-bearing to tolerance. Like the rigid cast, use of the T.A.I.P.E.R. limb results in reduced oedema, reduced phantom limb sensation, and improved healing. However, the T.A.I.P.E.R. air splint is easily removed for wound inspection, and damaging localized stresses on the stump are avoided. The inflation pressure increases when the patient bears weight, from roughly 30 to 60 mmHg, creating a pumping effect which promotes venous and lymphatic return. Shear stresses on the patient's stump are also lower with the pneumatic prosthesis than might be generated in a rigid cast, as the air bags are free to deform.

The device also permits an early evaluation of the patient's ability to cope with an artificial limb. This is particularly important in the case of the bilateral amputee, where it might be questionable whether or not the patient could cope with a prosthesis. The T.A.I.P.E.R. system may be used with any patient with little expense as it is reuseable, whereas fitting of the rigid post-surgical cast entails the time of a skilled prosthetist.

Design

The T.A.I.P.E.R. is shown in Figure 1. Two double-walled inflateable bags constructed of heavy semi-transparent plastic are used. The smaller bag fits over the distal end of the stump and permits adjustment of the distal end pressure. The frame is constructed of aluminum. The top ring is larger than the bottom, and the pylon clamp at the end of the frame is set out from the central axis by about 1.5 cm. The offset is unique to the T.A.I.P.E.R. Earlier pneumatic prosthesis with the pylon in line with the central axis tend to cause varus (bowing) at the knee on bearing weight which causes pain in the patient. Velcro straps are provided as a safety factor in the event of accidental puncture of the air bag.

The foot/pylon assembly is constructed of readily available prosthetic components. The SACH foot is attached to the shank pylon by means of a standard quick-lock ankle fitting and foot bolt. Pylons are available between 5 and 30 cm in length in 5 cm increments.

The shank pylon telescopes into the slotted pylon clamp on the main frame and is fixed in the desired position with a hose clamp. The clamp has a large steel ring welded onto it to obviate the use of tools. A second clamp is brought flush with the frame and tightened to provide a safety stop in the event of the main clamp becoming loose.

Method

The T.A.I.P.E.R. is applied by invaginating the distal air bag over the end of the patient's stump, which is covered with a tensor bandage. The bag is partially inflated with air. The skin of the leg is covered, and the main air bag is drawn up over the distal bag and the limb, well into the groin. The frame is positioned over the air bags such that its upper ring is mid-thigh, and the outer bag is inflated to about 30 mm Hg. When fully inflated the bag is in contact with the vertical struts of the frame. The velcro straps are then adjusted.

The foot/pylon assembly is inserted into the pylon socket and compared for length with the other leg. It should be two centimetres longer than the sound limb to allow for vertical movement on weight-bearing. The hose clamp on the main frame is tightened, and then the hose clamp on the pylon is brought flush with the end of the frame and tightened. To ensure a secure connection, the end of the pylon should be above the top of the slot in the pylon socket.

In most cases, the patient may weight-bear to tolerance; however, walking aids such as crutches or canes are always used as a precaution against accidental puncture of the bag.

Development and Evaluation

A total of ten T.A.I.P.E.R. systems were constructed for use and clinical evaluation at VGH. Preliminary trials with the prototypes indicated a high level of acceptance from patients, physiotherapists and doctors. The need to provide patients and therapists with weight-bearing feedback led to the incorporation of a load cell into the shank pylon. It is planned that a two-tone audible feedback unit be developed, to indicate when the desired weightbearing is achieved, and to warn when the permitted maximum is exceeded in those patients with a particularly fragile stump.

At present the T.A.I.P.E.R. is being applied for ambulation, but it is hoped to introduce the air bag component into the operating room, as is done in Australia[8]. The wound is dressed lightly and the air bag is applied and inflated to 25 mm Hg for the first 48 hours. Drains are not used. After 48 hours the air bag is deflated, the dressing checked, and both the air bag and frame are re-applied and the patient begins weight-bearing.

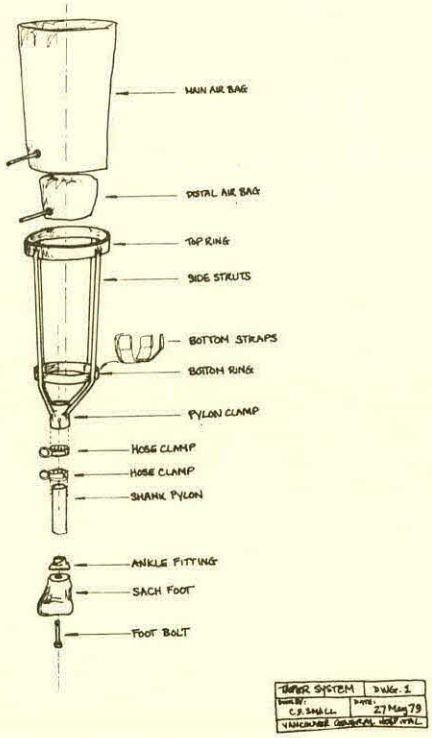
Other plans include identification of a softer and more transparent plastic than is presently used for the air bags, changing the air bag valve fittings to improve the bags' service life, and development of T.A.I.P.E.R. frames for above-knee amputees, those of below average height and children. It is felt that while individual patient progress can be adequately monitored by regular measurement of patient stump size, the overall effectiveness of the T.A.I.P.E.R. system must be determined by following the progress of many amputees over a period of years. A computerized study of lower limb amputee rehabilitation has been in progress for some years in VGH, and inclusion of T.A.I.P.E.R. use into the data collection system will permit analysis of the T.A.I.P.E.R.'s efficacy.

Summary

Use of the T.A.I.P.E.R. at Vancouver General Hospital over the past year and a half has demonstrated clearly that the temporary pneumatic prosthesis is a useful tool in the rehabilitation of lower limb amputees. It is particularly suitable for patients with peripheral vascular disease, as the stresses imposed on the stump are uniform in distribution and low in magnitude and the air splint can be removed at any time for wound inspection. The T.A.I.P.E.R. is a simple, easy-to-use and inexpensive rehabilitation aid which has been well-

accepted by patients and clinical staff; proof of its effectiveness can only be accumulated by monitoring the progress of many amputees over the coming years and comparing the T.A.I.P.E.R. results with the results of conventional treatment.

FIGURE I TAIPER System



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RETRIEVAL ANALYSIS OF TOTAL JOINT IMPLANTS

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ABSTRACT

Retrieval analysis is an efficient means of monitoring the problems associated with total joint replacements. In 5 years, 122 prostheses have been collected from 8 hospitals in the Chicago area, including 75 hips and 44 knees. The patient's case history, surgical observations and findings from laboratory examination of the implant were summarized and coded for computer analysis.

Total hip replacements are being removed at a relatively constant rate with no evidence of an increase in removal with time.

The major cause of total hip revision was idiopathic loosening (26/68) followed by trauma (15/68).

No correlations could be obtained using the current study data to demonstrate a relationship between prosthesis loosening and a radiolucent line demarcating bone and cement on x-rays or prosthesis loosening and poor cement interdigitation with bone.

Femoral stems loosened more often than acetabular cups; femoral stems often debond at the prosthesis/cement interface.

INTRODUCTION

Over the past twenty years, total joint replacement has proven to be a highly successful procedure for the rehabilitation and restoration of persons crippled by severe arthritis. The results following total hip replacement are particularly dramatic with recent clinical series reporting good to excellent long-term 10 year results of 91% for the patients followed (1).

Total joint replacement is not yet a perfected procedure, particularily for joints other than the hip. Indeed, the surgical replacement of certain upper extremity joints such as the elbow and shoulder remain in the experimental stage.

There are a number of methods for monitoring the problems associated with total joint replacement.

The most definitive is the large clinical series where all the patients of a physician or group of physicians are followed. By studying a large number of patients over many years, the problems unique to a particular procedure will be manifest. Although many clinical problems and their incidence can be determined through such a series, the causes of such problems or their solutions are often unclear. This is particularly true when multiple mechanical, chemical, biological and surgical factors are implicated. Also, many patients and years are required to complete such a study, and there is very little variable control during the course of the patient follow-up.

Another method which can be used involves laboratory analysis where theoretical, mathematical models, in vitro bench tests and in vivo animal experiments may be employed. Laboratory analyses can be completed relatively quickly and under close supervision where the number of experimental variables can be restricted and carefully chosen and where parametric analysis is possible.

However, the in vivo environment is difficult to replicate in in vitro tests and such experiments cannot model time-dependant biological responses. In vivo animal experiments more closely model the human condition but variable control becomes more difficult. Mathematical analyses are limited by the parameters included in the model.

Laboratory methods, then, are not so much monitoring systems as analytic tools and are most useful in studying problems which have become evident from clinical series or some other monitoring scheme.

A third means of monitoring problems associated with total joint replacements is through the retrieval and examination of removed implants. The study of such retrievals, when coupled with surgical observations at the time of arthroplasty revision and the clinical history of the patients, can yield detailed information on the in vivo function of the implant and the conditions which led to its ultimate removal.

Although retrieval studies cannot give absolute failure rates since the patient population from which the sample is drawn is unknown, failure modes and their relative rate can be determined. A second disadvantage is that not all implants which may be classified as a clinical failure come to removal so that retrieval studies do not monitor all failure modalities.

Nevertheless, retrieval analysis provides an efficient means of obtaining detailed information on the in vivo behavior of implants and is a good means of monitoring and following the failure modes of joint replacements in use.

METHOD

An ongoing project of our laboratory has been the collection and examination of removed total joint replacement prostheses from all the patients of 8 hospitals in the Metropolitan Chicago area. Over a period of 5 years, 122 prostheses have been collected, including 75 hips and 44 knees. It was attempted to collect all prostheses removed by the participating surgeons, and this was generally successful.

The study of the removed components consists of i) examination of patient medical records, ii) analytical review of x-rays, iii) observations at the time of the revision surgery, iv) post-operative gross examination of the removed components. A member of the project team studies all the pertinent medical records and x-rays and, when practical, attends removal surgery to record all surgical findings. The removed components are then studied post-operatively.

Data from these analyses are summarized on coded forms and then transferred to magnetic tape for storage and statistical analysis. Currently, 135 variables are recorded.

RESULTS

This report will be limited to a discussion of the 75 total hip replacement (THR) prostheses collected, and in particular to those prostheses which are bonded to bone using polymethyl methacrylate (PMMA) cement. Seven of these are of the metal-on-metal type and 61 are of the metal-on-plastic type.

Selected demographic patient data is given in Table 1 and some selected study statistics are given in Tables 2-3 and Figures 1-9.

Table 1
GENERAL PATIENT DATA

AGE: median age = 64, (9 40, 16 70)

SEX: 28 males, 40 females

WEIGHT: 30 normal, 38 moderately or markedly overweight

DISEASE: osteoarthritis - 30
rheumatoid arthritis - 8
other - 30

PRIOR REVISION: 25% had prior TJR revision

SYMPTOMS: 81% experienced symptoms less than 1 year prior to revision

DISCUSSION:

Figure 1 indicates that total hip patients appear to be at greater risk of developing problems during the 3 year period following arthroplasty and that after the initial 3 year time period, the revision rate may be more constant.

Table 2
GENERAL STUDY DATA

NO. OF CASES: 68

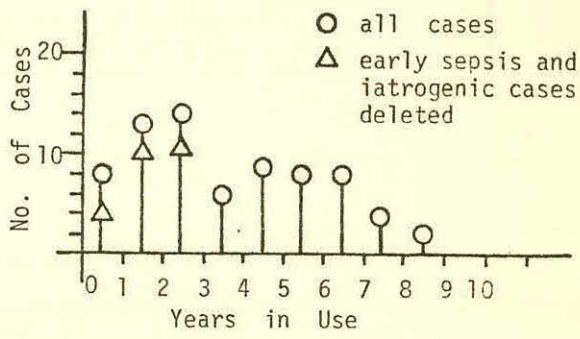
NO. OF HOSPITALS: 8 (2 hospitals contributed 76% of cases)

NO. OF SURGEONS: 21 (3 surgeons contributed 63% of cases)

CASE SOURCE: surgeon's own case - 72%
referral case - 27%

PROSTHESIS TYPE: 7 McKee-Farrar
19 Aufranc-Turner
4 Charnley
36 Charnley-Muller
2 other

Figure 1
LENGTH OF TIME PROSTHESIS IN USE BEFORE REMOVAL



However, when the removals due to early sepsis and iatrogenic factors are removed from the totals, the number of removals per year of use is seen to be relatively constant. In particular, there is no apparent increase in failure rate with time in use.

Figure 2
PATIENT ACTIVITY

	active/ working	active with support	inactive
before symptoms	42	18	7
after symptoms	13	28	26

Figure 2 clearly demonstrates both the rehabilitative aspects of THR and the disabling character of a failed joint.

For those metal-on-plastic cases in which iatrogenic or technical factors were the major incident leading to hip joint revision, reoperation occurred within 3 years in all 6 cases and in fact, symptoms developed within 1 year for 5/6 cases.

All of the 7 fractured femoral prosthesis stems showed roentgenographic or post-operative evidence of loosening prior to fracture, implying that the fixation of these prostheses had already been compromised prior to device failure. One explanation for such fractures, then, may be that certain regions of the loosened stem are subject to elevated stresses due to gross loosening and then go on to develop metal fatigue and fail.

Table 3

CAUSE OF REMOVAL:	Metal/Metal	Metal/Plastic
Idiopathic Loosening	4	22
Trauma or fall	1	14
Iatrogenic factors	2	6
Sepsis, Early	0	4
Late	0	6
Prosthesis Fracture	0	6
Other	0	3

Figure 3
PROSTHESIS LOOSENING

	Metal/Metal		Metal/Plastic			
			Non-septic		Septic	
	loose	firm	loose	firm	loose	firm
acet. comp.	3	4	10	39	2	10
fem. comp.	5	2	31	18	6	6

Figures 3 and 4 tabulate the loosening of the hip joint components as observed at the time of revision surgery. For all the prosthesis designs, the incidence of femoral component loosening is substantially greater than loosening of the acetabular cup and is certainly a reflection of the more adverse stress conditions experienced by the femoral stem. One exception is the apparently higher incidence of metal acetabular cup loosening. However, when the cases of prosthesis malalingment and impingement are removed, the rate of metal cup loosening more closely approximates the general trend.

Figure 4
IDIOPATHIC LOOSENING

	Metal/Metal		Metal/Plastic	
	loose	firm	loose	firm
acetabular component	1	3	5	20
femoral component	4	0	16	9

More than one-third of the cases (26/68) have been classified under the heading of idiopathic loosening as the cause of removal. These are cases for which the cause of removal could not be attributed to some of the more obvious failure modes listed in Table 3.

The ultimate loosening and failure of the prosthesis was probably due to the formation of an interfacial tissue leading to debonding between bone and cement. In 58% of the idiopathic loosening cases, a fibrous interfacial tissue was observed at the time of revision surgery; this tissue was reported in only 31% of the other non-septic removal cases. Additional fibrous tissue probably formed after loosening making this surgical observation more evident in these cases, but this cannot be known with certainty.

Idiopathic loosening is generally a long-term phenomenon. Chemical toxicity from methymethacrylate monomer, thermal effects from the curing polymethyl methracrylate, allergic reactions to cement or metal, and excessive stresses or motion at the bone/cement interface are among the reasons given to explain the formation of a soft tissue liner at the bone/cement interface. Although the exact mechanism for tissue formation is not known, the reaction is essentially a biologic one; a bone response to the presence of the implant.

Figure 5

RADIOLUCENT LINE AROUND COMPONENT
METAL-on-PLASTIC COMPONENTS

	Non-Septic		Septic	
	Line	No Line	Line	No Line
acetabular component	34	15	11	1
femoral component	22	27	9	3

The second leading cause of hip prosthesis revision is direct trauma or falling (15/68). Such failures could and did occur at any time period, ranging from 1-7 years post-operatively with no increased incidence in any given year.

A commonly reported intermediate and long-term x-ray observation is a region or line of radio-lucency separating and demarcating bone from cement. Some reports in the literature have taken such a lucent line, particularily greater than 3 mm in thickness, to be evidence of prosthesis loosening (2). The lucent line was almost always observed in the septic cases and was seen in about half of the non-septic cases, Figure 5.

Figure 6

CROSSTABULATION: PRESENCE OF RADIOLUCENT LINE VS COMPONENT LOOSENING

	Acetabular Component			Femoral Component	
	loose	not loose		loose	not loose
line	6	28	line	14	8
no line	4	11	no line	17	10

However, a cross-tabulation of the study data, Figure 6, to correlate the incidence of a lucent line on x-ray with the clinically observed incidence of loosening does not support the conclusion that the presence of a lucent line indicates loosening of the prosthesis.

Figure 7
LOOSENED INTERFACE

	Metal/Metal		Metal/Plastic Non-Septic		Metal/Plastic Septic	
	Bone/ Cement	Cement/ Prosth.	Bone/ Cement	Cement/ Prosth.	Bone/ Cement	Cement/ Prosth.
acet. comp.	3	0	12	8	3	1
fem. comp.	2	4	15	37	3	7

Not all components were loose.

Figure 7 gives a tabulation of the loosened interface for the prostheses which were found to be loose at the time of revision. Whereas two-thirds of the acetabular components loosened at the bone/cement interface, the femoral components primarily loosened at the prosthesis/cement interface (70%). This observation suggests that the relatively smooth surface of the femoral stems currently in use may not be bonding well to cement and that perhaps provision should be made to enhance the interdigitation of cement with the femoral stem surface.

Figure 8
CEMENT INTERDIGITATION

	Metal/Metal		Metal/Plastic Non-Septic		Metal/Plastic Septic	
	Good	Poor	Good	Poor	Good	Poor
acet. comp.	0	7	2	17	2	8
fem. comp.	1	4	12	17	4	6

Examination of the cement from removed prostheses often indicates that poor or marginal bone/cement interdigitation was achieved at the time of the original surgery, Figure 8. This bone/cement bond is sometimes further weakened by the entrapment of biologic debris such as blood clots at the interface creating, in essence, cracks along the bonding interface. In such cases, the bone/cement bond is probably at increased risk of failure, leading to loosening of the prosthesis component. However, the current study data is unable to show any correlation between poor bone/cement interdigitation, as defined for this study, and the clinical observation of component loosening, Figure 9.

Figure 9

CROSSTABULATION: CEMENT INTERDIGITATION VS COMPONENT LOOSENING

Acetabular Component	good	poor	Femoral Component	good	poor
loose	0	7	loose	9	13
not loose	1	7	not loose	3	3

Aseptic cases only

CONCLUSIONS:

1) Except for those cases where the patients develop early sepsis in their joints or where a technical problem occurred at the time of surgery, total hip prostheses appear to be failing at a fairly constant rate with regards to time of use.

2) The major time-related problem associated with total hip replacement failure is idiopathic loosening. Multiple factors are involved in the mechanism of idiopathic loosening. Short of the manifestation of frank symptoms, progressive idiopathic loosening is difficult to diagnose reliably through radiographs. Two of these factors, namely poor bone/cement interdigitation and the development of a lucent line on radiographs have been examined. The current study data is unable to provide a correlation between these factors and loosening.

3) The second major cause of long-term THR failure is trauma. This is an area which may be beyon the realm of prosthesis redesign although it is conceivable that prostheses could be fashioned to withstand the heavy but infrequent shock of a fall. This observation supports the idea that stress level is a factor in component loosening.

4) Femoral stems loosen more often than the acetabular cups and on loosening, femoral stems often debond at the prosthesis/cement interface. These are design related problems. The problem of stem loosening requires a more detailed understanding of the stresses which occur with the prosthesis/cement/bone/soft tissue system. The problem of loosening at the metal/cement interface, however, may be addressed by providing a means which allows the cement to bond and lock onto the metal stem more securely.

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ACKNOWLEDGEMENT

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A COMPARISON OF FEMORAL COMPONENT DESIGNS IN TOTAL HIP ARTHROPLASTY

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ABSTRACT:

This paper presents the results of finite element analysis of femoral component designs in total hip arthroplasty. The analysis predicts the distribution of stress throughout the proximal femur-prosthesis structure. Data is reported on the maximum tensile stress in the component stem, longitudinal stress in the proximal medial femur, maximum compression within the cement and maximum tension within the cement. These stress quantities are compared for a number of contrasting prosthesis designs.

INTRODUCTION:

The orthopaedic surgeon has a wide choice of prosthetic components in total hip arthroplasty. Many of the components have potentially significant functional differences. There are, however, few comparisons from which the surgeon can objectively determine prosthesis of choice. Reports of clinical experience with particular prostheses inevitably contain variations in patient selection, rehabilitation, and surgical technique. The functional differences of the prosthetic components are masked by these clinical variables. Laboratory produced comparisons of prostheses' function, which are either experimental or theoretical in nature, can provide a basis for component comparison free from clinical variables.

This paper reports on a comparison of several femoral component designs in total hip arthroplasty. The components are compared on the basis of a finite element stress analysis. This analysis, which is mathematical and theoretical in nature, represents a considerably simplified and idealized representation of total hip arthroplasty. The analysis is, however, free from the mask of clinical variables. The prostheses studied and compared include many of the major popular available femoral components and a component of original design which has resulted from our experience in femoral component design analysis.

METHODS AND MATERIALS:

Three-dimensional finite element stress analysis has been widely applied to femoral prosthesis design analysis by these researchers and others (1,2,5,6,7). A description of the method, validity and accuracy of this modeling technique is presented in previous publications (2). The prosthesis, cement and femur in-vivo comprise a structure that is complex with respect to its geometric form, material properties and loading condition, Fig. 1. Practical engineering solutions to the stress analysis of the proximal femur-prosthesis structure can be obtained only after significant simplification of the structure, Fig. 2. A segment of femur 150 millimeters long of linearly elastic and isotropic material was represented. A region of cancellous bone was represented in the region of the greater trochanter. Cross-sections of the femur were symmetrical about a medial-lateral axis. The femoral component stem was modeled as an isotropic material with modulus of elasticity equal to 200 GPa (steel) or 100 GPa (titanium). Fig. 2 demonstrates the model developed and analyzed for a Charnley type of prosthesis. Similar models were developed for each of the prostheses studied. Within each model it was attempted to fatefully reproduce the shape and size of a "standard" component.

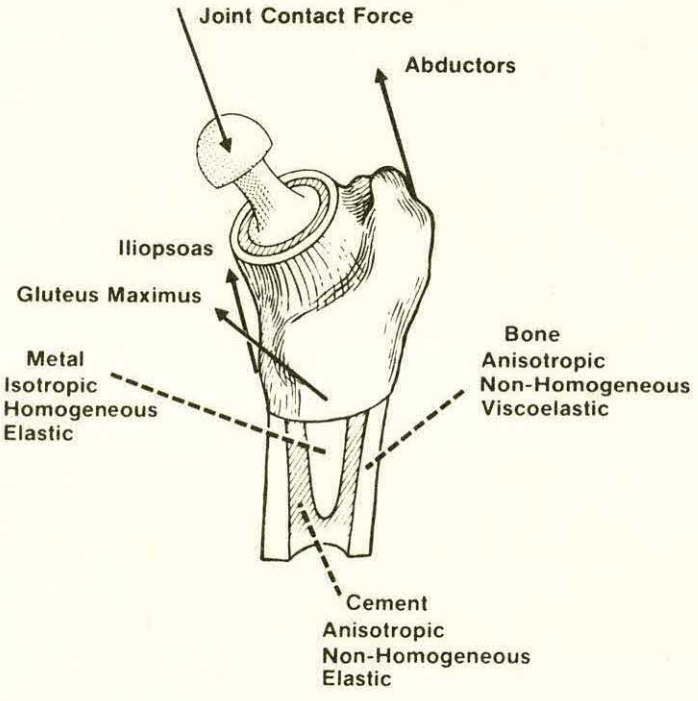

Figure 1

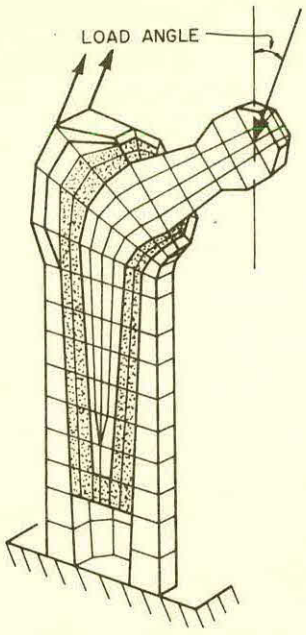

Figure 2

Previous work has indicated (3) that femoral component neck-femoral shaft angle and femoral component neck length significantly affect the forces generated during function at the femoral component head. The force present at the femoral head significantly affect the force and moment resultant present at the proximal stem-neck junction. These variable resultants will in turn produce different stress distributions within the proximal femur-prosthesis structure. The femoral component designs studied in this analysis were assumed to have identical force and moment resultants applied to the proximal stem-neck junction. This assumption makes it convenient to compare the function of the various stem designs, however, the in-vivo application of these devices may result in characteristically different femoral head force generation.

RESULTS:

The finite element analysis estimates for each prosthesis studied the state of stress throughout the proximal femur-cement and prosthesis. The prosthesis designs are compared on the basis of the longitudinal stress developed in the lateral aspect of the component stem, the longitudinal stress within the medial femoral cortex, and the principal stresses of tension and compression within the cement.

This investigation indicates that all of the prostheses studied result in significantly non-physiologic loading of the proximal medial femoral cortex, Fig. 3. At the most proximal level of the osteotomized femur, the longitudinal stress that is present as a result of the prosthesis loading, varies from about ten to sixteen percent of predicted physiologic levels. Although there is a sizable relative effect of the stress induced in this region of the femur with a different prosthesis studied, none of the prostheses resulted in stress levels that approached physiologic levels. With position distally along the medial femoral cortex, stress levels increased with all prosthesis types to approach physiologic levels. At any level along the femoral component stem, the STH prosthesis is seen to induce the greatest stress in the proximal medial femur, while the ITH prosthesis is seen to induce the least stress. This effect results directly from the relative rigidity of the femoral components.

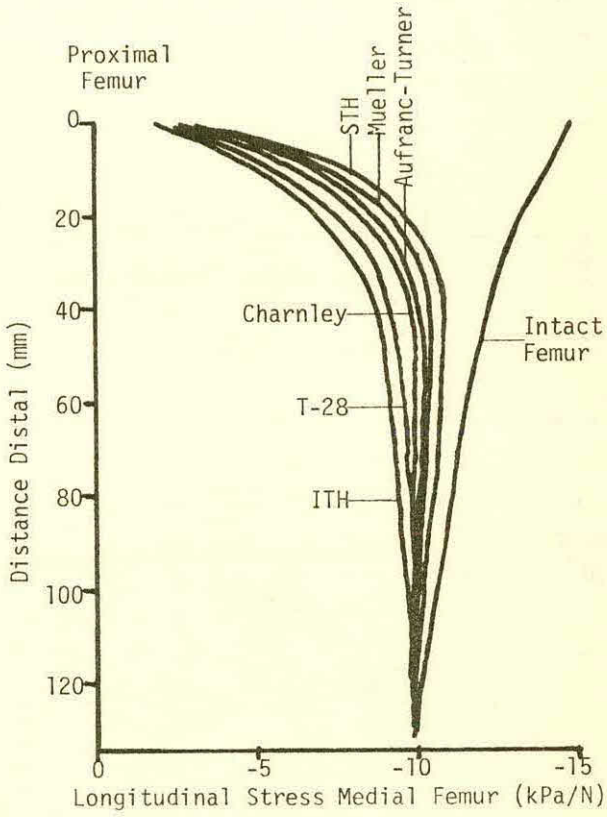

Figure 3

Femoral component design is seen to affect the longitudinal component or stress in the lateral aspect of the component stem, Fig. 4. The least rigid STH prosthesis (8) results in the lowest tensile stress in the component. The other prostheses in general experienced higher tensile stresses with the Mueller prosthesis experiencing the largest stress of those studied.

The maximum (principal) compressive stress occurring within the cement is seen to be significantly dependent upon prosthesis design, Fig. 5. There is a relative increase in compressive stress about both the proximal and distal portions of the femoral component stem. Stress levels through the mid-length of the stem are considerably lower. In generaly, prostheses with shorter stems and with lower rigidity experienced higher compressive stresses in the proximal region. The compressive

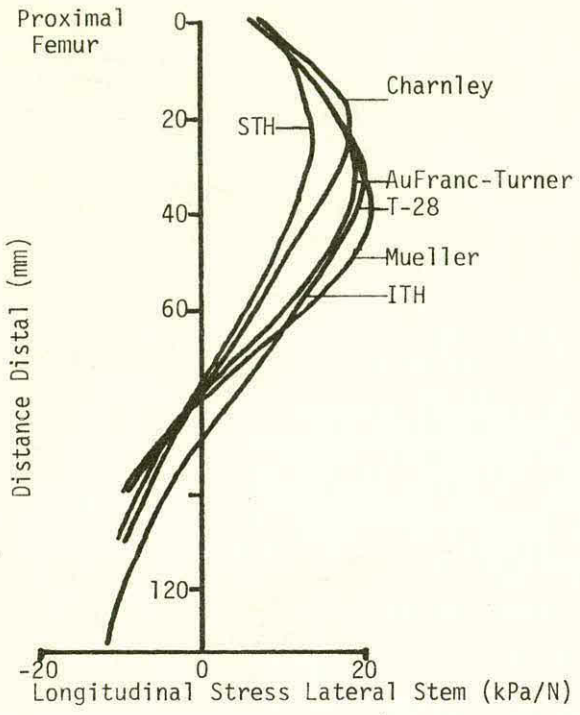

Figure 4

stress within the proximal cement was greatest with the STH prosthesis while this stress was lowest with a longer, more rigid ITH prosthesis. The maximum compressive stresses occurring about the distal tip of the prostheses were less affected by component design. The location of the maximum compressive stresses was, however, displaced more distally with the longer stem prosthesis.

The maximum (principal) tensile stress within the cement is affected by prosthesis design in a manner similar to that of the compressive stress, Fig. 6. Relative maximums are seen to occur both about the proximal and distal portions of the component stem. The STH prosthesis resulted in the greatest tensile stress in the proximal region of the cement, however, this prosthesis resulted in the lowest tensile stress in the distal region of cement. The tensile stress in the proximal region of the cement that is associated with the ITH prosthesis is the lowest of those studied. The maximum tensile stress within the cement about the distal tips of the prostheses is displaced more distally with the longer component stems.

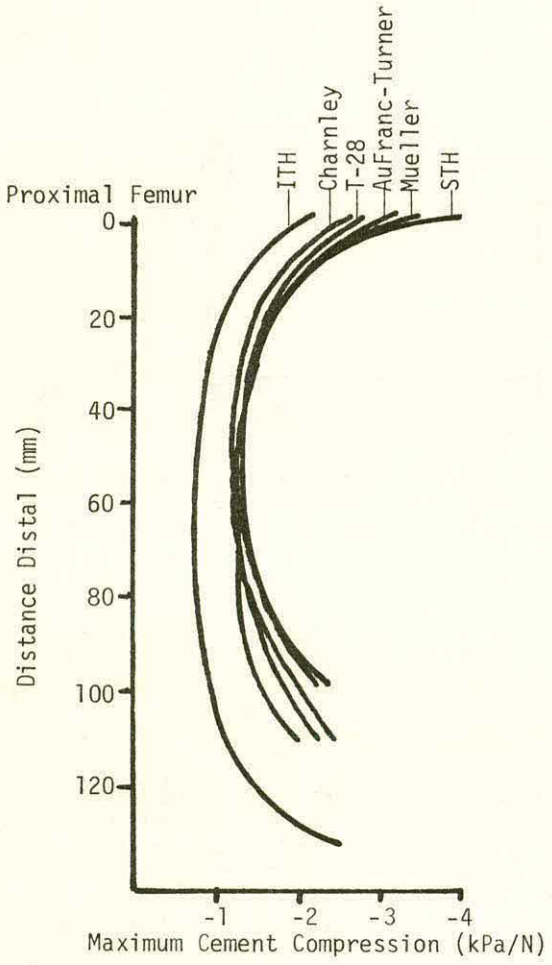

Figure 5

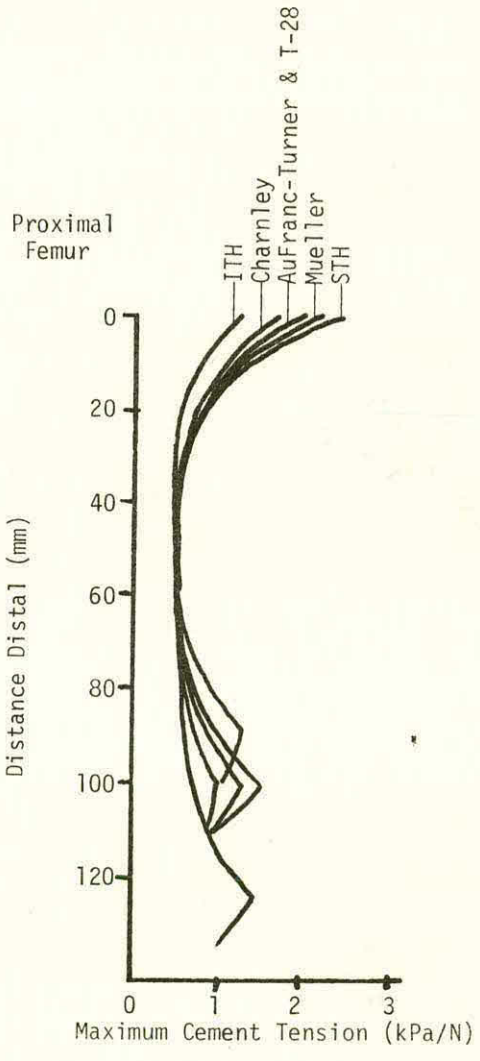

Figure 6

DISCUSSION:

This papers provides a comparison of the function of several femoral component types. The method of this analysis, which is mathematical and theoretical in nature, is free from variations in patient selection, rehabilitation, and surgical techniques that are inevitably present in clinical studies. Although the method of this report is objective and quantitative in nature, it has inherent limitations. Finite element stress analysis is an approximately method to calculate stress states. Its accuracy is dependent upon the degree to which the numerical model represents the real structure. Specific limitations and numerical accuracies of this technique of analysis have been extensively discussed in other publications (1,2). Although the results obtained by this general method of investigation are free from the mask of clinical variables, they are also without the definitive proof that may be obtained through long-term clinical studies. Studies of this type described in this paper are thus able to provide objective and quantitative information of surgical technique and device design. This information may be useful in designing improvements in surgical treatment which may be proven by clinical experience.

The prostheses studied in this investigation are all of the type that are intended to function primarily without proximal support of the prostheses through devices such as proximal medial collars. Modeling studies (2,4) have indicated that actual support of the femoral component through a collar on the proximal medial femoral cortex significantly affects the distribution of stress throughout the proximal femur, cement and prosthesis. Prostheses of this type were omitted from this study due to the concern of these investigators that clinical experience has yet to prove that collars, which in theory affect prosthesis function, do in actuality predictably obtain and maintain proximal medial femoral cortex support.

With the advent of improved manufacturing techniques and stronger new materials, considerations of the stress levels occurring within the femoral component stem need not be a dominant factor in component selection. In the absence of component loosening, fracture of the femoral component stem is highly unlikely, if ever, to occur, while in the presence of proximal stem loosening, the prevention of component fracture is not likely affected by femoral component selection.

The commonly seen remodeling of the medial proximal femoral cortex is logically the result of the significantly altered stress distribution within this region of the femur. The prosthesis type studied, although affecting stress levels, without exception resulted in substantially non-physiologic levels of stress within the region of the proximal femur. These authors feel that the changes in predicted stresses seen with the prostheses studied will unlikely affect this remodeling process. The selection from these components will not likely change the clinical evidence of proximal medial femoral cortex remodeling.

The change in the stress levels seen within the cement, particularly within the proximal region, is of concern to these authors. It is thought that the acrylic cement and its bond to the femoral component stem and the femur is the weakest link in present reconstructions of the hip. The relative increase in cement stress seen with some component designs may lead to premature loosening and failure of these components. It is believed that efforts intended to reduce stress levels within the cement are well advised. Component loosening may thus be forestalled. The components which result in generally reduced stresses within the cement may experience higher stresses within the femoral component stem and reduced stresses in the proximal medial femoral cortex. However, on balance this trade of lower cement stresses for high stem and lower femur stresses, may result in improved component longevity.

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COMPARATIVE EVALUATION OF TOTAL KNEE PROSTHESIS DESIGNS

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ABSTRACT

Total joint replacements are highly successful surgical procedures, but a significant number of failures occur. In particular, tibial component loosening is a major problem with total knee replacements. Finite element stress analysis is used to compare several tibial component designs based upon stress levels. One piece, metal components are found to be preferable to all polyethlene components or designs with separate condylar pieces. The single, central post design results in the lowest stress, but its use requires sacrifice of the anterior cruciate ligament. Of the designs compatible with retention of the anterior cruciate ligament, the three post design is found to be superior. The value of retaining the anterior cruciate ligament to tibial component fixation and the effect of non-perfect interface conditions on the stresses require further study.

INTRODUCTION

Arthritis, in its many forms, affects seventy percent of the adult population including ninety percent of those over sixty-five years of age. It is a severely disabling condition that results in pain, limited motion, decreased mobility, and dependency on others. It decreases the quality and enjoyment of life for these people physically, emotionally and economically; and speeds the development of bone, soft tissue, respiratory and vascular disorders. Thus, the rehabilitation of these patients through the relief of symptoms and return of function is an important medical and social goal.

The development of total joint replacements of the hip and of the knee has provided procedures for the surgical management of patients with severe arthritis of these joints. The pain is eliminated, motion is increased, and mobility is restored, in most cases, to a level of comfortable, independent living. The early, post-surgical mobilization that is possible with these procedures shortens the hospitalization and prevents complications associated with long, bedridden convalescences. These advantages not only aid the arthritic patient, but are also very valuable in the treatment of elderly trauma patients. It is estimated that 90,000 hips and 50,000 knees are replaced each year in the United States[1]. The use of these devices has greatly increased the available options and the effectiveness of medical treatment, and has proved to be an effective rehabilitation tool.

TOTAL KNEE PROSTHESIS RESEARCH

Total joint replacements are highly successful, but a significant number of complications and clinical failures do occur. In many cases, these complications lead to surgical revisions which are costly, medically dangerous, and destine the patient to a protracted period of hospitalization, convalescence, therapy and functional dependency.

Loosening of the tibial component, by separation of the cement-bone interface around it, is a primary cause of clinical failure and revision of total knee prostheses [2]. Many of these cases involve trauma or excessive functional demands upon the prosthesis, suggesting that direct mechanical failure due to high stress levels at the cement-bone interface is a cause in a significant number of tibial component loosenings. The prevention of such failures through prosthesis design is the goal of bio-engineering research on knee prostheses.

A variety of theoretical, experimental and clinical approaches have been taken in the past to knee prosthesis design problems. As a result, a great number of knee prosthesis designs have been proposed. Metal and plastic designs with none, one, two or more fixation posts on connected or separated tibial condylar components have appeared. Some of these designs allow retention of one or both of the cruciate ligaments; others do not. Short of clinical trials, which are the most dangerous and the least acceptable design screening method, there has existed no convenient way to compare these designs.

The finite element method provides a technique for examination of prosthesis designs with a speed and a control not offered by clinical trials or experimental methods. Thus, comparative evaluations of prosthesis designs can be quickly obtained.

In this paper, three-dimensional, finite element stress analyses are used to develop a comparative evaluation of basic knee prosthesis designs.

FINITE ELEMENT STRESS ANALYSIS OF TIBIAL COMPONENTS

The basic three-dimensional finite element

model of a tibial knee component implanted in the proximal tibia with PMMA bone cement is shown in Figure [1]. The bone model consists of cancellous bone of varying stiffness surrounded by a cortical bone shell of near anatomical shape. The external and internal geometries of the model were determined from a serially sectioned defatted tibia. The material properties of the internal cancellous bone were assumed to vary linearly with the area fraction of cancellous bone as measured in the sectioned tibia. All materials in the model were assumed to be linearly isotropic, and all material interfaces were assumed to be perfectly bonded. The model was rigidly constrained at the distal surface of the cortical shell. The finite element model consists of 1053 nodal points and 768 eight-noded, isoparametric, parallelopiped elements. The model and the mesh were defined, generated and refined using an interactive computer graphics package, NUFIG3D, developed and maintained by the Rehabilitation Engineering Program of Northwestern University [4].

The geometries of six tibial component configurations, shown in Figure 2, were generated within the bone model. Both metal and polyethylene components were considered. Six load cases, shown in Figure 3, were considered for each design.

FINITE ELEMENT RESULTS

The structural elements of the implanted prosthetic component system that appear, from clinical experience [2], to be most vulnerable to mechanical failure are: (1) the PMMA-bone interfaces beneath the plate and around the posts of the component, (2) the PMMA, and (3) the cancellous bone. There is not yet sufficient information available to say if failure of any of these elements will or will not lead to failure of the system as a whole. For the comparative evaluation purposes of this paper, each of the structural features will be considered independently with simple critical stress failure criteria.

Results for the tension at the PMMA-bone interface beneath the plate, the compression in the cancellous bone, the tension in the PMMA and the shear at the PMMA-bone interface along the posts were obtained for a load of 2000N (3 x body weight) for each load case and for each design. Each stress result was expressed as a fraction of the appropriate failure stress. It is not known which of the load cases is the most critical for each of the designs. Thus, in order to compare designs based upon convenient parameters, four basic indexes, formed by averaging the considered stresses over all the load cases, were developed, Figure 2.

The results of the analyses are used to discuss three problems of tibial component design. Should the tibial component be all polyethylene or should plastic articulating inserts on metal components be used? Should the tibial component condylar components be separate or connected? What is the optimum fixation post configuration?

METAL OR POLYETHYLENE COMPONENTS: Polyethylene components result in higher cancellous bone compressive stresses and higher PMMA tensile stresses than do metal components. In some polyethylene cases these stresses are near their critical values. Metal components exhibit higher tensile "tilting" stresses due to loads at the extreme edges of the component than do polyethylene components. However, polyethylene components show tensile stresses due to receding contact or "wing up" phenomena that are, in some cases, greater than the tensile "tilting" stresses seen in the metal components. Metal components appear superior to polyethylene components based on stress levels.

SEPARATE OR CONNECTED COMPONENTS: Higher cancellous bone compressive stresses, tensile "tilting" stresses, post interface shear stresses and PMMA tensile stresses are seen with the separate component designs, in both metal and plastic, than are seen in the corresponding connected configurations. The tension beneath the plate of the separate component designs may exceed the failure stress level. On the basis of these stress results, connected components are superior to separated components.

FIXATION POST CONFIGURATION: Considering only the metal posted components, the four stress indexes shown in Figure 2 indicate that the single-post design provides the lowest stress conditions. The two-post design and the large post, three-post design are next with little difference between them, followed by the small-post, three-post design. Of the metal components, the separated component design shows the highest stress levels.

It is also seen that the compressive cancellous bone stresses and the cement-bone interface tensile stresses, which are critically important to the integrity of the no-post design, are higher in that design than they are in the small-post, three-post design.

In the one-post and two-post configurations, a rotation axis appears to be formed by the stiff posts, resulting in higher "tilting" stress. Since the three-post design does not promote such axis, the three-post design would appear to be superior to the two-post configuration.

Of the configurations considered, only the no-post, the separated two-post, and the small post, three-post designs allow the retention of the anterior cruciate ligament. An intact anterior cruciate ligament limits the surgical working space so that large posts cannot be used. Of these three configurations, the small-post, three post design is superior based upon stress and fixation considerations. A comparison with the single-post design shows the latter to have lower stresses. However, the anterior cruciate ligament must be sacrificed in order to use the single-post design. The anterior cruciate ligament restricts the joint laxity and prevents the extreme edge landings which cause the highest stresses. Thus, the surface loadings may not be so severe with the anterior cruciate ligament present. Whether this load reduction will cause lower stresses in the anterior cruciate retaining design than in the single-post design has not yet been established.

CLOSURE

In applications to total joint prosthesis design the general numerical and modeling accuracy of the finite element method can be experimentally demonstrated. However, the comparison of implant designs based upon stress levels is not without deficiencies. The stresses of interest are those at or near the PMMA-bone interface, which are dependent upon the mechanical conditions at that interface. In the finite element model this interface is assumed to be perfectly bonded. This condition almost certainly does not occur in clinical applications. Voids and cracks occur which may propagate under load. What effect such discontinuities will have upon the stresses and the evaluation of designs presently considered is not known. Such discussion will have to await greater knowledge of the clinical interface conditions and their impact on the model. The finite element method as presently used remains the best available approach to the evaluation of the total joint replacement design.

ACKNOWLEDGEMENT

This research was supported by NIHR Grant No. 23-P-55898.

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FIGURE 3. LOAD CASES

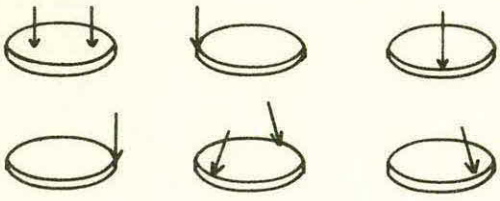


FIGURE 1. FINITE ELEMENT MODEL

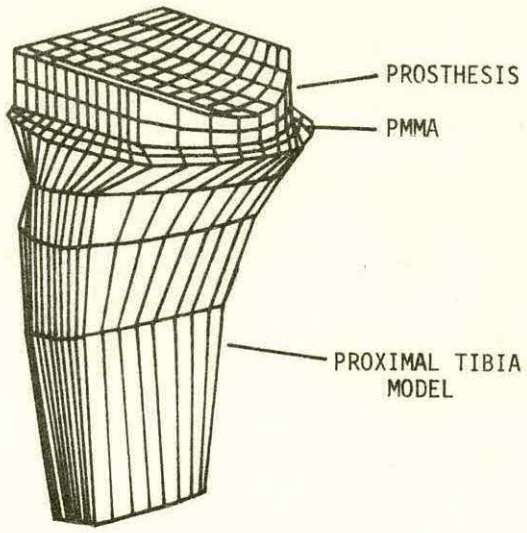
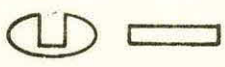
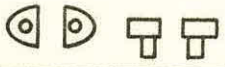
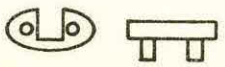
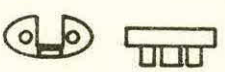
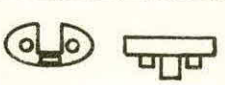
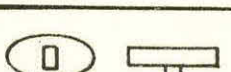


FIGURE 2. STRESS LEVEL INDEXES

CONFIGURATION	MATERIAL	A	B	C	D
	P	0.56	0.55		0.21
	M	0.27	0.65		0.07
	M	0.42	1.33	0.43	
	M	0.24	0.49	0.40	0.06
	M	0.24	0.47	0.42	0.07
	M	0.22	0.58	0.62	0.05
	P	0.47	0.49	0.30	0.21
	M	0.24	0.35	0.47	0.06

M - METAL
P - POLYETHYLENE

A - CANCELLOUS BONE COMPRESSION
B - PMMA-BONE INTERFACE TENSION UNDER PLATE
C - PMMA-BONE INTERFACE SHEAR ALONG POST
D - PMMA TENSION

THE INFLUENCE OF TOTAL HIP AND KNEE REPLACEMENT ON GAIT

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ABSTRACT

Forty-two patients following total hip or total knee joint replacement surgery were observed while walking. The parameters observed included basic time distance measurements, joint motion and joint moments. The patients gait was compared to a group of fourteen aged matched normals. It was found that hip patients recover a more normal gait than knee patients. The gait abnormality in knee patients had several common features, these included a shorter than normal stride length, a lower than normal flexion at the knee during mid-stance and abnormal pattern of flexion-extension moment at the knee. The abnormal gait in knee patients seemed to be indicative of a different than normal use of the extensor mechanism of the knee. Gait abnormalities were also found in hip patients based on the location of the reconstructed joint center. Hip patients that had the distance between the hip joint center and the line of action of the abductor muscles shortened walked normally, while patients that did not have this distance shortened had a shorter than normal stride length, reduced flexion motion and a higher than normal hip flexion moment.

INTRODUCTION

An understanding of the influence of joint reconstruction on patients function is fundamental to future improvements in these procedures. Quantitative gait analysis provides a means of objectively studying and evaluating the results of joint reconstruction in the lower extremities. In addition, this type of analysis will help determine the role of the design of the prosthesis in the restoration of normal function during walking and other activities of daily living. In general, there has been a trade-off between the restoration of normal anatomy for functional purposes and structural design of the implant devices. The purpose of this report is to demonstrate the use of quantitative gait analysis in the study of patients following total knee replacement and total hip replacement.

While there has been a number of important studies describing the gait of patients with joint replacement, there is still a need to identify basic biomechanical factors relating the mechanics of joint replacement design to patient function. The purpose of this paper is to identify gait abnormalities in patients following hip and knee joint replacement and to discuss the implication of these gait abnormalities to prostheses design and usage.

MATERIALS AND METHODS

Forty-two patients following joint replacement surgery were observed in the gait laboratory. Twenty-two patients had total knee replacement and twenty had total hip replacement. Subjects were selected on the basis of a successful clinical result and observed a minimum of twelve months following surgery. Each subject was observed bilaterally during level walking over a range of walking speeds. A clinical examination was taken at the time of the gait observation which included an assessment of deformity, joint stability, passive range of motion, and a subjective assessment of pain. The gait parameters observed included time distance measurements, joint range of motion, and three components of moments at the hip, knee, and ankle. In addition to the patient population, a group of fourteen normal subjects which were aged matched to the joint replacement group were observed.

The total knee replacement patient had one of four different type design of total knee replacement. Seven patients had Geometric design, seven had Gunston, seven had Duo-Patellar, eight had Total Condylar with patellar prosthesis. The total hip replacement patients were all treated with Mueller design prosthesis. The normal subjects were all observed following the same protocol.

Instrumentation included an optoelectronic system for motion analysis, a multi-component piezo electric force platform and a mini-computer based system for data acquisition and processing. Limb motion measurements were obtained by placing light emitting diodes (LED's) on the pelvis, greater trochanter, center of the lateral joint line of the knee, lateral malleolus, lateral aspect of the calcaneous, and the base of fifth metatarsal. The position of these light emitting diodes were tracked in three dimensional space using the optoelectronic digitizer. This positional information was acquired and processed at a frame rate of 75 samples per second using the digital computer. The multicomponent force platform was used to measure the three components of foot ground reaction force, the location of the center of pressure, and the vertical twisting moment beneath the foot. The motion and force information obtained simultaneously were used to determine the net resultant forces and moments at the ankle, knee and hip joints during support and swing phase. The force and moments vectors were

broken into components aligned with axes in the direction tending to produce flexion-extension, abduction-adduction, or internal-external rotation.

RESULTS

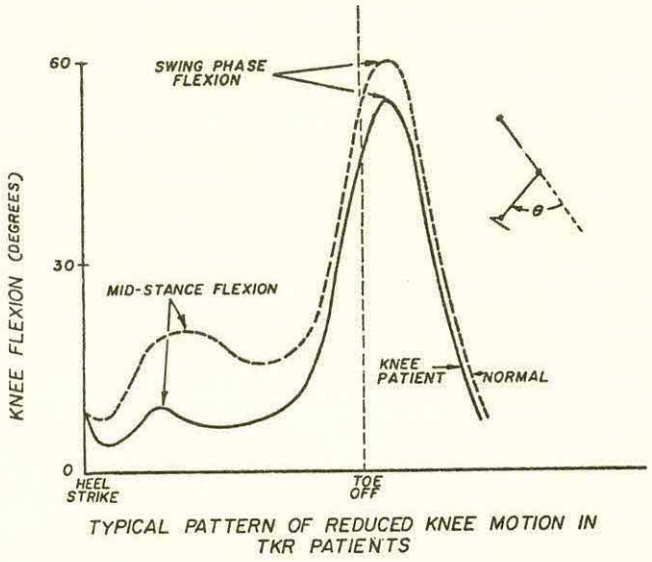
All gait measurements were compared at approximately the same walking speed (1 meter per second). A comparison of time distance parameters indicates that the patients with total knee replacement had a significantly shorter stride length (1.16 meters s.d. = 0.07) than normal subjects (1.28 meters s.d. = 0.12) while the hip patients stride length was normal. These results indicate that patients following total hip replacement surgery have a gait which more approximates that of normal than patients following total knee replacement surgery.

It is difficult to infer the cause for gait differences among knee patients and hip patients on the basis of only time-distance measurements. To analyze the cause of these gait variations among joint replacement patients, joint motion and net reaction moments at the hip, knee, and ankle joint for each group were observed. The following section describe the results for the knee patients and the hip patients separately.

Knee Patient Gait Observations

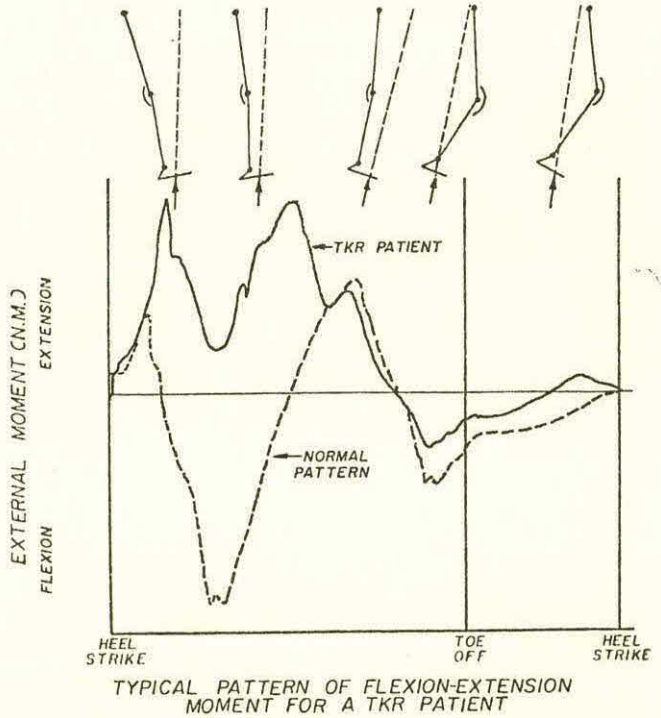
The range and pattern of knee flexion extension moments during support and swing phases was compared for normal subjects and knee patients. Typically, a normal subject will strike the ground with his knee near full extension, flex his knees to approximately 20^0 during mid-stance re-extend the knee just past mid-stance, and continue to flex the knee through toe-off and into swing phase (Figure 1). During swing phase, a normal subject will flex his knee approximately 65^0. The pattern and range of knee flexion-extension motion during support and swing phase was found to be different in patients following total knee replacement (Figure 1). Knee patients did not flex their knee the same amount as normal subjects during midstance. In addition, the amount of flexion during swing phase was reduced. In general, patients following total knee replacement use less flexion in mid-stance (6.8^0) than normals (20^0) and less flexion in swing phase (48.6^0) compared to (63.6^0) for normals. Thus, knee patients tend to walk with their knee near full extension throughout support phase.

The knee patients were also observed to have an abnormal pattern of flexion extension moment at the knee joint. The normal pattern illustrated in (Figure 2) cycles from extension to flexion, from flexion to extension just prior to toe-off. The knee patients walked with a gait that tends to produce an external moment in the extension direction almost the entire period of support phase. Thus, by walking with a shorten stride-length, reduced mid-stance flexion, the knee patients have acquired a gait which tends to minimize the use of the extensor muscle groups during walking. This observation was consistent among all the knee patient observations and was independent of prostheses design.



TYPICAL PATTERN OF REDUCED KNEE MOTION IN TKR PATIENTS

Figure 1: Knee flexion-extension motion for a normal and knee patient.



TYPICAL PATTERN OF FLEXION-EXTENSION MOMENT FOR A TKR PATIENT

Figure 2: An illustration of the differences in resultant flexion-extension moment at the knee for a normal and knee patient.

Hip Patients Gait Observations

As indicated previously, hip patients were found to have a more normal gait than knee patients. Both the patterns and magnitudes of the moments and motions were approximately normal for this group. However, gait differences could be observed when hip patients were separated on the basis of the location of the joint center. Two groups were identified: The first group had a joint center location such that the distance from the center of the joint to the line of action of the abductor muscle group was shorter than the unoperated side by an average of 0.9 centimeters, while the other group had essentially the same length as the unoperated side. This measurement was designated as the abductor moment arm. These measurements were determined from frontal plane post-operative x-rays.

The group of patients with the unshortened abductor arm had a lower than normal range of flexion extension motion at the hip joint. The average flexion-extension motion for this group was 27^0 compared to 40^0 in normals. Twenty-seven degrees of motion also significantly differed from the 38^0 of motion in the reconstructed hip of the group with the shortened abductor moment arm.

Both groups of hip patients had a normal pattern of flexion-extension moment, during support phase. Typically, the pattern of flexion-extension moment at the hip is sinusoidal in character, changing from a peak flexion moment after heel strike through zero at midstance, to a peak extension moment past midstance and back to zero at toe-off (Figure 3).
However, the magnitude of the peak flexion moment just after heel strike was statistically higher in the patient group with the unshortened abductor moment arm.

The gait abnormality in this group seems to be associated with a change in the action of the flexor and extensor muscle groups at the hip joint. It is interesting to note that no statiscally significant difference in the magnitude of the abduction-adduction moments could be correlated to the difference in abductor moment arms of the two groups.

Discussion

Gait analysis can be useful in detecting gait abnormalities in patients following total joint replacement. The relevance of this type of information for clinical evaluation is still an open question, since there are many other measures that can be used for patient evaluation. However, research gait studies of joint replacement patients do provide an understanding of the relevance of various gait measurements in evaluating this type of patient. If clinical evaluation using gait analysis becomes feasible at some point, this information from research studies will be needed. In addition, biomechanical data describing forces and motion at the joints is needed for input into future designs of prosthetic components

The time-distance parameters including lengths of stride, walking speed, time of swing, and time of support are among the simplest to measure and the most reproducible.

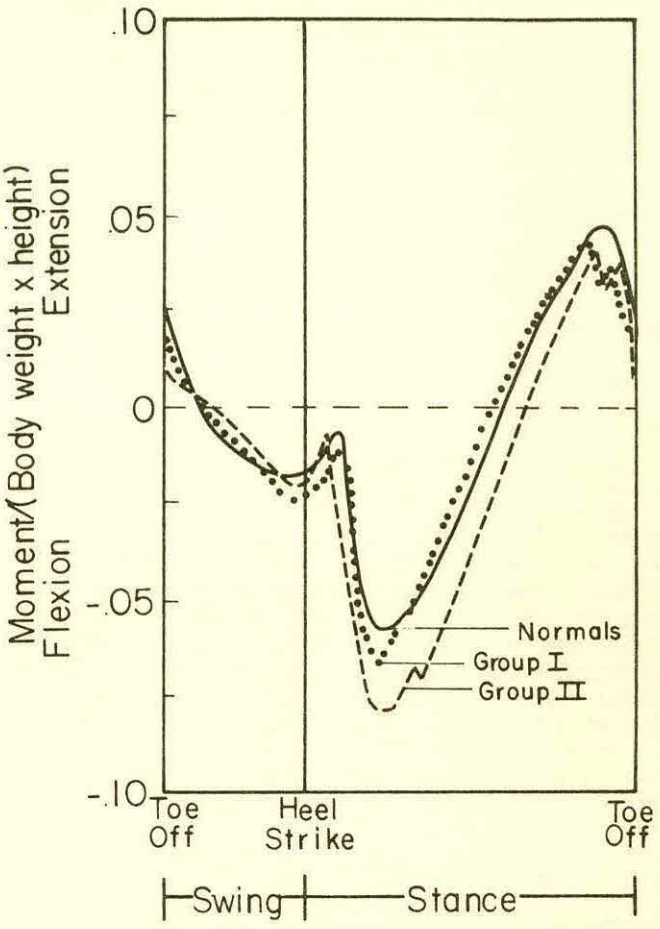


Figure 3: An illustration of the differences in hip flexion extension moment for normals, patients with shortened abductor moment arms (Group I) and patients with unshortened abductor moment arms (Group II).

Although these parameters are quite sensitive indicators of gait alterations by themselves, they do not indicate the cause of gait abnormality. Range of motion combined with time-distance measurement provide an additional bit of information relating to the source of the abnormality. However, a more d scriptive and precise parameter can be calculated by combining three dimensional positions of the joints with measurements of ground reaction force to compute the external moment acting at the joint throughout the walking cycle.

External moments are the net effect of all the external forces acting on the limb segments tending to produce rotation about a particular joint. These external forces are primarily the weight of the body and the inertial effects of the limb segment accelerating and decelerating throughout the walking cycle. The laws of mechanics require that external moments at the joints must be equilibrated by internal moments that are equal in magnitude and opposite in direction.

The internal moments are generated primarily by muscle forces acting at a distance from the joint center. For example, an external moment tending to flex the hip joint would require an extensor muscle group force to maintain equilibrium. Thus, it can be infered that the magnitude and direction of the external moments are indicative of muscle action.

In the material reported here, it was shown that knee patients as a group tend to walk differently than normals. Their gait abnormality could be identified by a shortened stride length at the same walking speed as the normal subject. It was found that by observing the joint motion and moments at the knee joint that this gait abnormality seems to be indicative of the knee patients not using their knee extensor mechanism.

Similarily, it was found that hip patients while having a more normal gait than knee patients can have their gait abnormalities related to the location of the joint center relative to the line of action of the hip abductor muscles. It seems that the group with a shortened distance of the length to the abductor moment arm is favorable from the view point of function during gait. The gait abnormality in this group seems to be associated with a higher than normal moment tending to flex the hip. Again, assuming that the external moments are balanced by internal moments generated by the muscles, we can infer that the higher the normal external flexion moment at the hip joint is indicative of higher than normal forces in the extensor mechanism at the hip to maintain equilibrium.

The findings presented in this report have several implications. It appears that quantitative gait analysis can be used in detecting gait abnormalities in patients following hip or knee replacement surgery. In general, hip patients will recover a more normal gait than knee patients. The difference in gait between the hip and knee patients can be observed by a simple comparison of stride length at the same walking speed. Thus, time-distance measurements are useful indicators for evaluating gait abnormalities. However, for a basic understanding of the cause and effect of gait abnormalities, parameters such as the joint angular motion and moments are needed. In studies presented here abnormal gait patterns were associated with abnormal motion and moments at the involved joints. The abnormal moments are related to variation in muscle action at the involved joint. While it is beyond the scope of this report to examine the causes for these gait abnormalities; there seems to be a mechanical cause and there is definitely a mechanical effect. One effect of these gait abnormalities is that forces at the joints in both the hip patient and knee patient groups will be different than normal. Both the magnitude and direction of the resultant loads on the prosthetic components will be different than those generated during normal gait. The analysis and design of current devices has been based on normal gait patterns. Thus, design criteria may be inappropriate for the type of gait occuring in patients.

Acknowledgements:

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FUNCTIONAL EVALUATION IN UPPER EXTREMITY TOTAL JOINT REPLACEMENT PATIENTS

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ABSTRACT

Objective, quantitative evaluation of joint function is available for joints of the hand, elbow, and shoulder. Isometric and isokinetic strength is measured with electromechanical devices. Range of motion measurements with conventional hand goniometry is augmented by a three-dimensional analysis which uses a biplanar radiographic technique. Functional ability is quantitated with peg board tests and simulated activities of daily living. A system to monitor rectified integrated EMG activity defines muscle function about the joints. This system is now being used to monitor total joint arthroplasty patients pre- and postoperatively, to define normal function, and to provide kinetic and kinematic data for the design of prosthetic devices.

INTRODUCTION

The study of normal and abnormal joint function plays a vital role in the management of total joint arthroplasty patients involving the upper extremity. Areas of objective evaluation necessary to adequately define joint function include measurements of strength, range and pattern of motion during activity, boney orientation, functional ability and quantitative EMG. These measurements provide information necessary for monitoring patients' improvement following joint replacement and throughout their rehabilitation, as well as for the design of prostheses.

A system has been established which provides objective measurement of these important parameters for joints of the hand, elbow and shoulder. Initially, each subject, normal or patient, is interviewed for information regarding their history, level of activity or exercise, pain in the involved or adjacent joints, and surgical status. Anthropometric measurements are taken and joint stability assessed. A series of objective, quantitative measurements are then carried out based on the specific joint being studied.

HAND

Patients undergoing total joint arthroplasty of the hand are assessed in the areas of strength, range of motion, boney orientation, and dexterity. The equipment is shown in Fig. 1.

Hand strength is measured with a group of devices consisting of properly calibrated strain-gauged metal beams capable of registering strength values displayed on a digital readout meter (1). This apparatus, which allows measurement of pinch, grasp, and lateral deviation, quickly gives an accurate measurement and is readily adaptable to a variety of hand sizes and deformities. Dexterity is quantitated by the subjects' performance on a timed peg board test (2). Test results are transmitted to a PDP 11/34 digital computer via A/D, D/D channels for on-line data analysis.

Fig. 1. Hand evaluation equipment: (A) Strength meters, (B) Peg board, (C) EMG amplifier, (D) Computer terminal for data retrieval and analysis.

Range of motion is measured with hand goniometers, and the three-dimensional orientation of the joints is established with a biplanar radiographic technique (3). Small metal "T"-shaped markers are attached to the dorsal surface of the hand over the appropriate boney segments for the joints being studied. The hand is placed within a plexiglas frame with an imbedded metal grid system (Fig. 2). Horizontal and vertical x-rays are taken of the hand in positions which define the joints' extreme range of motion and daily activities. These x-rays are reduced with a sonic digitizer. The computer is used to calculate the three-dimensional joint orientations and range of motion.

A system for quantitative EMG analysis is being used to study muscle activity in a variety of hand functions in normal individuals. Fine wire electrodes are inserted in the muscles being

studied. Eight muscles plus the readout from strain-gauge strength meters can be simultaneously monitored. As the subjects perform specific functions, a strip-chart recorder records the raw data, and the computer records the rectified, integrated EMG signal plus the total external force (Fig. 3). As previously documented by Cnockaert, et al (4), this integrated rectified EMG was found to be proportional to the isometric tension developed. This result provides valuable information for understanding normal and pathologic joint function.

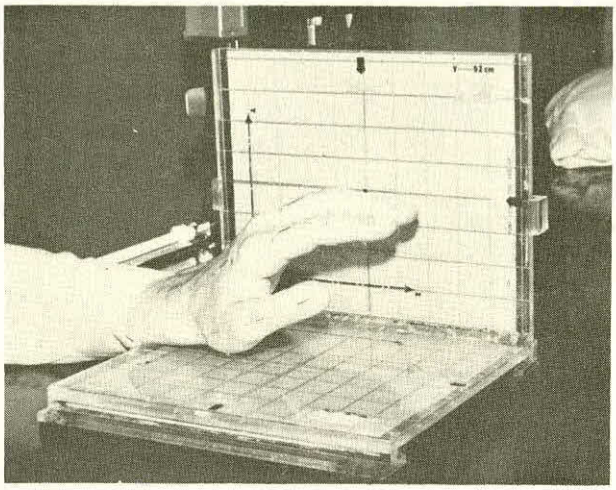

Fig. 2. Hand positioned for biplanar radiographs.

One group of patients now being studied have undergone total joint arthroplasty of the thumb CMC joint. To date, 28 patients of a total of 40 have been evaluated 1 to 2 years postoperatively. Strength in grasp, 3 types of pinch, and thumb abduction and adduction is measured. The subjects' forearms and wrists are stabilized in orthoplast splints to standardize the testing procedure without limiting the specific functional activities. Their dexterity is assessed with the timed peg board test, and hand goniometry measurements are

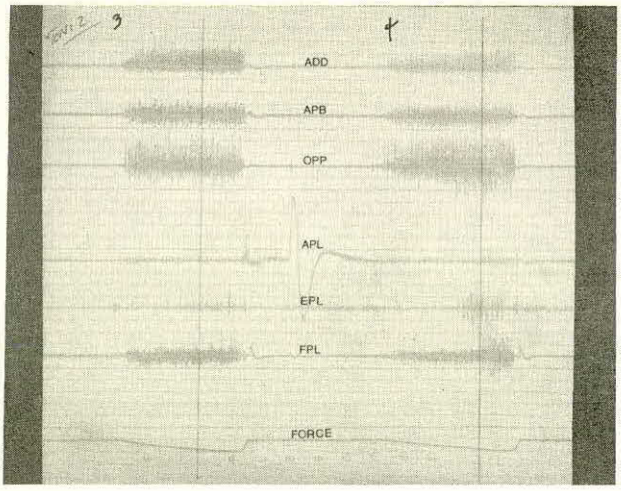

Fig. 3. Raw EMG and force recordings.

taken for active and passive movement of the thumb IP, MCP, and CMC joints. The three-dimensional range of motion occurring at the thumb CMC is determined using the biplanar x-ray technique. This technique must be utilized since such information is unavailable based on regular methods of measurement.

Preliminary results for this group of patients show their mean strength measurements to be 50 to 65 percent of normal. Time required for the peg board test is 60 percent greater than normal. Range of motion at the CMC joint, based on an axis system defined by the actual joint surfaces (5), is 60 to 70 percent of normal (Fig. 4).

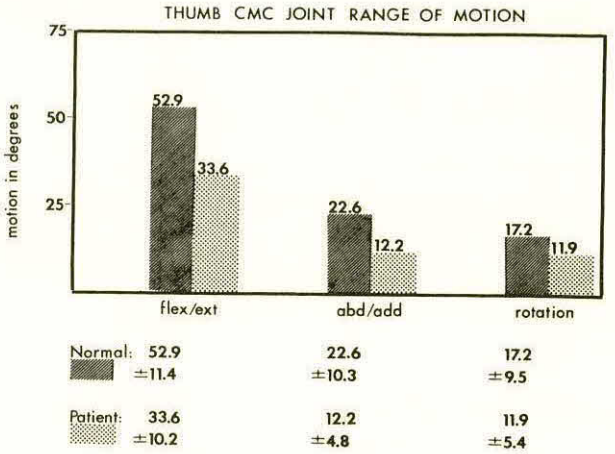

Fig. 4. Range of motion following CMC joint arthroplasty.

ELBOW

Evaluation of elbow function is based on measurements of strength, range of motion, and functional activities. Isometric strength of elbow flexion, extension, pronation and supination is measured with torque cell dynamometers with the measurements displayed on a digital readout device (Fig. 5).

A triaxial electrogoniometer is employed to simultaneously monitor the three-dimensional motion occurring at the elbow (Fig. 6). This device consists of three mutually orthogonal potentiometers attached to the arm and forearm to record the motions of flexion-extension, pronation-supination, and carrying angle changes (6). The analog output of elbow motion measured by the electrogoniometer in different functions is shown in Fig. 7. This electrogoniometer is used to evaluate the maximal range of motion available in normal and patient populations. With a specially designed function board, a variety of simulated activities of daily living are measured to define the necessary joint motion requirements in order to perform these functions.

This system is used to evaluate patients undergoing total joint arthroplasty of the elbow, pre- and postoperatively. Also, a study has recently been completed which documented normal

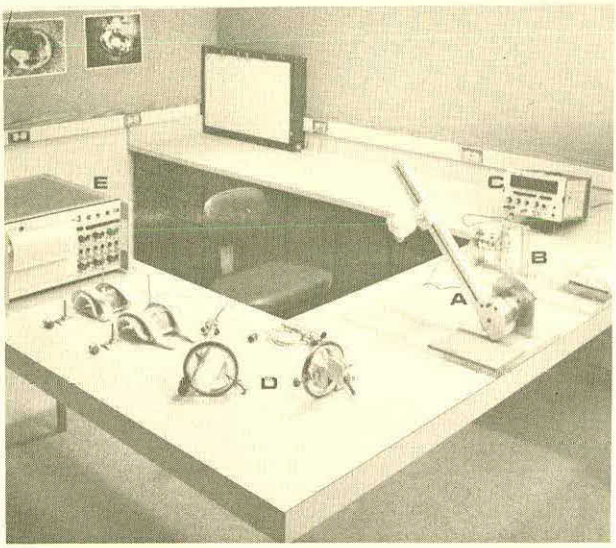

Fig. 5. Elbow evaluation: (A) flexion/extension strength testing device, (B) pronation/supination strength testing apparatus, (C) digital readout for elbow strength, (D) 3-D elbow electrogoniometers, (E) Honeywell Viscorder to monitor elbow motion measured by the goniometer.

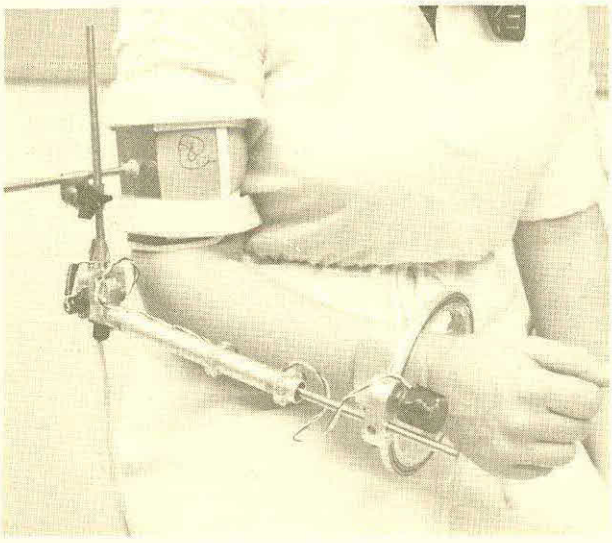

Fig. 6. Triaxial electrogoniometer for the elbow.

elbow motion in a series of 15 daily activities, designed to establish the common range of motion occurring at the elbow. The results shown in Fig. 8 are the mean and standard deviation of the maximal flexion, extension, pronation and supination used for each of the functions. In general, most of the activities could be performed with a flexion-extension range of 90°, an arc between 40 and 130° of flexion. Flexion of less than 40° from the fully extended position was infrequently seen. A functional arc of about 100° of forearm rotation, equally divided between pronation and supination, was adequate for all of the activities.

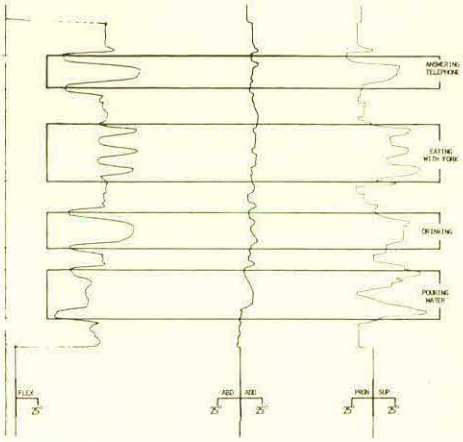

Fig. 7. Three-dimensional elbow motion during activities of daily living.

Elbow Position, Hand to:	Flexion/Extension	Supination/Pronation
Head vertex	118.6 ± 6.1	-46.6 ± 16.0
Occiput	144.0 ± 7.0	-2.0 ± 23.0
Shirt waist	100.4 ± 13.2	-11.9 ± 23.8
Shirt chest	120.0 ± 8.2	-29.4 ± 19.2
Shirt neck	134.7 ± 5.2	-40.9 ± 16.3
Sacrum	69.7 ± 12.4	-55.8 ± 20.1
Shoe	16.0 ± 6.3	19.0 ± 17.2
Range of Motion Required	**Flexion/Extension**	**Supination/Pronation**
Pour from pitcher	35.6 ± 9.5 → 58.3 ± 10.5	-21.9 ± 13.3 → 42.9 ± 19.4
Glass to mouth	44.8 ± 10.4 → 130.0 ± 6.2	-13.4 ± 17.9 → 10.1 ± 11.7
Cut with knife	89.2 ± 9.1 → 106.7 ± 10.1	26.9 ± 26.7 → 41.9 ± 21.2
Fork to mouth	85.1 ± 13.7 → 128.3 ± 8.4	-51.8 ± 19.6 → 10.4 ± 28.1
Use of telephone	42.8 ± 10.0 → 135.6 ± 7.3	-22.6 ± 18.0 → 40.9 ± 15.6
Reading newspaper	77.9 ± 14.1 → 104.3 ± 11.3	7.1 ± 29.3 → 48.8 ± 15.5
Rising from chair	20.3 ± 13.6 → 94.5 ± 12.3	9.5 ± 18.2 → 33.8 ± 19.1
Opening door	24.0 ± 11.1 → 57.4 ± 16.8	-23.4 ± 23.1 → 35.4 ± 18.0
	+ = Flexion - = Extension	+ = Pronation - = Supination

Fig. 8. Normal elbow motion required for activities of daily living.

SHOULDER

An isokinetic dynamometer (Cybex II, Lumex, Inc., Bay Shore, New York) is used to assess shoulder strength and range of motion (Fig. 9). This commercially-available system has been extensively modified by our laboratory to provide the stabilization and adaptability required for our patient studies. The device monitors a subject's strength throughout his available range, thus evaluating a dynamic function. Limb motion is kept at a constant velocity, and any force applied is met by an equal counter-force (7). The motion velocity can be varied from isometric to fast functional velocities. The resistance is in both clockwise and counter clockwise directions, thus reciprocal patterns of motion can be performed and monitored.

A dual-channel recorder simultaneously displays the torque and angular displacement of the joint (Fig. 10). This type of recording provides valuable information for understanding joint function and pathology. The range of motion recordings are reliable at low velocities (7) and with proper stabilization.

Fig. 9. Modified Cybex equipment to measure shoulder isokinetic strength.

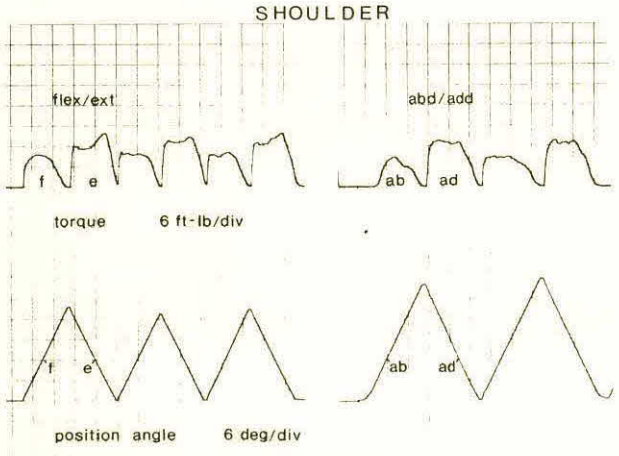

Fig. 10. Torque curves and position angle display from dual channel Cybex II recorder.

This apparatus is presently being employed as part of the functional assessment of patients with proximal humerus replacement after bone tumor resection or for salvage purposes after failure of regular total shoulder replacement arthroplasty. Due to the absence of rotator cuff and the surrounding deep muscles, the range of active abduction and external rotation is significantly reduced. The amount of joint strength in these directions is also limited. However, these patients can have adequate flexion, extension and internal rotation function to carry out necessary daily activities. Normal elbow and trunk motion can be utilized to compensate for the shoulder functional deficiency.

SUMMARY AND CONCLUSIONS

Instrumentation and techniques have been developed which provide quantitative assessment of joint function in terms of strength, range of motion, boney orientation, functional ability and quantitative EMG. Electromechanical devices are employed to objectively measure isometric and isokinetic strength. Conventional hand goniometry for joint range of motion is augmented by three-dimensional analysis which uses a biplanar radiographic technique. Functional ability is assessed with peg board tests and simulated activities of daily living. Electromyographic (EMG) analysis of muscle function is accomplished with wire electrodes. A PDP 11/34 computer records and correlates the integrated rectified EMG signal from eight muscles plus the forces produced during specific functions.

These techniques are used in normal studies and for assessing pre- and postoperative total joint arthroplasty patients involving upper extremity joint abnormalities. Information is available to define normal function, assess deficits, and monitor results of surgery and rehabilitative techniques. Quantitative kinetic and kinematic information is also produced to aide in the design of prosthetic joints.

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FUNCTIONAL GAIT ANALYSIS OF TUMOR PATIENTS FOLLOWING CUSTOM TOTAL JOINT AND SEGMENTAL REPLACEMENT ARTHROPLASTY

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ABSTRACT

Little interest has been shown in the past for en bloc resection of primary osseous lesions due to inadequate methods of skeletal reconstruction. Recent advances in reconstructive orthopedics and custom total joint and segmental prosthesis design have generated renewed interest in local resection of bony tumors. With the primary objective of life preservation, this new method of treatment offers the added advantage of retaining normal limb and joint function. However, the patients' postoperative function must be monitored in order to establish effective rehabilitation guidelines and to protect the implant for prolonged in vivo service. The functional results obtained also provide the basic data for the improvement of prosthetic design and surgical technique. The purpose of this paper is to illustrate the use of quantitative gait analysis as a clinical tool to monitor patients' functional progress from which pertinent information can be derived to enhance the effectiveness of this procedure.

INTRODUCTION

The surgical procedure is carried out through en bloc resection of a localized lesion with an envelope of surrounding normal musculature. The defect is then replaced by a custom-designed segmental and total joint prosthesis. The location and extent of the tumor must be accurately determined to enable removal of all tumorous tissue to prevent local recurrence, but not so extensive as to prevent restoration of useful function. The present report includes only the patients with hip and knee involvement.

The custom total hip replacement (Fig. 1) was initiated at this Institution in 1971.[1] To date, 37 patients have undergone this type of surgery. The mean age of this group is 49, ranging from 13 to 81 years old. Since 1974, 26 patients have undergone segmental replacement at the knee (Fig. 2) as a result of tumor.[2] This group ranged from 14 to 68 years old with a mean age of 34 years.

Any reconstructive surgery involving prosthetic replacement must attempt to satisfy two functional requirements; 1) maintain sufficient range of joint motion for essential activities of daily living, and 2) provide adequate strength with the existing musculature to perform required functions. At follow-up, the 63 custom replaced patients (37 hips and 26 knees) were questioned concerning their pain, functional limitations and the use of walking aids. The majority of these

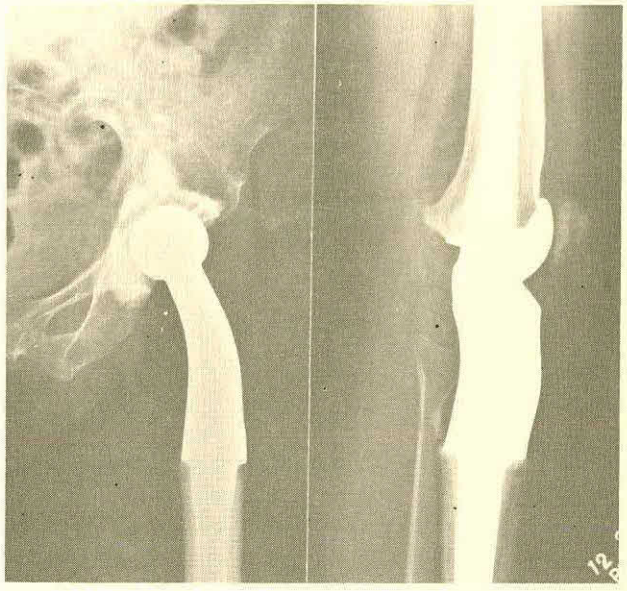

Fig. 1-Custom Total Hip

Fig. 2-Custom Total Knee

patients have no pain and can achieve necessary functions without limitation. The remaining patients with limited function had metastatic disease. They were not included in the present gait evaluation program since the surgery served as a palliative measure.

Through objective gait analysis, a patient's functional status can be continuously monitored. This may allow early detection of mechanical and clinical complications. This data also allows the determination of prosthetic joint forces and moments. Such data can aid in the establishment of an optimal rehabilitation program designed to compensate for muscle weakness caused by radical resection and immobilization while preventing possible overloading of the implant.

METHODS

Gait analysis data includes the three-dimensional joint motion, foot-floor reaction forces, foot switch contact patterns, quantitative joint strength, and temporal and distance factors. When the values of these parameters are compared between patients at different disease stages and those who are under various types of treatment regimes, the effectiveness of the reconstructive

method and the in vivo performance of the prosthetic device can be evaluated objectively.[3]

The gait laboratory (Fig. 3) contains two 14-meter walkways. One is utilized for level walking, while the second is used for varied ground conditions, including a 15° ramp, stairs, and a side slope. A triaxial electrogoniometer (Fig. 4) is used to measure motion occurring at the hip and knee.[4] The patient's foot-floor reaction force is measured by a piezoelectric force plate concealed in the walkway (Fig. 3F). The temporal and distance aspects of gait are measured by four footswitches taped to the subject's shoe at the heel, great toe, first and fifth metatarsal heads. Individual step length and width are recorded by two instrumented plexiglas mats (Fig. 3M) with conductive stainless steel strips placed at 1 cm intervals. Walking velocity is determined by two infrared optical switches (Fig. 3L). These lights also serve as triggers for the computer to begin and end data collection.

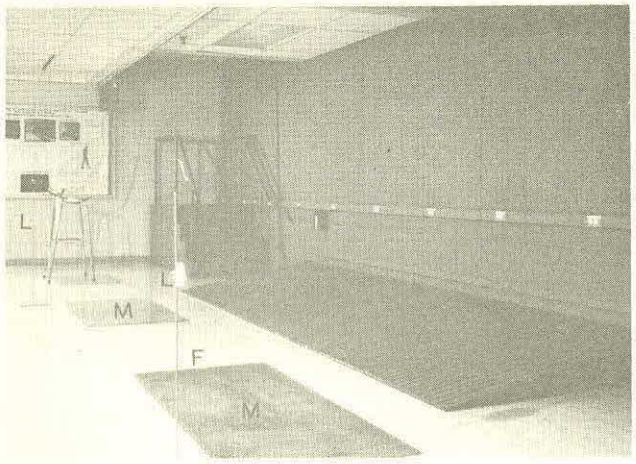

Fig. 3 - Gait Laboratory

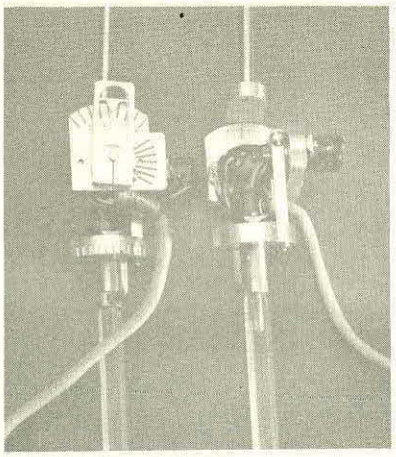

Fig. 4 - Anterior and posterior views of the electrogoniometer.

The entire data collection and analysis is automated by a PDP 11/34 computer (Fig. 5). Average gait patterns and parameters are available 5 min. after evaluation. A physician's summary report is also prepared to be included in the patient's history (Fig. 6).

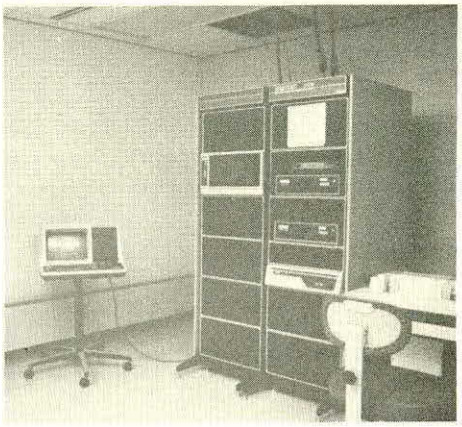

Fig. 5 - PDP 11/34 Computer

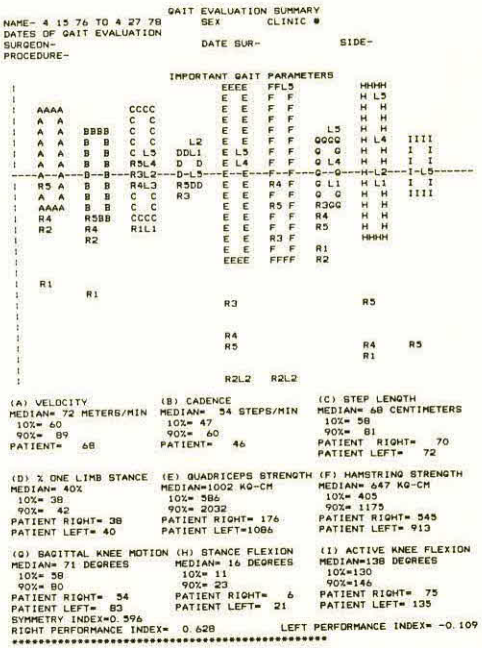

Fig. 6 - Evaluation summary for a custom knee replaced patient.

RESULTS

A typical evaluation summary is shown in Fig. 6. This patient had a custom knee prosthesis and was followed at six-month intervals for 2½ years. The subscripts 1 indicate the subject's initial evaluation, and 2 - 5 represent the subsequent evaluations. The letters R and L refer to the subject's right and left legs respectively. For velocity and cadence only the letter R is used. The bar diagrams represent the 10th and 90th percentile of the parameter among the normal population. The horizontal line stands for the median value for each parameter. The letters making up the border of each bar refer to a specific parameter defined at the bottom of the chart.

This example subject has made continued progress throughout the postoperative period. The

majority of gait parameters were found to be within the normal range, except for the right quadriceps torque, stance phase flexion, and active knee flexion. It is interesting to observe the reduction of the right quadriceps strength with time. This weakening is occurring in spite of the fact that this individual is a very active young man. This indicates the need of continuous quadriceps exercises. The reduced stance phase knee flexion on the right is probably due to the quadriceps weakness.

The gait data is also subjected to parametric, harmonic, and phasic analysis. The harmonic spectrum analysis provides a quantitative evaluation of the motion pattern. The phasic analysis examines the coordination of the joint motion on the same extremity or on contralateral extremities for bilateral evaluations. Figures 7 and 8 are the computer averaged wave form for a custom knee patient and phase plots displaying the interrelationships of the various planes of motion. Figure 7 displays the motion occurring in the replaced knee. The minimal separation seen in the phase plots is due to the hinged prosthesis allowing only planar motion. Figure 8 is a tracing of the subject's unoperated knee motion. The range and pattern of joint motion, as well as the phasic relationship, have the typical features found in normal subjects.

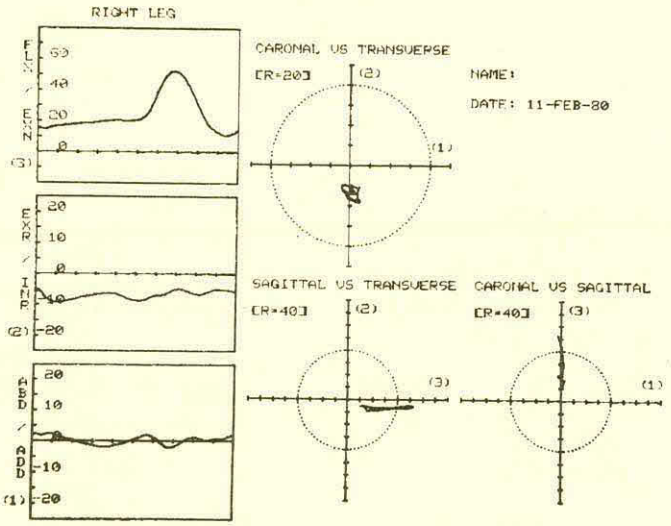

Fig. 7 - Tracing and phase plots from the custom replaced knee.

Two indices are also calculated, one to examine the functional performance during walking of each leg and a second index is used to evaluate the functional symmetry between the right and left legs (Fig. 6). Such indices, as well as other analyses results, will provide reliable objective assessment of patients' joint normality after reconstruction. The compensatory action of the normal side can also be documented.

DISCUSSION

The evaluation is currently being utilized to follow the functional progress of all the custom joint replacement patients done at this institution. To date we have evaluated 43 custom

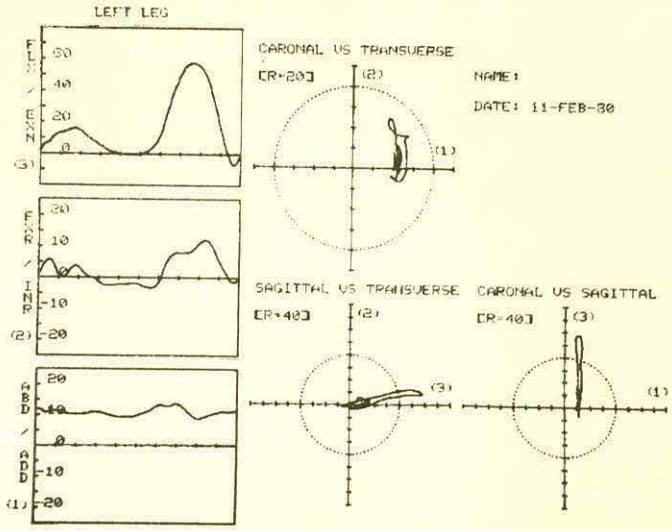

Fig. 8 - Tracing and phase plots from unoperated knee.

patients, a number of which have been evaluated more than once. Table 1 is the average results obtained from the custom knee patients grouped according to the length of time between surgery and gait evaluation. Table 2 represents the average data for the custom hip patients.

The results shown in Tables 1 and 2 reveal that patients' functions are approaching normal and such status is maintained beyond 2 ½ years. Certain patients with postoperative problems were also identified by the gait analysis. Such findings have substantiated the importance of long-term gait evaluation follow-up on these patients. This information can single out the most significant functional deficit of a patient so that a proper rehabilitation program can be recommended.

The gait analysis method is particularly valuable to document the functional changes in individual patients during subsequent evaluations. For example, the tracing in Fig. 9 is from a custom hip patient who shortly following surgery began to have problems with the prosthetic hip. The tracing on the left is the hip motion three months following surgery. The rotations in the sagittal and coronal planes display minimal motion with slightly out-of-phase patterns. Motion in the transverse plane was increased and reversed as compared to normal. These deviations are reflections of laxity and instability about the hip caused by the surgical resection of the hip capsule and musculature. Based on these results, a tight-fit technique was recommended and surgery was performed subsequently to eliminate subluxation. The tracing to the right is the patient's hip motion two months following revision. All parameters are improved, showing increased sagittal motion, increased coronal motion with the normal stance hip adduction, and reduced transverse motion with the normal stance phase internal rotation of the femur relative to the pelvis. This result truly reflects the value of gait analysis as a means to identify functional problems in this group of patients.

CUSTOM KNEES

	Follow-Up: Under 1 yr. (N = 15)	1 + 1.5 yr. (N = 11)	1.5 + 2.5 yr. (N = 7)	Over 2.5 yr. (N = 6)	Normals (N = 50)
Quads (kgcm)	302 ± 155	336 ± 319	371.7 ± 294	340.1 ± 252	1172 ± 527
Hams (kgcm)	319 ± 221	474 ± 228	462 ± 287	471 ± 328	747 ± 304
Stride Length (cm)	116 ± 14.6	117.3 ± 21.74	119 ± 20	112.7 ± 13.8	137.2 ± 17.15
Step Length (cm)	58.3 ± 7.8	59 ± 11.17	58.1 ± 11	56.5 ± 8.36	68.8 ± 8.5
Stride Length/LEL	1.29 ± .17	1.31 ± .19	1.31 ± .17	1.26 ± .13	1.52 ± .14
% Stance	58 ± 3	58.27 ± 2	61.4 ± 1.8	59.3 ± 4.3	59.4 ± 1.8
% Single Limb	36.9 ± 3	37.9 ± 3	37.1 ± 1.2	35.6 ± 3.3	40.66 ± 1.9
Cadence (strides/min)	43.5 ± 4.79	46.7 ± 5.1	45.9 ± 5.5	46.7 ± 3.69	54.45 ± 4.88
Velocity (m/min)	51.97 ± 10.9	58.6 ± 4.9	57.4 ± 8.4	56.11 ± 11.59	74.82 ± 11.6
Total Sagittal (degrees)	41.53 ± 13.8	54.8 ± 9.16	55.7 ± 9.9	53.17 ± 9.11	70.3 ± 8
Stance Sagittal (degrees)	7 ± 3.6	6.64 ± 3.35	8.2 ± 3.4	6.5 ± 4.72	16.8 ± 4.7
Terminal Stance Sag. (degrees)	12.13 ± 3.4	18.6 ± 7.69	27.4 ± 7.8	28.8 ± 4.6	30 ± 5.5
F_1 (% B.W.)	96.3 ± 5.89	98.36 ± 5.6	97.2 ± 5.2	98.17 ± 8.6	111.18 ± 9.5
T_1 (% stance)	31.47 ± 7.98	27.7 ± 6.9	25.7 ± 5.28	27.3 ± 4.5	26.4 ± 3.8
F_2 (% B.W.)	87.7 ± 4.27	86.8 ± 4.56	84.7 ± 5	84.5 ± 5.75	72.2 ± 9.2
T_2 (% stance)	49.67 ± 9.12	50.8 ± 8.13	40.3 ± 6.8	46.8 ± 7.17	48.8 ± 5.7
F_3 (% B.W.)	99.8 ± 8.18	101.8 ± 5.72	102.1 ± 5.4	102.3 ± 6.9	111.6 ± 8.1
T_3 (% stance)	74.53 ± 6.86	75.7 ± 3.93	71.4 ± 10.1	73.5 ± 8.6	78.46 ± 4.1

Table 1 - Average values for custom knee patients at various periods of follow-up.

CUSTOM HIP

	Follow-Up: Under 1 yr. (N = 2)	1 + 1.5 yr. (N = 10)	1.5 + 2.5 yr. (N = 6)	Over 2.5 yr. (N = 4)	Normals (N = 21)
Stride Length (cm)	114.5 ± .5	100.5 ± 15.1	101.3 ± 7.8	109.0 ± 15.8	134.7 ± 13.9
Step Length (cm)	57.5 ± .5	52.9 ± 8.2	51.7 ± 5.1	55 ± 8.2	63.8 ± 16.4
Stride Length/LEL	1.24 ± .12	1.03 ± .07	1.17 ± .09	1.02 ± .13	148.6 ± 12.6
% Stance	59.5 ± .5	60.5 ± 2.3	61.3 ± 5.2	60.3 ± 4	60.9 ± 2.7
% Single Limb	38.5 ± .5	35.9 ± 5.2	38.5 ± 2.1	30 ± 5.5	39.9 ± 3.2
Cadence (strides/min)	44.4 ± .9	47.5 ± 4.9	48.4 ± 6.5	45.9 ± 5.4	52.9 ± 4.9
Velocity (m/min)	54.4 ± 0	50.7 ± 9.4	52.2 ± 6.6	52.5 ± 12.5	72.4 ± 8.5
Total Sagittal (degrees)	30.5 ± 1.5	27 ± 7.1	29.3 ± 5.5	31.3 ± 6.6	41.9 ± 4.7
Stance Sagittal (degrees)	22.5 ± 2.5	18.9 ± 4.5	18.3 ± 7.4	21 ± 5.4	21.9 ± 8.3
Terminal Stance (degrees)	5.5 ± .5	6.2 ± 1.5	9.3 ± 5.3	6.5 ± 2.6	16.3 ± 6.9
Stance Coronal (degrees)	4 ± 0	6 ± 2.6	6 ± 1.4	10.5 ± 2.2	7.6 ± 2.5
Stance Transverse (degrees)	8.5 ± 2.5	8.1 ± 2.8	9 ± 3	9.8 ± 3.9	9.9 ± 3.2
F_1 (% B.W.)	93 ± 2	95.2 ± 6.9	96.7 ± 7.5	98 ± 4.6	112.2 ± 8.6
T_1 (% stance)	26.5 ± 2.5	33.3 ± 5.5	31.5 ± 4.9	30.8 ± 3.9	23.9 ± 4.1
F_2 (% B.W.)	80.5 ± 4.5	82.2 ± 5.5	85.7 ± 8.9	86 ± 7.9	74.1 ± 8.1
T_2 (% stance)	38.5 ± 1.5	51 ± 6.5	53.3 ± 4.3	45.8 ± 3.9	45.2 ± 7.5
F_3 (% B.W.)	103 ± 4	94.4 ± 5.3	98.8 ± 4.1	96 ± 2.7	113.4 ± 6.7
T_3 (% stance)	74.5 ± 6.5	72.9 ± 5.4	76.7 ± 2.6	70 ± 2.2	74.5 ± 5.5

Table 2 - Average values for custom hip patients at various periods of follow-up.

SUMMARY

Treatment of primary bone tumors involving adjacent joint and soft tissue by radical resection and custom segmental and total joint replacement has shown great potential for curing the disease and restoring the patients' normal functional ability. Gait analysis is used as a means to assess the normality of the hip or knee joint after prosthetic replacement. The results produced are used to improve the design of the implant, as well as the associated surgical techniques. Continuously monitoring the patients' functional status can help to identify problems and proceed to establish effective rehabilitative management for fast functional recovery. Certain activities that may have deleterious effects on the integrity of the prosthetic implant can also be determined. Thus far, we have studied 37 hip patients and 26 knee patients. The results show

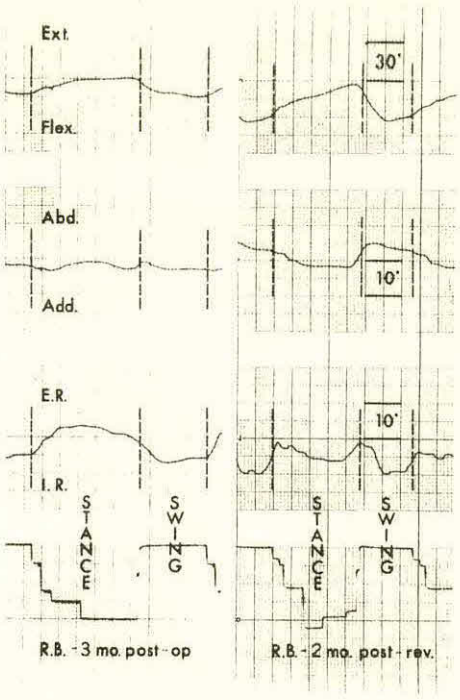

Fig. 9 - Three-dimensional hip motion in a custom replaced hip.

that patients with primary tumors have excellent functional abilities after replacement surgery. However, for those who had metastatic lesions, the postoperative results were less dramatic, primarily due to the extent of the disease, as well as the short life span preventing complete functional recovery. As encouraged by the good results obtained thus far, this technique is now being used to salvage failed total hip and total knee arthroplasties with significant bone loss. Hopefully, a standard procedure can be developed to treat patients with extensive bone and joint loss due to tumors or other orthopedic diseases.

ACKNOWLEDGEMENT

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Communication Rates for Nonspeech Expression as a Function of Manual Tasks and Linguistic Constraints

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The development of nonvocal communication devices has focused upon the matching of existing physical movements and appropriate electronic devices. The ability of an individual to adapt to such systems and to acquire rapid communication has not been fully researched. This paper reviews relevant literature regarding output rates for various nonspeech modes of communication and attempts to develop an understanding of the motoric and linguistic constraints on expressive language.

In a normal conversation the typical adult speaks at a rate that ranges from 126 to 172 words per minute (Perkins, 1971). This rate can be voluntarily modified (faster or slower) by an individual speaker (with a ratio of 2.6 to 1 (Grosjean, 1980)).

It is apparent from this discussion that the average human is equipped with a flexible and reasonably fast means of expressing himself in his native language. When one attempts to replace human speech as an individual's primary mode of expression, much of this flexibility and speed seem to disappear. Current efforts to provide alternative communication modes for severely physically disabled individuals (cerebral palsy, ALS, etc.) demonstrate this problem. Attempts have been made to use alphabetic and other graphical outputs to replace speech. The maximum output rate for these systems is typically below 15 words per minute.

The error free maximum rate for the eye-gaze encoded ETRAN has been calculated at 12 words per minute (Rosen, 1978) while the similar rate for a single key typing device has been shown to be 8 WPM (Crochetiere, et al., 1977). (These measurements have been made on subjects with no known disabilities). The remainder of this paper will explore the use of alternative modes of expression and will address the consequences and expression of these relatively slow output rates.

Typewriting as an Expressive Mode

As a predictable replacement for speech it might be suggested that handwriting or typewriting be used as a substitute. With accommodation for physical disabilities, this is what has been done in the alternative systems mentioned earlier, and described in detail in Vanderheiden (1976), one may initially recall the secretarial typing rate of 50-60 WPM (Seibel, 1972) and compare this to the maximum output rate for the alternative systems. This comparison seemingly diminishes the hope for rapid and flexible expression for the disabled until one realizes that the typical typing rate is for transcribing existing text, rather than generating original messages. Therefore, the extent to which these numbers can be compared legitimately is not altogether clear.

In a series of studies, the Hopkins Laboratory of Engineering Psychology at the Johns Hopkins University has examined human interaction through various modes of expression. The general paradigm in all aspects of the work was to provide two individuals with separate information and a common problem. Neither could individually solve the problem. However, by working together a solution was possible. This design provided the experimenters with an opportunity to study, among other things, the effect that different modes of communication had on such measures as time for solution (an indirect measure of information communicated) and words per minute of communication. Among the modes studied, those of interest to this paper were the "Communication Rich," allowing for both visual and spoken communication; "Handwritten," allowing for real time exchange of written messages only; "Experienced Typing," allowing for communication through a pair of computer terminals by subjects who were experienced typists; and "Inexperienced Typing," allowing for communication through a pair of comptuer terminals by subjects who were not experienced typists.

Examination of the data for Chapanis et al. (1972) shows that the typing rate for experienced (pretested at $>$ 40 WPM) and inexperienced (pretested at $<$ 25 WPM) typists were 18.1 WPM and 10.2 WPM respectively. The corresponding rate for the "Communication Rich" mode was 190 WPM. While the latter does not disagree with other rates for expressive speech, the typing rates are considerably different from the transcription rate of an experienced typist. It is of interest to note that the original rate difference between experienced levels of typing becomes considerably smaller when the subjects are in the expressive mode.

One might have expected the variations in typing experience to have produced a larger rate difference between groups of subjects.

Weeks, Kelly, and Chapanis (1974) compared the ability of experienced vs. inexperienced typists in problem solving situations and found similar results. Again the experienced typists were tested to have a transcription rate in excess of 40 WPM. The expressive rates for the two sets of subjects were 27 and 18 WPM, respectively, compared to their mean transcription rates of 43 and 21 WPM. This shows that the experience of the typist, while important for transcription, has a much smaller effect on expression rates.

Weeks et al. (1974) describe this outcome as being counterintuitive and explain it... "Our pretest of typing skill, intentionally like those used to select typists, was a measure of speed and accuracy in copying prepared text. By contrast, our experimental task... required a great deal of planning and decision making as our subjects composed their messages at the typewriter."

It may be inferred from this that while the transcription process may be unequally practiced among the typists, the expression process using a typewriter is more nearly equally unpracticed.

Handwriting as an Expressive Mode

Returning to the data of Chapanis et al. (1972), the expressive rate for the handwritten interchange was 23.5 WPM. This exceeds the rates of both levels of typists. One would not expect this since handwriting is generally considered to be slower than typing. Seibel (1972) reports that subjects using free cursive handwriting to copy individual words reach a rate of 80 characters per minute or approximately 20 WPM. In a small pilot study this author asked subjects to transcribe in "legible" handwriting a passage of magazine text. The mean transcription rate was calculated to be 33 WPM.

The difference between these two rates can be attributed to task differences. In the latter, subjects were copying meaningful text onto standard sized writing paper. In the former, the subjects were presented individual words which were copied into predefined boxes on forms.

The significance of the handwriting data is that there is only a very small drop (if any) in output rate from transcription to expression. Interpretation of this in light of the performance of the experienced typists is aided by consideration of the use of handwriting as compared to typing.

The primary function of typewriting among the subjects (all either high school or college students) can be assumed to be the preparation of final drafts of academic work. This, like most clerical functions, is transcription.

Conversely, the principal function of handwriting is the expression of ideas onto paper. Not since early penmanship practice in elementary school would the subjects have used handwriting for transcription. The writing of letters, essays, and even class notes are processes which are more related to the "conversational" tasks in the Johns Hopkins studies.

This analysis is supported by Muhich (cited in West (1969)) who reports that on straight copy work, second semester high school typists showed a mean rate of 41.7 WPM. When they were asked to type rough drafts, tables, or letters that rate dropped by more than one half. This final rate is more or less in line with the expression rates found in the Johns Hopkins studies.

Other Manual Communication Modes

Conversion of thought into written or typed words is not the only non-speech mode available as a replacement for natural speech. Several highly developed manual systems are presently available for the deaf population.

In American Sign Language (ASL), movements of the hands, either together or individually, define unique signs which can be received visually and interpreted by the receiver. The signs used in ASL has its own linguistic structure and is generally considered to be its own language, not a modified form of English (Bellugi and Fischer, 1972).

Fingerspelling represents another manual communication system for the deaf. The American Manual Alphabet describes a variety of finger configurations, each representing one of 26 letters and 10 numbers (Zakia and Haber, 1971). Each word to be expressed is therefore represented by a sequence of representations of its constituent letters. Fingerspelling generally follows the linguistic rules of English. Fingerspelling may often be used to compliment ASL. Some ASL words are frequently fingerspelled (off, do, is, was, etc.). In the studies reported in Bellugi and Fischer (1972) between 2% and 12% of the words were fingerspelled.

Both of these systems offer an opportunity to study the capabilities of well-practiced individuals using non-speech modes of communication.

No replication of the Johns Hopkins studies using native users of ASL or fingerspelling exists. However, related data can be used for comparison. Grosjean (1980) reports transcription rates for both speakers and signers reading "aloud." (Reading "aloud" is this author's descriptive term to indicate copying from printed material to another form without the transcriber altering the content). He reports a speaker rate of 224 words per minute compared to 81 signs per minute.

In another study Bellugi and Fischer (1972) asked bilingual subjects (children of two deaf parents) to retell in their own words some story from their childhood, or a story they knew well. The subjects showed mean articulation rates of 4.7 words per second, and 2.3 signs per second. These

numbers cannot be directly translated into meaningful word per minute or sign per minute rates since all pauses between words and signs were omitted. (This was not done for other reported output rates.) Using the percentages of communication time devoted to pauses presented by Bellugi and Fischer a simple computation can be made for corrected word and sign per minute rates. These become 231.6 words and 119.63 signs per minute respectively.

There is obviously no way in which the outcomes of these two studies can be compared without knowing more about the subjects, and the material in question. One observation that can be very cautiously made is that there appears to be no dramatic difference between transcription and expression as there was in typewriting exercises.

(At the risk of comparing two studies, one might give some thought to the appearance of higher rates for both modes of output in expression over transcription. This topic would make for interesting future study).

Rates for fingerspelling are reported in Zakia and Haber (1971) at .2 second per letter, or one word per second for a 5 letter word in fingerspelling context, and in Bellugi and Fischer (1972) 2.7 words per second in ASL context. The high rate for the latter is undoubtedly due to the use of fingerspelling for the short words described earlier. Bellugi and Fischer report the average length of these to be three letters, thus yielding a rate of 8.1 letters per second or .12 second per letter. Zakia also reports rates of .162 second per letter during an experiment where experienced fingerspellers copied written words presented out of context. Again since these data are from two different studies, it would be difficult to attribute significance to the small relative differences caused by expressive and transcriptive modes. It is, however, of note that no large difference does occur between rate data for the two modes.

Manual Task Limitations on Expression

With the exception of typewriting by experienced typists there can be no evidence found that the rate of expression using the various modes is appreciably less than the rate of transcription.

The rate of speech as well as rates of handwriting and signing appear to have an upper limit imposed by the physical dynamics of the neuromotor system. If one is to assume that the expression process adds in some way to the cognitive load, that increased load does not appear to noticeably reduce the output rate of the individual.

The addition of a more cognitive load such as the requirement that subjects speak and sign simultaneously (Bellugi and Fischer, 1972) only slightly reduces output rate. The upper boundary is still primarily determined by motoric considerations. Kelly and Chapanis (1977) report a study where subjects were required to communicate via typewriter using 300, 500, and unlimited word vocabularies. The word-per-minute rates were similar to those found in other Johns Hopkins studies. There was, however, no significant effect due to the vocabulary limitations. In spite of the constraint of using only words within the limited sets, and the added effort to plan messages accordingly, the subjects were able to maintain their output rate. The typing rate seems to also be insensitive to increased cognitive load.

If typing rate is insensitive to this added processing requirement, why then is typing rate so dramatically affected by the transition from transcription to expression? Quite possibly the significance rests only on the degree to which typing is well practiced or automatized for the particular function being tested. Traditional typists are trained almost exclusively in copying text without reading the content. Use of a typewriter in the expressive mode is therefore a function in which an "experienced" typist may have very little practice. Looking at the Chapanis data for both experienced and inexperienced typists one sees that indeed there is a greater absolute drop in rate for the experienced typists. The ratio of expression rate to transcription rate, however, is approximately 50% in both cases. This value corresponds to the reported drop found by Muhich (cited in West, 1969).

This decline in rate would presumably be attributed to the increased conscious effort to translate thoughts into finger movements. As shown in fingerspelling and ASL research, however, this conscious effort can be greatly reduced with practice, with a corresponding increase in output rate. One would likely expect, on this basis, that typists with a great deal of practice in the use of a typewriter for expressive communication (as opposed to copying) might attain very high rates approaching their dynamic physical limits. A study by this author is currently being planned to investigate this concept. Subjects will be selected for their ability to compose original work at a keyboard (i.e. newspaper reporters, authors, etc.).

Linguistic Aspects of Alternative Modes

All discussions so far have related to rates of communication based on words per minute. This measure is fortunately only one factor in the applicability of alternative modes for fluent communication. Included in Chapanis et al. (1972) was an indirect measure of the amount of information rate of transfer. In each of Chapanis' problem solving tasks the time for solution was measured. Since no one team member could solve the problem alone, some minimum amount of information must have been exchanged to allow for correct solution of the tasks. Solution time is an indirect measure of this. Although this communication rich mode allowed for a rate of 190 WPM compared to the 23.5 of the handwriting mode, (a ratio of

approximately 8 to 1) its mean solution time was 29.0 minutes compared to 53.3 minutes for the handwriting. This is less than a 2 to 1 ratio. Clearly something has occurred which has allowed the relatively slow handwriting to nearly keep pace with the much more rapid speech channel. Chapanis et al. (1977) performed a linguistic analysis of the messages exchanged between their subjects. They found the spoken exchanges to be verbose. Comparing averages of speech to handwriting, in 29.0 vs. 53.3 minutes, 1563.8 vs. 118.5 unique words used. The researchers found the handwritten and typewritten modes (the latter showed solution times of 66.2 and 69.0 minutes for inexperienced and experienced typists) to exhibit drastically modified linguistic structure. They highlight a literal copy of an exchange between two inexperienced typists by saying, "If one adheres to all orthographic and typographic rules, there is not a single correct sentence in the entire table (referring to the sample), or, for that matter, in the entire protocol." Examination of the samples shows that the subjects informally adopted rules for "telegraphing," abbreviating, and otherwise modifying their messages, while allowing the content to be understood.

Horowitz and Newman (1964) compared spoken and written expression from the point of view of the analysis of units which were defined as ideas, subordinate ideas, ancillary ideas, etc. They report that speech was considerably faster than writing, but report a different spread of units within the two modes of expression. Written expressions contained a skewing toward ideas with a relatively small number of subordinate ideas. Spoken expression showed more ideas per minute by a factor of approximately 1.5 to 1 (small, considering the ratio of output WPM rate possible) and showed much higher numbers of subordinate ideas. This study supports the concept of linguistic modification to accommodate the slowness of the expressive medium.

The theory behind ASL and its structure is that it is more appropriate for the medium of manual signing than is the English language. In the story-telling experiment of Bellugi and Fischer (1972) the linguistic unit was taken to be the proposition. Although the number of signs per minute was less than the number of words per minute, the number of propositions per minute was the same for each system. The number of propositions is again an indirect measure of information exchange. The same story can be told (with most important features included) in two systems in the same length of time. The only differences are the articulators (lips, vocal tract vs. hands, arms) and the linguistic structure of the messages.

CONCLUSIONS

The speech mechanism appears to be a highly practiced, automatized process. The subconscious use of the speech articulators allows for rapid communication of thoughts without demanding full attention to the function of speaking. It is this automatization, or release from conscious control, that is to be sought in alternative forms of communication. In the preceeding discussions it has been shown that speech may not be the only motoric process capable of reaching this state.

Alternative functions may in fact have limitations due to their physical nature. The speech articulators are undoubtedly faster and more coordinated than the fingers and arms. The speech process may not however be thought to have an inherent advantage in being able to subconsciously access language. The data for transcription and expression using handwriting show only a minor advantage to the transcription mode. If the transcription mode is thought to be the upper limit of the ability of the musculature to produce legible words then the expressive mode does not lag too far behind. In manual signing, Bellugi and Fischer (1972) have shown that bilingual subjects can tell a story simultaneously in both spoken English and in ASL without appreciable reduction in output rate. This task is clear evidence of the ability of the hands and arms to become as automatized for language as the lips, the tongue, and the vocal folds.

The superiority of speech in terms of words per minute is also a function of its compatibility with the spoken language. The evidence presented by Bellugi and Fischer (1972) regarding the ASL linguistic structures which allows the signer to communicate "information" at rates equal to that of a speaker, clearly shows that the motoric slowness can be overcome by making additional accommodations. The linguistic analyses of the written and typed communication by Horowitz and Newman (1964) and by Chapanis et al. (1977) indicate that the modifications to the English linguistic rules made informally and without advanced planning aided subjects in maintaining reasonable information rates despite dramatically low word-per-minute rates. One might begin to suspect that with appropriate design a set of syntactic rules could be generated to allow for written and typed communication to be increasingly productive.

This returns the discussion to the original topic regarding the physically disabled. There needs to be continued work in the area of human engineering to increase the motoric output of the individual. Ignoring for the moment any cognitive handicap, the disabled person should be able to learn to express himself at the rate allowed by his motoric ability in conjunction with any processing technology required. His ability to maximally use his abilities should be no different from that of his able-bodied counterpart who expresses himself nearly as fast in handwriting as he can copy, or the person who can fingerspell or sign conversation as fast as the fingers and arms will allow.

The data from the Johns Hopkins studies indicate that the users of the nonvocal communication systems described earlier, with their 12 and 8 word-per-minute rates, are in the same range as the inexperienced typist (10.2 WPM). The indirect

measure of information transfer (inverse of solution time) for this mode was only slightly less than one half that of 190 words-per-minute speech.

If the nonvocal individual, using existing assistive devices and adopting the linguistic compromises of the inexperienced typists, can reach nearly 50% of normal speech rate information, what can be expected if the man-machine interface is improved, and structure analogous to ASL is adopted?

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LARYNGECTOMEE SPEECH AID WITH PROSODIC CONTROL

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Design principles of a hands-free, variable-pitch, controlled-volume, unobtrusive speech support system for laryngectomees are described. The scheme is to make the best use of residual articulatory faculties of laryngectomees and to exploit digital signal processing techniques and devices. The following novel features characterize the system now under development:

(1) The vocal tract is excited by a miniature, low-power driver affixed inside the oral cavity; ensuing sound vibrations are sensed by an intraoral miniature acoustic sensor.
(2) Control and information signals to and from the mouth are transmitted by electromagnetic coupling.
(3) Actual speech sounds are radiated, after amplification and spectral correction, from a chest-level concealed loudspeaker.
(4) Intonation and dynamics are regulated by the speaker via the rate of exhalation, which is sensed by a miniature device positioned at the stoma opening.

INTRODUCTION

The long-term goal of this ongoing project is to establish a theoretical and empirical basis for the development of a hands-free, variable-pitch, controlled-volume, unobtrusive speech support system for laryngectomees.

Such a system, whose installation would not require major surgery, is envisioned to incorporate an intraoral sound source as well as a sensor, and derive its pitch and intensity parameters from tracheal airflow. Consequently, in this research we are exploring the effects on speech of non-conventional excitation of the vocal tract, as well as the suitability of tracheal airflow for controlling time-varying analog parameters. The objective of the latter is to derive both pitch and intensity control signals from tracheal airflow rate.

Some of the fundamental research associated with these objectives may also be of value in other health-related fields, specifically, in mobility and communication aids for quadriplegics, and in phonation disorders other than laryngectomy.

RATIONALE FOR THE PROPOSED SYSTEM

The starting premise of this project is that by exploiting recent advances in the understanding of speech acoustics and in speech signal processing it should be possible to combine the medical simplicity of intraoral vibrators with the (assumed) acoustic advantage of implanted artificial larynges. We also note that whilst the costs of surgery, hospitalization and post-operative treatment rise constantly, the price of sophisticated electronic prostheses declines. Also, patients cannot always sustain additional surgery immediately after laryngectomy. It then follows that every attempt should be made to find a technological alternative to implantation into the pharyngeal wall. Since trans cervical vibrators have several intrinsic limitations (back radiation, high visibility, etc.) which probably cannot be overcome, one is left with oral vibrators.

A basic problem with an oral vibrator is that no matter where it is installed, its location differs from that of the larynx it replaces. Consequently, the resonant pattern set up by its vibration in the vocal tract would, generally, also differ from the natural pattern. When this difference becomes noticeable, it decreases speech intelligibility and naturalness.

An additional limitation of any oral vibrator is the low level of undistorted power output, due to its necessarily small size. This is particularly noticed outdoors or in a noisy environment.

Both problems could be resolved by placing in the mouth a miniature sound sensor and applying to its output post-processsing (to reduced distortions) and amplification (to provide sufficient acoustic power). Moreover, the intraoral sensor developed for a fully artificial system might also be used by esophageal speakers. In fact, there is no reason why even the vibrator plus sensor system could not beneficially coexist with a limited degree of esophageal speech, with the latter providing the air necessary for fricatives and plosives, and the former augmenting vowel production with variable pitch as an added feature.

In order to vary pitch and volume it seems natural to utilize the pulmonary air flow at the stoma by deriving from it appropriate control signals. (The feasibility of training subjects to control pitch by breath flow is discussed below.) This would restore to the respiratory system the role it plays in speech production, and free the user's hands for other purposes.

Four key questions thus emerge which must be resolved as part of the design task for a superior speech aid for laryngectomees:

(1) What type of distortions in the speech signal are encountered, and how can they be corrected, when the phonating source is transferred from the glottal area to some-

where in the front of the oral cavity, and the resultant sound is also sensed intraorally?

(2) What type of intraoral driver and sensor meet the physical and environmental requirements of the task ?

(3) By what method of energy transfer, modulation and coding can the intraoral devices be effectively linked to the body-worn equipment ?

(4) How can rapid and accurate control signals for pitch and intensity be derived from pulmonary airflow ?

Answers to these questions are the main goal of our current research. In what follows we shall describe the broad outlines of our system.

OVERALL DESCRIPTION OF THE PROPOSED SYSTEM

Our current conception of the prototype system, whose design would be based on findings of ongoing research, can be described as the interconnection of the following subsystems:

(a) Pulmonary airflow sensor (over the stoma)
(b) Excitation signal generator and transmitter
(c) External (neck) antenna loop
(d) Internal (oral) antenna loop
(e) Receiver and electro-acoustic driver (dentally mounted)
(f) Acoustic sensor and signal transmitter (dentally mounted)
(g) Signal processor and power amplifier
(h) Loudspeaker (worn over the chest)
(i) Battery pack

The physical layout of these components is illustrated in Figure 1, and their interconnections in Figure 2.

Operation of the System

Activated by the Pulmonary Airflow Sensor (a), the Excitation Generator (b) produces a substitute larynx tone which is modulated onto a high frequency carrier and then radiated by the External Antenna (c). The radiated energy is picked up by the Internal Antenna (d) and is converted into and acoustic signal by the dentally mounted Electroacoustic Driver(e). The sound so created is selectively amplified by the vocal cavity, and can be augmented by fricative and plosive sounds produced by buccal or esophageal air.

The aggregate speech sounds are sensed by a likewise dentally mounted Acoustic Sensor (f), which then modulates them onto a different carrier from that used before, and re-radiates them via the same Internal Antenna (d). The signal picked up by the External Antenna (c) is electronically processed and amplified by the Signal Processor (g), and is then radiated at an adjustable average power level via Loudspeaker (h). Power for the external components is provided by the Battery Pack (i).

At first sight this may seem to be a rather complex system, but it should be noted that items (c), (d) and (i) are readily available, (b) and (g) can be implemented by integrated circuits, and only finding suitable types of electroacoustic transducers (a), (e), (f) and (h) may be problematic.

It may be feared that prospective users would

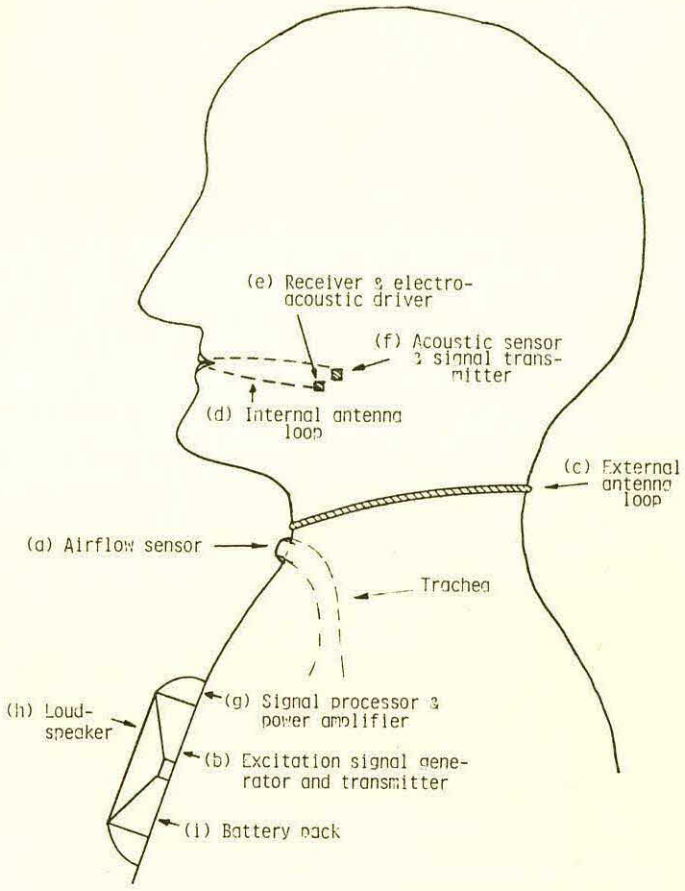

FIGURE 1. Physical layout of the proposed system

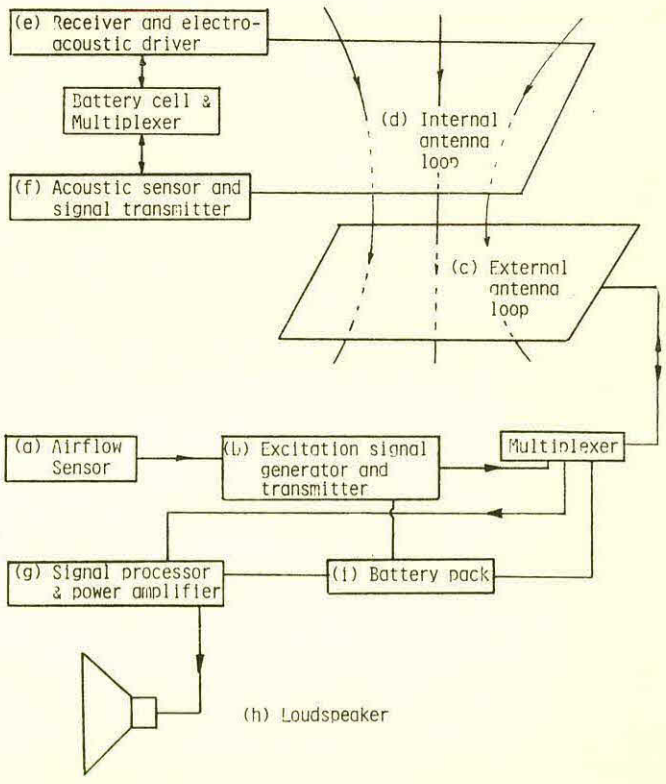

Figure 2. Interconnection of components of the proposed system

become self-conscious about their amplified voice emanating from a flat loudspeaker strapped to their chest. However, disparity between auditory and visual cues for sound are are already present with laryngectomees using shirt pocket amplifiers and in any television set.

In what follows we shall discuss in some detail one of the main innovative features of our system, namely the control of pitch and intensity by tracheal airflow.

AIRFLOW CONTROL OF PITCH

In normal speech the fundamental frequency (Fo) is determined by a delicate interplay of laryngeal muscle activity and subglottal air pressure. The underlying mechanism is nowadays fairly well understood [1,2,3]. Little of this knowledge is directly applicable, however, to a laryngectomee. Two remarks are worth quoting, though: "All stressed syllables are accompanied by an increase in subglottal pressure produced by the action of the respiratory muscles; many pitch changes are also accompanied by an increase in subglottal pressure" [3]. Also: "Whenever (the cricothyroid) muscle is passive, Fo will be controlled by subglottal air pressure" [1].

These facts encouraged us to experiment with the use of airflow as a control signal. In our first series of tests the subject received visual feedback from an oscilloscope screen which displayed both a "target" curve and the time variations of his own breath flow. A detailed description has appeared elsewhere [4] and will not be repeated here.

In a second series of tests the feedback was auditory, and the subject aimed at mimicking the intonation pattern of a short utterance recorded on an endless tape loop. The experimental arrangement is shown in Figure 3. The subject, a 23 year old female graduate student, was a native English speaker. She first recorded several test sentences which were then passed through a Frøker-Jensen Trans-Pitchmeter. The resulting pitch contours were recorded on endless tape loops, and she listened to these in one ear while in the other ear she heard a triangular wave whose pitch was proportional to her rate of exhalation. A test tape of the pitch contour imitations has been prepared and will be played at the Conference.

In the next phase of our experiments we have employed a similar setup but for controlling the Fo input to a Computalker Model CT-1 Speech Synthesizer. Pre-recorded test sentences were synthesized on the Computalker, using its own programmable intonation pattern. Next, the same sentences were synthesized but with the Computalker's Fo control signal replaced by one derived from the subject's airflow. Fairly natural sounding intonation patterns were generated. (A tape will be played at the Conference.) The pneumotachograph used for laboratory experiments is unsuitable for monitoring airflow at the stoma. Instead, we propose incorporating in a standard stoma button a miniature airflow sensing device, such as a self-heated thermistor, pressure-sensitive transistor, hot-wire anemometer etc. It is important for the device to respond to changes on the order of 50-100 ms.

Our experience so far indicates that it will be desirable to restrict pitch control to the exhalation phase, with the inhaled air passing through the stoma unimpeded. (This is especially important in view of some reported opposition of laryngectomees to a new, disposable humidifier at the stoma air intake). We consider using a special two-way valve for separating air intake and outflow [5].

Occasional interference with the pitch control is expected from the user coughing or clearing his stoma. Coughing implies a rather rapid exhalation, and a threshold could be set in the signal amplifier to recognize and ignore it. Clearing the stoma could present a problem for the airflow sensor; in the final implementation it may be necessary to make the sensor readily removable for this purpose.

Figure 3. Experimental Arrangement for Airflow Control of Pitch

While the above discussion was limited to the control of pitch via airflow, our scheme actually calls for the control of intensity as well. These two parameters are generally thought to be related, but few data exist to support this. In Figure 4. we illustrate this relationship: a test sentence spoken by an American male was sampled at 12,000 samples/second and analyzed by computer. When voicing is present the logarithm of intensity is seen to follow pitch variation fairly closely. We thus hope

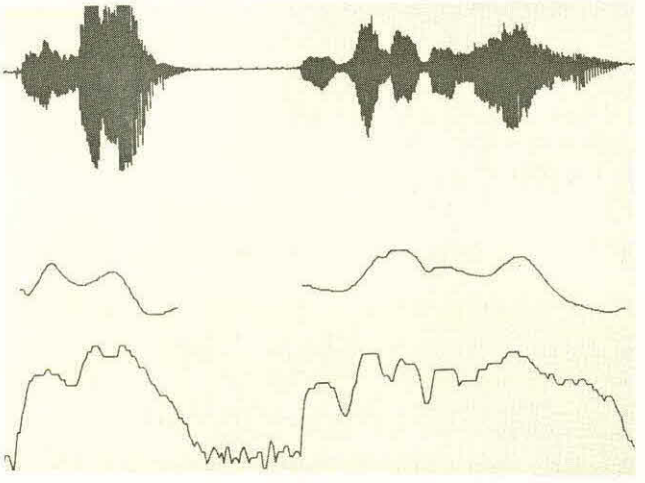

Figure 4. Prosodic features of the sentence: "Oh my! He really will marry me!"
Top trace: speech wave
Middle trace: Fo variations (linear scale)
Bottom trace: Intensity variations (log scale)

to derive from a single airflow rate two different, albeit functionally related, control signals for pitch and intensity.

VOCAL TRACT MODELLING

Reference was made earlier to distortions arising from the unnatural placement (in the forward half of the vocal tract) of the substitute larynx. We have studied the effects of unconventional excitation of the vocal tract both by the use of a mechanical analog [6] and by computer simulation [7]. Preliminary results suggest that correction for spectral distortions could be accomplished by digital - and if necessary adaptive - filtering.

PERFORMANCE CRITERIA

We feel that satisfactory progress towards an improved speech support system for laryngectomees will have been made if we succeeded in demonstrating the following:

(a) Representative English sound can be produced by non-conventional excitation of the vocal tract, followed by post-processing.
(b) Fo and intensity can be controlled by airflow sufficiently rapidly and accurately to create natural sounding speech intonation and dynamics.
(c) Means can be found to turn the excitation source on and off by actions of only the articulatory and respiratory system.
(d) Motivated laryngectomees find the system acceptable, after an initial traning period, for regular use.

CONCLUSIONS

We see the main significance of our system under development in the breadth of its attach on the problem of speech aids. Instead of merely improving on the concept of an "artificial larynx", we intend to lay the foundations of a systematic approach to a broad class of phonatory aids. Such an approach, we hope, will also have an impact on the design of prostheses for sufferers of the afflictions such as:

Vocal cord paralysis
Spastic dysphonia
Glossectomy
Pathological pitch deviations
Inadequate esophageal speech
Other congenital or traumatic laryngeal deficiencies

Our modelling studies of acoustic coupling characteristics could lead to standards by which future proposed schemes for intraoral speech aids might be evaluated. Specifically, our models and data should make it possible to predict the acoustic performance attainable with any suggested transducer and sensor location. Surgical hazards (or other difficulties) associated with specific implantation procedures or prostheses could then be weighed in the light of such expected performance.

The information to be gathered on breath control should find application in the design of mobility and communication aids for quadriplegics.

The airflow sensor on the stoma, when developed, may find applications for monitoring the breathing of patients in intensive care units after tracheostomy, following the removal of the breathing machine. In a modified form it may prove useful in the measurement of airflow during normal speech or singing.

The data to be collected on the relationship between Fo and intensity could lead to some economy in channel capacity requirements in those analysis-synthesis systems in which these are transmitted independently.

ACKNOWLEDGEMENT

Most of the work reported here has been performed under grant #PDT-113 of the American Cancer Society Inc. Valuable contributions to the results reported were made by Karen Doyle, Jon Forrest and Steven Hiller.

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RATE OF LANGUAGE PRODUCTION WITH A SPEEC NONVOCAL COMMUNICATION SYSTEM

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This paper will consist of two parts

1) A brief description of a nonvocal communication system called SPEEC, which is based on analyses of the frequency of phonemes and phoneme clusters in spoken English, and a discussion of the rationale of this approach and its implications for communication rate.

2) The presentation and discussion of data on rate of output taken from videotapes of adolescents who have been working with this system since October, 1979.

INTRODUCTION

Communication rate is an extremely important factor for nonvocal people who must rely on some means other than speech every time they interact with others. Use of the alphabet to spell words letter by letter affords the user precise and unrestricted message production, but slows communication to a point where receivers are tempted to use various avoidance strategies to ease the burden on their patience.

Description of SPEEC

SPEEC (Sequences of Phonemes for Efficient English Communication)[3] is based on a technique designed to answer the following questions: given that the number of items on a nonvocal communication lapboard be restricted to N, what are the N items which will allow totally unrestricted communication with the least number of item selections (e.g. pointing acts) per word? The SPEEC formats which are now being tested by nonvocal people are lapboards containing either 256 or 400 items, in alphabetic or modified alphabetic order with more frequently used items grouped to reduce motor demands.

Letters are used in a simplified, consistent orthography to represent the phoneme groups (i.e. word parts) and all phonemes. Numbers from 0 through 10 and a space (normally used only for disambiguation) are also included.

Rationale for SPEEC

Native English speakers probably have active vocabularies well in the tens of thousands.[5] There is much evidence that nonvocal people may have fully as extensive and elaborate linguistic competence.[1] Communication boards using word lists alone are woefully inadequate for the needs of such people, even when words are ingeniously combined to provide clues to or definitions of missing words.

Lists of highly frequent words along with the alphabet offer the most useful format of this type of board. However the nonvocal person who uses a "telegraphic" style may not benefit from a list of high frequency words, which are all function words and are often omitted in telegraphic style (i.e. the, of, is, where). Users must have recourse to the alphabet for content words, with an average of about 5 letters (5 motor acts in direct selection) per word.

The SPEEC technique selects items for their contribution to minimalizing the number of motor acts per word. The corpus of words used to derive the present system is a list of 9,699 different words and their frequencies in 250,000 words of spoken text.[4] This corpus is considered to be representative of spoken English, based on earlier studies which showed the list of frequent syllables in a smaller sample of Parisian French to be almost all (but 17 out of 188) contained in the list of highly frequent syllables in a larger sample of Montreal French.[2]

With the 400-item board, the English corpus referred to can be produced with about 1.5 selections per word. Counts of items needed to represent other samples of written and spoken English range from 1.2 (for a sample of a young child's speech) to 1.6 items per word.

The reduction in number of motor acts per word offers the potential for improving output rate in three ways

1) First, and most importantly, the user needs to execute fewer motor acts per message (than by spelling), which should consume less time.

2) There are fewer points in the message

where the interpreter has to intervene. This reduces the time when the process is stopped, and in particular provides fewer points at which the interpreter can make errors and retard the exchange of information by going off track.

3) The interpreter's guesses or anticipations (an inevitable part of nonvocal communication) are based on more information sooner in the process, and are thus more likely to be correct sooner. However the large number of items involved (version 26 letters of the alphabet) may mean a slower, less accurate item-indication on the part of some disabled users.

Data from SPEEC Users

Videotapes have already been made of 5 of the 6 adolescents now learning SPEEC and more are scheduled. Rate of progress varies greatly among the six participants, ranging from Lesson 3 of the SPEEC Manual, about 45 items after 10 weeks, to most of the 400 items. Items not yet presented are covered by opaque labels except in the case of the most advanced student, who has no labels on her board.

Data is offered here from the first videotaping session for the two most advanced students with the largest numbers of items. The videotapes were made of SPEEC training sessions conducted by the language teacher who usually works with each student, and included previously learned as well as new items. Timing was done by pressing a stop watch when the teacher uttered the last word of the question or the dictated word, and again when the teacher's verbal response indicated that the student had pointed to the correct answer.

Rate of Output Using SPEEC 400 Boards

Preliminary data from 2 users after 10 weeks (2-half hour sessions per week).

LG
Age 14
Manual Pointing
all 400 items uncovered

single item selections
(in response to dictation
or questions)
N=23 3.6 seconds/item

multiple-item selections
(in response to questions)
N=5 4.2 seconds/item

JL
Age 17
Manual Pointing
about 160 items uncovered

single item selections
(in response to questions)
N=2 9.0 seconds/item

multiple-item selections
(in response to questions)
N=28 3.5 seconds/item

The amount of data collected so far is extremely small. Further taping sessions are scheduled with all six participants, and comparisons with output produced by alphabet spelling will be included.

This data concerns output rate rather than communication rate. The users' responses are almost totally predictable by the receivers, and the receivers are experienced with the system and with these users. Future tapes will include less predictable conversation. However these data indicate that the frequent phoneme sequence technique may well have a contribution to make towards significantly improving the rate of communication for some nonvocal people.

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THE QUICKTIC COMMUNICATOR

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The QuickTic is a new row-column scanning communicator for severely disabled people. It is operated by a single switch and allows the user to select words and phrases as well as letters of the alphabet and numbers. The user can program fifty of his own frequently-used words or phrases for quick selection. The QuickTic incorporates an anticipatory spelling feature that predicts the two most likely letters to follow the previous three, and it has been designed in a modular way to keep costs as low as possible. It is designed to be a battery-operated personal communicator that mounts on a wheelchair for portability.

WHY ANOTHER SCANNING COMMUNICATOR?

All the available scanning communicators lack two important factors: first, sufficient speed for face-to-face conversation and second, a physical arrangement that allows face-to-face communication. The QuickTic was designed to address these two problems.

The Biomedical Engineering Center of Tufts-New England Medical Center has had about seven years experience with scanning communicators for severely disabled people. The Tufts Interactive Communicator[1] (T.I.C.) represented the beginning of scanning communicators with its front-panel layout of letters by frequency of use and its emphasis on portability for personal use. Its video screen allowed full-page composition and the strip printer produced hardcopy output. It provided the ability to compose any message desired and gave severely-disabled people the freedom to create without another person in attendance.

The QuickTic represents an evolution of the original TIC. The QuickTic is more portable and speeds communication through innovative techniques but still is controlled by the closing of a single switch. It is designed for face-to-face communication and presents no barriers to visual contact.

DESIGN DECISIONS

Operational Speed Considerations

Speeding up a scanning communicator is not an easy task. Actual scan speed can not usually be changed, as it is set by the motoric and comprehension ability of the user. Therefore, improvements can only come through giving the user the ability to select a desired letter or word without first going through all possible combinations or by reducing the number of switch closures required. The front-panel layout of the TIC is designed by frequency of letter usage in the English language with the most frequently-used letters closest to the "home" square where the scanning begins after each selection. The QuickTic retains this spelling feature and adds anticipatory spelling to further increase the number of desired letters presented without scanning the entire array. The anticipatory spelling is based on quadgrams of the English language which predicts the two most likely letters to follow any three (including spaces). The first and second anticipatory letters presented by the QuickTic are correct approximately eighty percent of the time. Using this technique can increase word output speed by thirty-three percent. Anticipatory spelling can also produce a thirty percent decrease in the number of switch closures required.[2]

A second method to increase communication speed is to add words and phrases to the choices available. If the user can choose larger pieces of information for communication, then speed should increase.

The front-panel of the QuickTic is arranged in a fifty-six box matrix (seven high by eight wide). Each box contains four possible choices which may be letters, words, numbers, or blank spaces available for user programming. There are 108 preprogrammed words and nine phrases which were selected through questionnaires and interviews with non-vocal people.[3]

The method of choosing within an individual box is experimental. At present a set of four tones indicate the different levels within the selected box. This method may be changed to four different colored lights if the tones prove unworkable. Since the cost of four different lights in every box would be prohibitive, the tones were chosen to avoid forcing the user to look elsewhere during the selection process.

The user must make three separate switch closures to select an individual item in a specific box. The addition of the third switch closure to select the level in the box does add more time and effort to the selection process. The tradeoff of time versus larger selection has not been researched for effectiveness but it will be checked as experience with the QuickTic grows.

As one would expect, a dictionary of 117 "communication pieces" can not represent a vocabulary tailored to a specific individual. Everyone has some unique set of words that they use more often than others. The QuickTic has fifty-two spaces for the user to program their own custom "communication pieces". These spaces allow any combination of words or letters up to twenty characters to be programmed by the user into a specific space. Stick-on labels could serve to record the programmed word(s). The user can reprogram these words at will, although a label change would probably require help.

Cost Considerations

Communication devices for handicapped people are expensive. This fact is due in part to the limited market for devices, but is also sometimes due to a "Cadillac" philosophy. The QuickTic has been designed as a low-cost modular unit by allowing additions of such units as a speech synthesizer or a television interface but not incorporating them into the basic unit. The basic unit contains a twenty column printer, a 24-character display, and a standard interface connector for any desired options. Extensive use of an RCA 1802 microprocessor keeps parts count low and costs down.

Other Considerations

The QuickTic has been designed as a personal communicator for a non-ambulatory person. It is designed as a lap tray for a wheelchair and is constructed of heavy-duty plastic able to ignore spills and stains. It is battery-powered for portability and incorporates low-voltage warnings to prevent the user from losing messages.

The ten prototype QuickTic's will collect statistics on the frequency of use of the words on the front panel. These statistics will be collected internally in the QuickTic and printed out on the printer for further information on our choice of front-panel words. The word choice will then be revised for the final QuickTic front-panel vocabulary.

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THE DESIGN OF A LINE OF GAZE INTERFACE FOR COMMUNICATION AND ENVIRONMENT MANIPULATION

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This device is intended to be used by severely physically involved individuals. The device monitors eye movements and calculates where in the visual field the user is looking. Head movement is not restricted, the user is allowed to move freely within a one cubic foot range. Visual line of gaze is continually calculated independent of head movements. As a communication device a user can make a selection by fixating on the desired entry for a pre-selected dwell period. This adjustable dwell period is typically 400 to 500 milliseconds. The following discussion outlines the design process followed to develop a line of gaze interface for a communicator.

INTRODUCTION

There is a large population of severely physically disabled people for whom verbal or written communication is impossible. This has led to the development of many different communication devices, with each device being suited to a particular class of motor dysfunction. The current minimum requirement for any device is at least one controlled repeatable muscular movement. In many cases this is possible, however there is a substantial number of people for whom motor-controlled communication is laboriously slow or even impossible.

It is known that ocular muscular control is frequently one of the last motor functions to be systematically affected by degenerative disease. Ocular control impairments are not directly associated with traumatic injury such as cerebral palsy, spinal cord injury, etc. When poor eye tracking skills are observed in the above, it is caused by additional neurological or developmental problems. Development of a low cost device capable of determining eye gaze direction would offer a fine alternative in communication. Theoretically it is possible to increase the communication rate using eye gaze over that of existing row scan techniques. This is because direct selection of a letter is inherently a much faster technique than row-column scanning.

Determination of visual line of gaze is a skill that everyone is capable of evaluating when we look into another's eyes and project where that individual is gazing. To build hardware that is capable of making this same determination is a very difficult and expensive task. The hardware must find the position of the eye in three dimensional space and then determine its angular orientation. For a computer this information is most easily manipulated if it is described in terms of a vector representation. The line of gaze vector is the line that originates at the individual's eye center and ends at the object being looked at in the visual field. The design of hardware capable of accomplishing this task involves understanding the physiology of the user population. It is then equally important to determine the most efficient mechanism through which the user can control the machine.

DESIGN PROCESS

The design of a line of gaze communication system begins with an understanding of nonvocal communication and the user population. It is important to understand very early in the discussion, the disabilities which leave an individual nonvocal. The diagnosis and implications associated with the disability provide information that will impact the success of the final design. For example, it is important to know why cerebral palsy does not usually affect visual perception and eye tracking skills. This knowledge aids in correctly diagnosing the individual's capabilities and limitations. An understanding of the pathology and physiology of the population that requires the device, provides a solid foundation for the hardware design.

A communication device based on line of gaze must have an operating scenario which is carefully designed from a basis of knowledge and not merely selected for ease of implementation. There are two categories of concern that need to be addressed in the design of an instrument which first monitors a function and then analyzes the dynamic behavior to make appropriate decisions.

The first category involves physiological data that defines specific features of the eye and the head as they relate to known monitoring techniques. The eye, for example, has features important for monitoring purposes which may be quantified. Some of the important features are: maximum eye rotations, optimum comfortable eye rotation, cornea size, speed of eye movement, typical dwell time, frequency and duration. These are quantities which must be known, for design of effective monitoring equipment.

The second category involves the mechanism through which the eye (Line of Gaze) communicates the selection of a desired character. This is a much more elusive understanding to gain than is the physiological data. The patterns of eye movement contain enough information for a computer to decipher a selection from a scan. This information needs to be obtained, quantified, and analyzed for the design of a communication device that will not cause excessive fatigue, physical or mental. The requirement is to obtain the necessary data to ensure an efficient usable design. This includes not only the dynamic behavior of eye movement but also such items as on/off switches, daily setup procedures and a calibration routine.

The daily setup operation is likely to be performed by a person with no technical background. The procedure must be straightforward and within the capabilities of this person. The most significant problem related to setup involves adjusting the eye position monitor to center the corneal reflection image once the eyeglass frames have been mounted on the user. . This operation takes time to learn before it can be efficiently performed. A procedure needed to be developed that would reduce the skill and time of performing this alignment.

Another procedure which is required each time the unit is mounted is the system calibration routine. This is inherently a difficult operation because it involves the user dynamically interacting with the machine and possibly some operator input. The machine must be calibrated so that line of gaze is correctly calculated and the intersection with the object board is oriented properly. To simplify this operation and requiring a minimum of user understanding, the procedure must insure success through its design. This involves a computer program that must look at the incoming data and decide if it is reasonable. Based on this decision, calculations are made for the location and orientation of the line of gaze.

The foundation of this point is comprised of: facts which are quantified, dynamic behavior for which the variables are known, and flow charts of expected operating procedures. This is the information that is needed to begin design and construction of the hardware.

The developed L.O.G. eye monitoring system has the following general configuration and operating characteristics. Calculation of line of gaze is determined by measuring both eye and head position. The corneal reflection of a subjects eye is monitored with an imaging array mounted on a pair of eyeglass frames. The imaging array is similar in function to a T.V. camera. The position of the corneal reflection image is proportional to the eye position. Head position must then be monitored for true eye gaze direction to be determined. The technique employed, involves an ultrasonic ranging device that determines the position of three points on the eyeglass frames. Having eye position and head position allows calculation of line of gaze.

A board may then be placed in front of a subject, containing letters or symbols from which communication of a message would be possible. There are a number of posible techniques that can be implemented such that visual fixation on a desired letter or symbol results in that selection being "recognized" by the device. An example of one technique is: a subject visually fixates on a desired selection for 400 to 500 milliseconds, moves to the second selection and dwells for a similar time period etc. The computer can recognize these dwell times and display only the desired selections and not the transition points between fixations.

Implementation of this line of gaze system as a communicator involves software that will manipulate the results provided by the hardware monitoring system. This approach that has been implemented requires the user to dwell on a desired selection for a set period of time. This dwell indicates to the machine that the fixated point is a desired selection that should be entered. For the machine to accomplish this task, a sampling rate of ten times per second is sufficient. The software is set up to compare at least 4 out of 5 entries of a fixation point on the object board. If our of the entries are within a line of gaze angle of ± 1.2 degrees, the program recognizes the point as a selection. It is seen that the user is required to dwell on a selection for approximately 400 milliseconds to indicate an entry. Once the selection is made, the letter corresponding to the fixated point is displayed on the video terminal. With some practice this technique can be used with a minimum of errors.

CONCLUSION

Although the primary need for the multiply handicapped individual is a communication device, the next step in development is a general line of gaze system that could be used in the areas of environmental control, workstation control in an employment setting and in prosthetic control.

VOICE CONTROL: CLINICAL EVALUATION BY PERSONS WITH SEVERE PHYSICAL DISABILITIES WITH AND WITHOUT SPEECH IMPAIRMENT

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Three studies were undertaken to obtain data on various aspects of voice input for control of standard rehabilitation aids. A user-adaptive, isolated speech recognition system was utilized to obtain data on recognition, operation and user acceptance by a spectrum of speakers including persons with spinal cord injury muscular dystrophy, cerebral palsy and Friedreich's ataxia. Results indicate that good recognition can be obtained by motivated disabled users when the mode of operation of the controlled device does not induce stress in the user. Speakers with mild to moderate dysarthria are potential candidates for voice control. Speakers with severe dysarthria may be candidates but a small inventory of available commands may limit application. Additional effort is required to develop better means of maintaining the position of the microphone with respect to the disabled speaker.

INTRODUCTION

Since August, 1977, the IRM/NYU REC has had prototype speech recognition instrumentation available for clinical trials. Two studies (A and B) were undertaken to explore applications of the system assuming that the rate of recognition for disabled users would be comparable to the system's excellent performance in industrial installations.

Initial review of the findings of A and B, revealed that the recognition rate--considered to be good because of satisfactory performance with one peripheral and excellent with others--was significantly lower than the industrial experience. Studies by others have shown that recognition must be greater than 98% to be readily accepted by users for work applications (1). It was hoped that analysis of our data might provide a possible explanation for the higher error rate. A third study, C, was initiated to obtain base-line quantitative data on our Voice Recognition System (VRS) to verify that recent modification of the unit had not adversely affected recognition.

The scope of this paper is to analyze and correlate the results of the three studies.

SYSTEM DESCRIPTION

The IRM/NYU prototype voice recognition system (VRS), previously described (2,3), utilizes a modified Threshold Technology T-500 System to provide isolated word recognition. A total of 32 different spoken commands can be recognized, 5 of which control the status (on, off, etc.); 4 commands control a modified Craig dictaphone; 15 are for a standard Prentke-Romich ECU-1/ADT environmental system (ECS); and 8 control a modified E & J 3P powered wheelchair. The operator must familiarize the VRS for his/her pronunciation of each command by means of a "training" process which requires that each command be repeated ten times.

STUDIES A, B, AND C

Objectives

Study A. To evaluate and compare voice input with conventional transducers ("puff" and "sip", etc.) for operation of rehabilitation aids by persons with various severe physical disabilities.

Study B. To evaluate the suitability of voice control by ten subjects representative of that segment of the cerebral palsied population with speech impairment.

Study C. To obtain data on VRS recognition by disabled and non-disabled untrained operators.

Methods

Subjects for A and C were patient recruits. Additional subjects for C were volunteers from the professional and service staffs. Subjects for Study B, referred by community agencies, were screened for their ability to produce a minimum of seven spoken commands. The first ten successful candidates were evaluated by speech pathology and psychology services to assess quantitative and qualitative variables of language/intellectual levels.

Clinical VRS trials were conducted by occupational therapists from the REC or Technical Aids Unit. After a voice level was obtained, each subject trained for all control

functions to be evaluated. Subjects in A evaluated the chair only if he/she used modified (pneumatic, etc.) mobility controls. All subjects in B evaluated the chair; successful chair operators with additional vocabulary also evaluated the ECS. All subjects in C tested all commands, the powered chair was non-operational except for speed control; the ECS operational. Subjects in A were encouraged to initially use standard word commands (Table 1). Subjects in B initially used standard word commands or their personal versions. Subjects in C were required to initially use standard word commands.

Alternates were sought for all unreliable (inconsistent recognition) word commands and subjects proceeded to device operation only after a "reliable" vocabulary had been trained. At the end of training, commands (words or sounds) were recorded. For chair driving, non-operational trials preceeded maneuvering. In A and B, data were recorded on vocabulary, ease of operation, user acceptance and problems; in C, on vocabulary and recognition of each input.

Table 1. VOCABULARY

Code No.	Standard	*Individualized Mod. Dysarthria
0	GO	o
1	JOG	og/ah
2	ERASE	- -
3	STOP	op
4	ECU	u
5	TAPE	-
6	WHEEL(CHAIR)	eel
7	FORWARD	on
8	REVERSE	ack
9	LEFT	eft
10	RIGHT	fi
11	FASTER	-
12	SLOWER	-
13	BRAKE	-
14	CANCEL	no
15	PLAY	-
16	RECORD	-
17	REWIND	-
18	TELEPHONE	one
19	DIAL	aisle
20	LIGHT	ight
21	RADIO	a
22		
23	TV	tee
24	DOOR	pay
25	ASSISTANCE	-
26	DRAPES	-
27	FAN	-
28		
29		
30	TURNPAGE	elp
31		

*Study B Speaker

Study A Results

Training was easily accomplished by all but three subjects. The "stop", "left" and "right" functions were frequently not recognized during wheelchair driving. The only command rejected during ECS operation was "stop" when utilized for interruption of the counting process for selection/dial utilizing the standard Prentke-Romich ADT-5 telephone. Speakers with excellent recognition on the ECS had significantly lower recognition during wheelchair maneuvering (Table 3). The only subjects with excellent recognition of wheelchair commands did not actually drive the chair but rather "talked" through typical driving maneuvers during non-operational testing.

Excellent recognition was obtained during ECS operation when speakers utilized the microphone mounted on the flexible arm (Figure 1). The microphone frequently moved out of speaking range during chair movement and also picked up the background sound of chair frame vibration. The headset mike proved better during driving but did require minimal repositioning to maintain optimum recognition. Most operators were dependent for readjustments.

Voice input was preferred for ease and speed over conventional transducers for all ECS functions except telephone dialing (utilizing the Prentke-Romich ADT-5) which was proved to be too time consuming, complicated and prone to error. Voice input was rejected for driving of the wheelchair due to reduced maneuverability as compared to pneumatic controls and the requirement for constant input. One subject commented that "it would be great if I could tell it to go to a particular room but didn't have to "talk" it there every step of the way".

Study B Results

Degree of speech impairment ranged from mild to very severe dysarthria; intelligibility ranged from unintelligible to excellent. Intellectual function ranged from very superior to significantly below age level. Identification of an optimal vocabulary was generally a result of trial and error. One subject required alternate words for all commands; most required assorted alternates in a random pattern. "Stop" "forward", "reverse" and "cancel" most frequently required alternates.

All subjects could operate the wheelchair under supervision. The subject requiring alternates for all commands (Table 1) had the best over-all performance. The need for precise timing of inputs provided the greatest complication of driving. ECS operation was uncomplicated with exception of telephone dialing. Due to the requirement for precise timing, dialing was too time-consuming, frustrating, anxiety producing and prone to error.

During motion most subjects experienced difficulty maintaining satisfactory position with reference to the microphone of Figure 1. The headset mikes showed potential but were uncomfortable when tightened to maintain proper position.

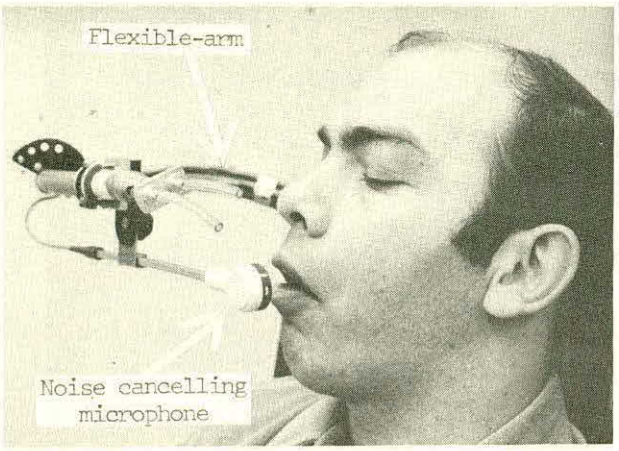

Figure 1.

Study C Results

All but three subjects had excellent final recognition rates comparable to those in industrial installations. Failure of the three to obtain excellent recognition appeared to be due to: slow speech patterns, insufficient word boundaries or inability to self-monitor consistent voice volume levels. Words most frequently requiring alternates were: stop, jog, brake, left, right, record, rewind. Mode of operation was easily learned by all subjects. All used the headset microphone and all required occasional readjustment to maintain recognition.

Table 2. POPULATION PARAMETERS

	STUDY A	STUDY B	STUDY C
DIAGNOSIS			
Hi-Level S.C.I.	13		7
Guillian Barre	1		
Duschene M.D.	2		
Frieder. Ataxia	2		
Congenital C.P.		10	
Able-bodied			5
DISABILITY			
Severe Physical	18	10	7
Speech Impair.	2	10	
AGE			
16-26	14	6	3
27-45	3	3	8
46-65	1	1	1
SEX			
Male	17	5	9
Female	1	5	3

DISCUSSION

Analysis of test results indicate that recognition problems were almost exclusively related to either loss of optimum microphone position or operator stress elicited by specific aspects of control of the peripheral devices.

Table 3. TRAINING & RECOGNITION

Diagnosis	No. Hours / No. Sessions	Voice Level: Soft	Medium	Loud	Normal LED Illuminated During Train: All	50%	50%	Rejected Standard Vocabulary: None	1-2	2-5	Many	Vocabulary Requiring Alternate Words or Retraining: *By code--see Table 1	Recognition with Final Vocabulary: ** W/C	ECS
Study A														
SCI	12 5	x				x				x		3,9,10,11,12	U	E
	4 3	x			x					x		3,9,10,12	S	-
	4 2½	x								x		3,9,11,12	S	E
	9 8	x				x				x		3,9,10	S	E
	4 4	x				x				x		3,9,10	S	-
	9 9	x				x				x		8,27,30	S	E
	1 1	x			x				x			27	-	E
	1 1	x				x					x	Most	-	U
	1 1	x				x		x					-	E
	1½ 1	x				x			x			3,10	*E	E
	1 1	x				x			x			27	-	E
	4 2		x		x				x			3,27	U	E
	3 3	x				x					x	All	U	U
GB	39 47	x				x			x			9,27	S	E
MD	2 2		x		x					x		3,9,10	*E	E
	1 1	x				x				x		17,20	-	E
FA	1 1	x				x					x	4, most	-	U
	1 1	x				x			x			4	-	E
Study C														
NONE	½ 1	x				x					x	1,3,6,7,11,14,15,16,17	U	S
	½ 1	x				x					x	3,4,7,13,17,20,30	U	U
	½ 1		x		x					x		3,9,16	E	E
	½ 1	x			x					x		2,3,9	E	E
	½ 1	x			x				x			9,17	E	E
SCI	1 2	x				x					x	1,3,5,7,18,20,21,26,27	S	S
	½ 1	x			x					x		1,3,9,10	E	E
	½ 1	x			x					x		1,10,13,16	E	E
	½ 1	x				x			x			10,13	E	E
	½ 1		x			x					x	1,3,10,13,17,24,30	U	S
	½ 1	x				x			x			13,16	E	E
	½ 1	x			x					x		0,9,16,17	E	E

*Optimum commands for underlined functions had not been identified at end of trial.

**E Excellent
S Satisfactory
U Unsatisfactory
* Non-operational

Initially there was concern that recognition was being affected by inadequate speaker volume. However, able-bodied speakers of C with equally low volume (i.e. same display verification and amplification settings - see Table 3) had reliable recognition. Voice fatigue was identified as the problem for one SCI subject.

Subjects in all studies experienced increased rejection rates and/or total loss of recognition due to variation of mouth-to-mike distance. Close spacing is essential for noise cancelling microphones which are required in noisy environments. Able-bodied subjects in C spontaneously repositioned the mike when recognition began to be affected. In studies by others headset mikes were projected as a solution to noise background problems, with reduced cosmesis and functional interference with other activities being the only foreseen disadvantages (4).

Operator stress was ultimately realized to be the factor in almost all remaining recognition problems. An examination of the most frequently rejected commands (Table 3) provides this conclusion.

Although "stop" and all of its alternates were reliable during non-operational trials, they were the most frequently rejected during wheelchair driving and telephone dialing. Similarly, "right" and "left" were reliable during non-operational trials, but were frequently rejected during driving, especially when turns were attempted while the chair was already moving forward. Commands for turns from a "stopped" status were not difficult for experienced operators. Subjects expressed that the need to input some commands at very precise times-- as was characteristic of all the frequently rejected commands--was undesirable and anxiety provoking.

It was encouraging to note that the VRS accomodated a wide range of intellectual abilities and speech impairments. No boundaries regarding intellectual levels were noted. Mild to moderate dysarthria and scanning speech should not be considered contraindications for voice control. Within the limitations of command vocabulary, severe degrees of dysarthria may be accommodated.

Many psychological and physiological questions, raised by our studies, can only be answered by further controlled and continuous clinical evaluations of many months duration. Voice fatigue, respiratory insufficiency and the effects of breath noise due to respiratory involvement may affect the quality of voice control when used to accomplish all or the majority of tasks of any job. Although previous discussions suggested the use of voice input as a replacement for keyboard input (5,6), our data suggests that the use of limited vocabulary, isolated word systems, can become physically and psychologically fatiguing.

Isolated word systems are acceptable when they trigger a sequence of functions or where there is no other solution. In the future, continuous speech recognition systems for unrestricted vocabularies and syntax will facilitate general use. Our results suggest that voice currently has as its most appropriate application the control of tasks which parallel those which have been utilized successfully in industrial applications. In those situations the users accomplish some but not necessarily all of their tasks by voice control.

CONCLUSION

Clinical trials have demonstrated that in the absence of stress, persons with various physical limitations and a wide range of intellectual levels can produce word commands which can be recognized by speech recognition instrumentation. The following are proposed as future efforts which can result in the ultimate successful utilization of such technology by persons with severe physical limitations. 1) Developmental work to eliminate the weak microphone-support link. 2) Controlled, operational trials to assess the physiological and psychological effects of the continuous utilization of speech recognition equipment as applied to work tasks by a spectrum of diagnostic speakers. 3) Efforts to identify jobs or educational activities which have specific tasks for which voice control may be optimum.

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HEAD MOVING MICRO-LOOKSIGHT SYSTEM AND POCKETABLE VOICE SIGNAL SYSTEM COMMANDED BY POCKETABLE MICROCOMPUTER, AND ITS APPLICATION ON THE COMMUNICATION SYSTEM TO SEVERELY HANDICAPPED

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Abstract — The command systems stated here were developed for the purpose of controlling electrical prosthesis.
The advantages of these command systems are that the intention of the user (handicapped persons)can be transmitted to prosthesis correctly and the user has no burden to operate the command system (he can manage it easily), which has formed so-called man-machine system.
One is head moving looksight system and the data quantity of communication enables 9 but 9 (=729 species)for practical purposes. The other is pocketable voice command system which contains in a metal box measuring 70 x 145 x 20 mm, and about 36 codes are utilized for the orders of electrical prosthesis. And about 2 days of training are required to enable use of prosthesis.

I. INTRODUCTION

Command system must satisfy the following conditions:

i) The intention of user can be transmitted correctly to controlled system.

ii) In case of transmission of intention, the burden of user (physically, mentally) should be few as far as possible.

iii) It is easy to connect the user with command system and to detach mechanically.

iv) There is not loss of time from transmission of intention to excution.

v) Command system should be pocketable.

As the command systems satisfying these conditions, we have developed head moving looksight system which selected menu prepared previously from head moving, and voice command system which selected codes prepared previously from pitch extraction of voice vocal signal (humming).

Head moving looksight system selects the menu of CRT display by the motions of direction of head rotation (X axis) and inclination of head moving up and down (Y axis) (the range of 40° in maximum).

The menu of one screen has 9 divisions and new menu of 9 divisions comes out again after selection of the one, which enables to 9 . Therefore, a dialogue system with CRT display was taken.

Voice command system is the method extracting pitch of the vocal chord and discriminating from loudness and change of sound (intonation) by attaching throat-microphone to the wall of Cartilago thyroidea. 36 kinds of code are utilized now to control prosthesis.

II. HEAD MOVING LOOKSIGHT SYSTEM

Fundamental principles

The space coordinate of position was detected in control of electrical prosthesis, but communication system linked head rotation and head moving up and down with the coordinate system of CRT relatively.
Head rotation was detected by electrical direction indicator (flux sensor) which standardized horizontal force component of the magnetic field of the earth,and the angle of inclination against gravity component was detected by electrical inclinometer concerning head moving up and down.

Fig.1 shows the coordinate system attached flux sensor and inclinometer.

Flux sensor as the change of Z axis rotation,

$$F_1 = e \sin\theta_1 \qquad .....(1)$$
$$F_2 = e \cos\theta_2$$

Inclinometer as the change of X axis rotation,

$$I_1 = e \sin\varphi_1 \qquad .....(2)$$

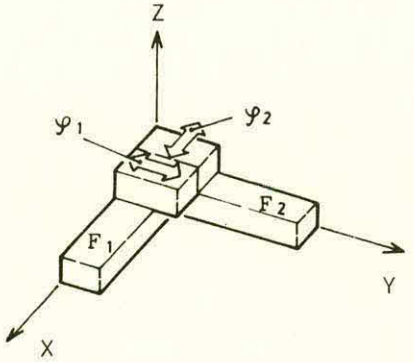


Fig.1 Coordinate system attached each detectors

as the change of Y axis rotation,

$$I_2 = e \sin\varphi_2 \qquad \ldots\ldots (2)$$

Correction calculation

The magnetic field of the earth varys according to places and it also varys according to near magnetic material and non-magnetic material in case of the room. On this account, we make the correction of data as the output change differed according to places.

Now, by (2) equation,

$$\cos\varphi_1 = \sqrt{1 - \sin^2\varphi_1}$$
$$\cos\varphi_2 = \sqrt{1 - \sin^2\varphi_2} \qquad \ldots\ldots (3)$$

and the correction for head inclination,

$$\cos\theta = \frac{\cos\theta_1 + F \sin I \sin\varphi_2}{\cos\varphi_2}$$
$$\sin\theta = \frac{\sin\theta_1 + F \sin I \sin\varphi_1}{\cos\varphi_1} \qquad \ldots\ldots (4)$$

where, F shows the whole magnetic force and I shows an angle of declination, which are values decided previously according to places.
Besides this, the measured error of horizontal force component is corrected in the moment of detection of peak value.

Consequently,

$$\Delta H = \sin^{-1}(\theta - \theta_0)$$
$$\Delta V = \sin^{-1}(\varphi - \varphi_0) \qquad \ldots\ldots (5)$$

From (5) equation, the change of angle is obtained, where φ_0 and θ_0 are standard angle settled early. And θ and φ change continuously for this angle.

The microcomputer of user side calculates to (5) equation and judges in what division the output (ΔH, ΔV) exist.
Fig. 2 shows the connection of the output value (ΔH, ΔV) corresponding to the angle of head rotation with 9 divisions.
Each division is difined by codes from 0 to 8.
The display input/output unit side calles up the information prepared in floppy-disk according to difined code, and expresses in monitor TV.

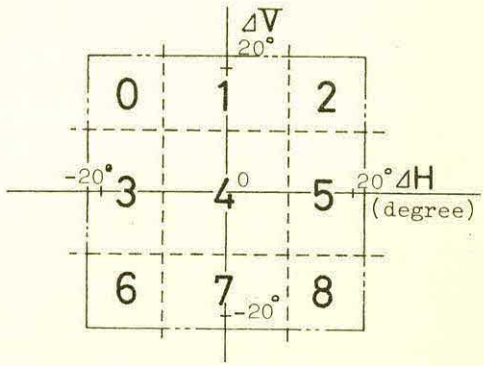


Fig. 2 The area of looksight detector and defineded codes

Environmental controller applicating looksight system

Fig. 3 is the block diagram of head moving looksight system as an environmental controller.

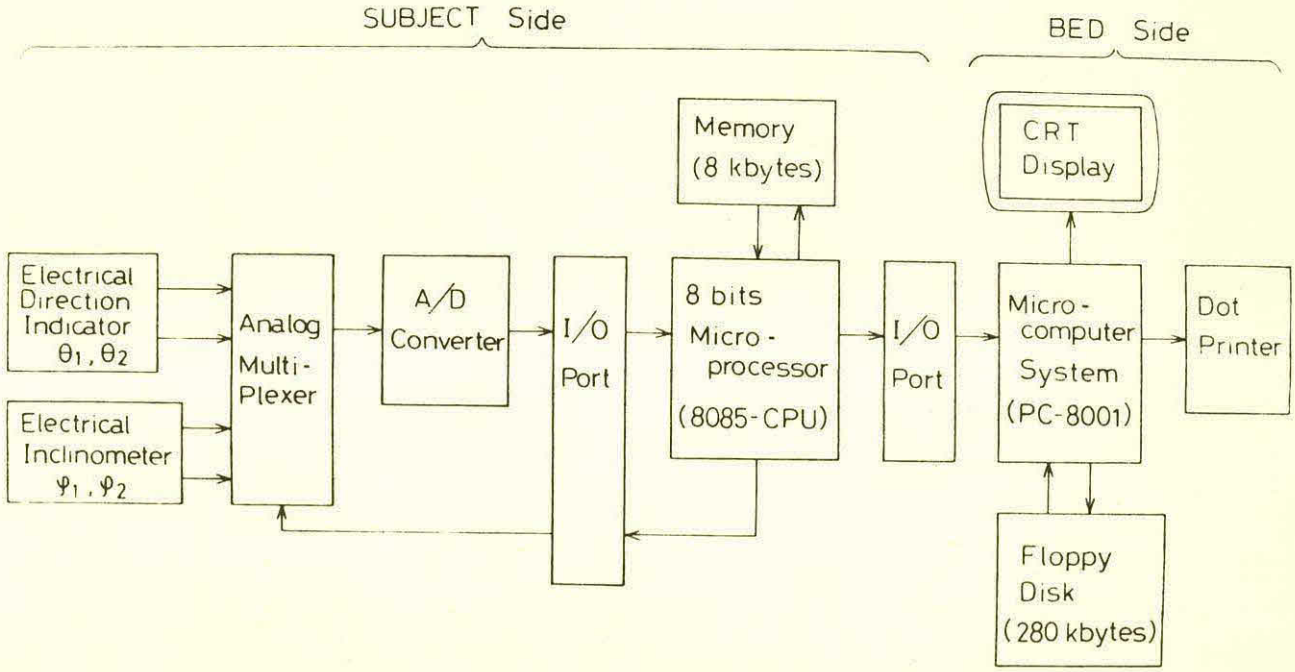
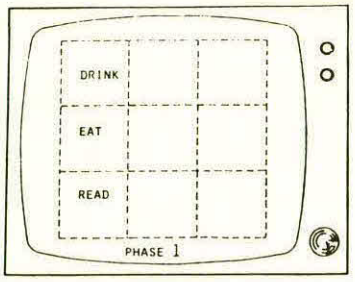
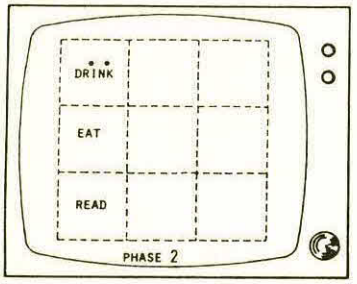


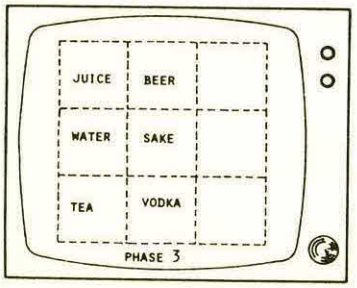
Fig. 3 Block diagram of head moving looksight controller system



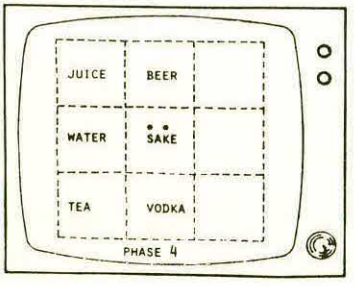
ques: What do you want to do ?
(ans: Drink)



* * Sign go on and off (All right)



ques: What do you want to drink ?
(ans: Sake)



* * Sign go on and off (All right)

Fig.4 The information of menus feed to user

The device of user side is composed of each sensor and microcomputer control device. On the other hand, there is a display input/output unit which is composed of monitor TV, microcomputer, floppy-disk and printer, which is fixed to the bedside. The microcomputer of user side calculates to (5) equation, and generates power continuous values of ΔH and ΔV.

The standard screen has 9 divisions so that the user may see it easily.
From the resolution power of sensor and head moving, the screen is possible to 36 divisions. This display of input/output unit has menus as shown in Fig. 4, and its capability may be used to directly control external devices such as TV, radio, lights or other appliances.

(Phase 1) The items of request are expressed on the screen and the condition of display input/output unit will wait for next command signal.

(Phase 2) The user acts head rotation and angle motion corresponding to a division on the screen which user wants. Voice command or mechanical switch feeds the command code of looksight controller in the position.

(Phase 3) If the user selects the "DRINK" command, for example, the screen will sign** go on and off. And then, the screen will express the concrete contents of the "DRINK" (juice, water, tea, beer, etc.).

(Phase 4) If the user selects the "SAKE" command, the screen will sign go on and off, and give next command to another device directly.

Pocketable voice command controller

Fig. 5 shows an appearance of pocketable voice command controller. The controller used a microprocessor of z 80-CPU type, RAM 256 bytes and PROM 2 kbytes with display includes the program of 65 command codes and 90 trainer menus.

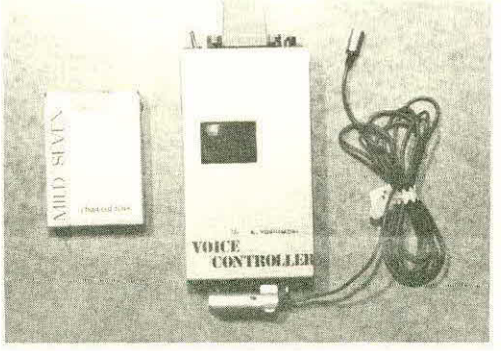


Fig. 5 Appearance of pocketable voice command controller

Table 1 . The menu of total arm prosthesis movement

Number of Mora	Accent (Example of movement) 0	1	1	1	1	2	2	2	2
1	(stop)					o ; Short humming ⬭ ; Long humming ; Pattern possible to use			
2A	(restart)	(hand horizon)	(reset) horizon)						
2B	(normal)	(fast)	(slow)						
3A	(central point)	(under left)	(upper left)	(upper right)	(under right)	(to mouth1)	(onto table)	(push button)	(turn knob)
3B	(eating)	(combing)	(shaving)	(tooth brushing)	(original position)	(grasp)	(let go)	(drinking water)	(sipping soup1)
3C	(pause)	(A-point)	(B-point)	(C-point)	(D-point)	(to mouth2)	(hair brushing)	(soup 2)	(soup 3)

It is generally said that the extraction of vocal vibration from the wall of trachea is the one of methods to extract the pitch of vocal chord. The voice controller extracts voice vibration by throat-microphone from the wall of trachea near the larynx, and obtains the loudness of pitch and smooth change of sound (intonation).
The method is to distinguish the change of pitch and intonation from the mora and the change of accent and pitch standardized previously.

Utterance is performed by humming, and recognized on condition of constant accent for the pattern decided previously. Therefore, there is not correct musical interval and rule, and it is recognized on condition of constant gamut.

Table 1 shows the correspondence between the pattern which is easy to utter and the movement decided by microcomputer control device from the connection of accent standardized with mora.
The pattern prepared is about 65 species and was actually utilized 36 species.

CONCLUSION

One of the difficult problems in designing and operating command system is to impose a burden on the user.
Voice command system is suitable for the command signal such as the connection between each device and user, the change of menu, standby, pause and emergency stop. Therefore, we used jointly voice command device and head moving looksight command device.

As an environmental system,

i) The standard command signal of head moving looksight controller has 729 species defined by 9 .

ii) The menus defined by 9 can be increased by memory capacity.

iii) By dialogue system the user and menu of display input/output are selected.

iv) In this voice command method, the user hums a series of "sylables" (differentiated by pitch and duration) into a throat-microphone. As a result, the control device has become pocketable, has been designed and fabricated.

v) A voice trainer was also developed; about 2 days of training are required to enable use of the electrical prosthesis.

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The author also wish thank all the members participating on this special project research.

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MICROCOMPUTER BASED COMMUNICATION SYSTEM

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ABSTRACT

An inexpensive microcomputer based prosthetic communication system designed specifically to meet the needs of functionally nonverbal physically handicapped users is described. Computer/"real world" interfacing and BASIC program are explained. Educational and recreational benefits of microcomputer systems are examined and future plans outlined.

Background

The microcomputer revolution has provided the handicapped with an exciting and extremely powerful new tool. Today a sophisticated computer system readily adapted for the handicapped user can be purchased for less than $700.00. Affordable microprocessor based "intelligent" consumer products are appearing in increasing numbers and hold tremendous promise for reducing dependency. A major frustration facing todays habilitation/rehabilitation engineer is finding enough time to explore and keep up with current technologies. Cost and availability are in most cases no longer top priority concerns.

The Northwest Louisiana State School Habilitation Engineering Department is investigating the potential of microcomputers for multiply handicapped, severely and profoundly retarded persons. As of this writing, all efforts have focused on using the Commodore PET 2001-8 and Rockwell AIM 65 microcomputers. Both machines are the personal property of a school employee and most programs and hardware modifications have been developed independently of state employment.

The full typewriter-style keyboard, 20 character alphanumeric LED display, dual cassette recorder interface, on-board 20 column thermal printer, romed 8K basic and no-fuss interfacing make the AIM 65 an excellent candidate for habilitation/rehabilitation applications. A total package system complete with enclosure, power supply and basic ROM can be had for as little as $600.00

Interfacing

Most habilitation/rehabilitation applications require the addition of external switches tailored to the physical abilities of the handicapped user and necessitate real-world/ computer interfacing. The AIM 65 and PET computers both have extremely flexible interface hardware on-board making child's play of what can easily become a stumbling block when working with other machines. All connections to the AIM 65 are made via a 6522 versatile interface adapter chip (VIA). A minimum configuration requires only two connections (See Figure 1). Sound requires the addition of two more wires and an inexpensive speaker - amplifier. If the control switch exposes the handicapped user to possible electrical contact with the computer a simple battery powered optical isolator or reed relay can be inserted between the control switch and 6522 input/output port (See Figure 2).

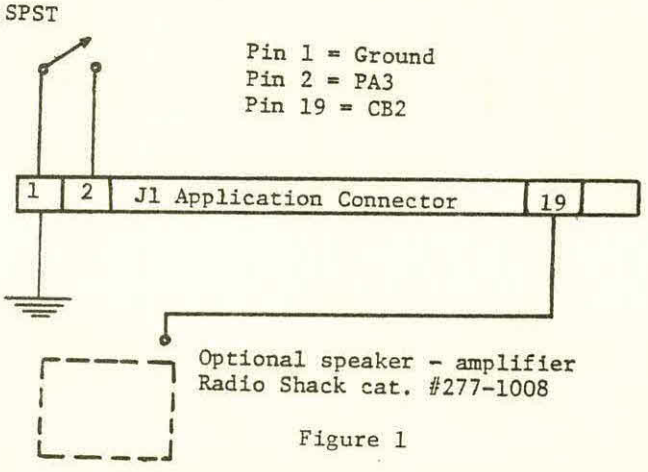

Figure 1

CONTROL SWITCH/AMPLIFIER CONNECTIONS

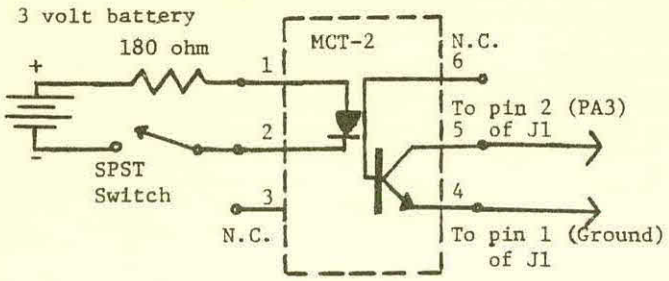

Figure 2 SAFETY ISOLATION

At the conclusion of this article is a listing of an AIM 65 program designed to enhance communication for speech-handicapped persons. The system consists of an AIM 65 microcomputer with 4K of RAM memory, 8K ROM basic, power supply, 44 pin edge connector, compatible cassette tape recorder, momentary contact SPST switch, battery powered speaker-amplifier and Blissymobolics Communication Foundation, 10 x 10, 100 vocabulary Bliss board. Other communication boards and vocabularies may of course be used providing responses can be identified via vertical and horizontal coordinates and the vocabulary listed in data statements starting at line 1000 are replaced accordingly.

Program Operation

Operation is simple and straightforward. When the program is run, the computer responds with "SCAN RATE?". The number entered in response to this query will determine the speed at which vertical and horizontal coordinates are displayed. The larger the number the slower the scan rate. Values between 50 and 60 have proved workable for most of our physically handicapped users. Once the scan rate has been entered and the RETURN key pressed, the numbers one through ten are alternately displayed on the left side of the LED panel and a "beep" is emitted through the attached speaker-amplifier. If, for example, the user wished to communicate the word "help", he would first locate the corresponding symbol and activate the control switch when the appropriate numeric and alphabetic coordinates are displayed, i.e., 7,D. Therefore, when the number 7 is displayed, the user momentarily closes the control switch. The computer responds by emitting a high-frequency "beep" and beginning sequential display of the letters A-J. When D is presented, the user again closes the control switch. The computer generates a short tone signaling recognition of his selection, displays the English equivalent of the symbol in the approximate center of the LED panel and resets to numeric scanning. If the user wishes to print a displayed word, he activates the control switch when either the 9,A or 10,A coordinates are displayed. In the special case of "help" an auditory alert is sounded until the user again presses the control switch.

The basic coordinate scanner program is being expanded to provide enhanced editing and print formating capabilities. Future plans include the development of a rechargeable battery power supply and ruggedized enclosure suitable for wheelchair mounting.

As of this time, two functionally non-verbal residents, classified as either profound or severely retarded and physically handicapped, have learned to use the AIM 65 communication scanner. The computer generates an immediate translation and written record of the user's responses. Early observations suggest that instantaneous translation of Bliss into English reinforces acquisition of English. The potential of this system as an educational tool beyond establishing viable communications appears great.

In addition to practical uses in communication and education, the AIM 65 and PET computers have been very well received as a source of entertainment. The flexibility and accessibility of these machines allow the development of both individual and group games which can accommodate a wide range of physical and mental limitations. The computer serves as an equalizer - making it possible, in some cases for the first time, for handicapped persons to play games with one another without outside assistance.

Program Listing

```
LIST
 100 REM* HELP
 105 POKE 40963,0
 110 S$="ABCDEFGHIJ"
 115 INPUT"SCAN RATE"; SR
 120 FORX=1TO10
 130 FORY=1TO10
 140 READ D$(X,Y)
 150 NEXTY:NEXTX
 160 REM SCAN ROUTINE
 170 FORX=1TO10
 180 PRINTXTAB(5)L$
 185 GOSUB2100
 190 FORZ=1TOSR
 200 P=PEEK(40961)
 210 IFP<255THENGOSUB2000
 220 IFP<255THEN250
 225 NEXTZ
 240 NEXTX
 245 GOTO170
 250 REM Y COORDINATE
 260 FORY=1TO10
 265 M$=L$
 270 PRINTMID$(S$,Y,1)
 275 GOSUB2100
 280 Z=Z+1:P=PEEK(40961)
 290 IFP<255THEN350
 300 IFZ<SRTHEN280
 310 Z=0
 340 NEXTY
 345 GOTO250
 350 L$=D$(X,Y)
```

```
 355 GOSUB2000
 360 IFL$="---"THENPRINT!M$:L$=M$
 362 IFL$"HELP"THENGOSUB2200
 365 GOTO170
 1000 DATA0,1,2,3,4,5,6,7,8,9
 1005 DATAHELLO/GOOD-BYE,?,I/MY,LIKE,HAPPY
 1010 DATAACTION,FOOD,PENCIL,FRIEND,ANIMAL,
PLEASE,WHY,YOU,WANT
 1015 DATAANGRY,MOUTH,DRINK,PAPER,GOD,BIRD,
THANKS,HOW,MAN
 1020 DATACOME,AFRAID,EYE,SLEEP,BOOK,HOUSE,
FLOWER,I'M SORRY
 1025 DATAWHO,WOMAN,GIVE,FUNNY,LEGS,TOILET,
TABLE
 1030 DATASCHOOL,WATER,OPPOSITE,WHAT,FATHER
 1035 DATAMAKE,GOOD,HAND,PAIN,TELEVISION,
HOSPITAL,SUN
 1040 DATAMANY,WHICH,MOTHER,HELP,BIG,EAR,
CLOTHING,NEWS
 1045 DATASTORE,WEATHER,MUSIC,WHERE,BROTHER,
THINK
 1050 DATAYOUNG/NEW,NOSE,OUTING,WORD,SHOW,
DAY,---
 1055 DATAWHEN,SISTER,KNOW,DIFFICULT,HEAD,
CAR
 1060 DATALIGHT,ROOM,WEEK-END,---,HOW MANY,
TEACHER
 1065 DATAWASH,HOT,NAME,WHEELCHAIR,GAME/TOY,
STREET,BIRTHDAY
 2000 REM ALERT
 2005 Z=0
 2010 POKE40971,16
 2020 POKE40970,15
 2030 POKE40968,200
 2040 FORU=1TO500:NEXT
 2050 POKE40968,0
 2060 RETURN
 2100 REM BEEP
 2110 POKE40971,16
 2120 POKE40970,15
 2125 FORV=1TO10
 2130 POKE40968,100
 2135 FORF=1TO20:NEXTF
 2140 POKE40968,0
 2150 NEXTV
 2160 RETURN
 2200 POKE40968,65
 2205 PRINTTAB(5)"HELP"
 2210 P=PEEK(40961)
 2220 IFP=225THEN2210
 2230 POKE40968,0
 2240 RETURN
```

Program Remarks

1. Line 105: Sets the A data direction register to input mode.

2. Line 110: The S$ string variable defines the the horizontal coordinates and may be replaced by any set of 10 alphanumeric characters. For example, "0 1 2 3 4 5 6 7 8 9".

3. Lines 120-150: Reads and defines D$ string variables listed in data statements starting at line 1000.

4. Line 180: Displays the L$ string variable which holds English translation of selected Bliss symbol.

5. Lines 190-225: Looks at A side of the VIA. If the control switch is closed, pin 2 is brought low, the variable P is set to 251, the L$ string is defined by the current X,Y values and the "beep" subroutine is called.

6. Line 360: If the L$ string array is either 9,1 or 10,1 (i.e.,---) the M$ string is printed. The M$ string was set equal to the preceeding L$ string variable in line 265. The L$ string is also set equal to M$ to ensure that the current English translation is displayed.

MOBILITY

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Abstract: A more comprehensive view of what constitutes "mobility" is offered and examples are cited which indicate what is being done at the University of B.C. in relation to this view of mobility. Three items are considered: (1) wheeled mobility with power; (2) vertical mobility by means of a lift module; and (3) supporting the disabled person for optimal function in relation to mobility.

Imagine yourself lying on a surface, powerless to change your position in any significant way. Consider what mobility would mean to you then. Eventually you must see it as the means by which you would position yourself at any endpoint and position yourself at any angle you wished. You would see mobility in terms of moving forward and backward, from side to side, up and down, while seeing it also in terms of such angular tilts around the various axes of your body as you required. With these six degrees of freedom you could improve your comfort, improve sensory access to the environment, get needed feedback from it and optimize your position for functional processes. It is in terms of six degrees of freedom that mobility should be viewed. Nor should consideration be limited to the gross and obvious movements, but include also such movements as need to be possible to adapt any support surface to changing needs, such as the need to relieve pressure or enhance circulation or relieve pain. Looked at in this way, one can see that many gaps exist in the present systems which are designed to deal with the mobility needs of disabled persons.

Considering mobility in such a comprehensive way and intending to digress toward such simplifications as practical constraints might impose, we conceived of a "lift-save-chair-bed", a system which would provide (a) planar mobility, (b) vertical mobility and (c) support in a variable way on a surface adapting to the specific needs of the user. As we envisioned it, electronic aids would facilitate sensory inputs and would facilitate control over the environment for functional and social gains.

In the simplification accepted as a basis for design, the same functions as are commonly found in typical powered wheel chairs were the ones we would include for planar mobility. We selected as the target group small children who would not be given power. We expected such a capacity to liberate the parent or child from such dependencies as would otherwise exist in a variety of situations. The child would be able to follow the mother at will. A more comfortable position could be adopted without aid. A new element of play would be added. Enhanced social value would accrue. We accepted that we would tackle vertical mobility apart from such things as angular tilts, sticking to the idea that the vertical mobility ability would serve most those people who must now be lifted for transfer and who are a particular problem because of their weight or flaccidity. The problem of designing an adequate support surface would be tackled last, and it would be tackled as a static system, perhaps hinging at strategic locations to allow improved postures only. All three areas related to mobility have been tackled and are at different stages of development. When each system has advanced sufficiently, it should ultimately be possible to combine two or more of these elements into such configurations as suit the needs of particular people, or to use each separately. This discussion describes the status of each of the systems and considers how they relate to the best meaning of the word "mobility" that a needing person might provide.

1. MOBILITY POWER UNIT (MPU)

The MPU consists of two electric motors driven by gelcel batteries from which the power is controlled by means of a proportional control system activated by means of a model airplane joy stick. The final configuration of the present MPU has evolved out of two previous prototypes and represents, as far as we can see, a nearly optimized example of the basic approach

we have taken to produce a simple friction drive system which can be attached to any wheeled device of the sort familiar in a disabled person's environment. Among the criteria defined for evolution of the MPU were that it permit folding of a standard wheel chair for auto trunk storage, be light enough for a woman to get it into the trunk, be under $500. to make and be organized to allow hand or push drive without removal of the power unit or modification to the wheeled device. It was also a requirement that the MPU should be suitable for application in the clinic, not requiring any modifications to be made to the vehicle or requiring very special techniques or tools for its application.

Figure 1 shows the system attached to a standard adult wheel chair.

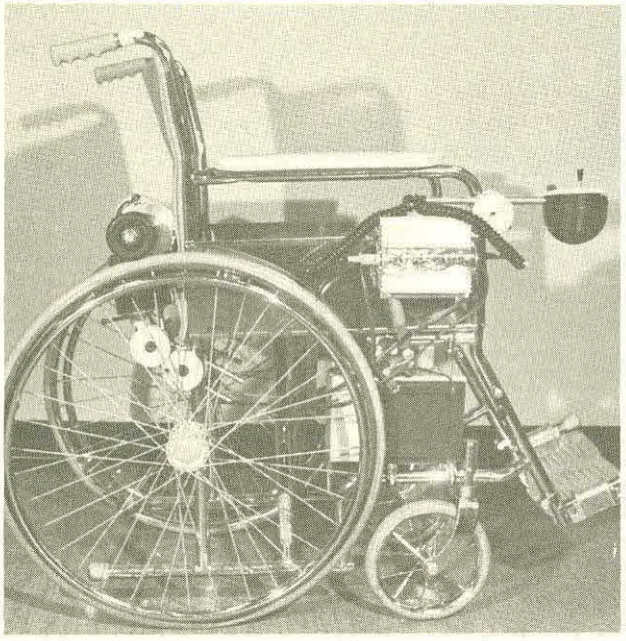

Figure 1 - MPU

The design is sufficiently easy to adjust and modify that different components could be substituted for the ones we have used, or additional batteries could be attached to extend the period of use or the power delivered at the wheels. This design flexibility has been achieved by separating out and making adjustable, the elements which comprise the MPU. We have referred to the system as a "multipack" system as opposed to the "unipack" system previously designed where all the main elements were in a single box and hung, one on each side, against the wheels to drive the chair. In addition to making design flexible so that improved elements can be easily incorporated, the multipack approach allows the system to be adapted to a variety of wheeled devices. It is this format that we would claim is of value rather than the particular system or particular components used. However, there is no escaping the fact that the components also affect the format. The Winnipeg clamp, designed to attach various items such as trays to a wheel chair, influenced us to create an arm-like support for attaching the motors. Position is secured by locking the support at elbow and shoulder so to speak. This permits quite accurate positioning of the drive wheel against the wheel of the device. Another mandatory adjustment is alignment of the axis of the drive wheel with respect to the axis of the driven wheel. This is achieved by means of an attachment bracket which can be rotated around the long axis of the motor body when the holding clamp is loosened (a hose clamp). The tubular sections of the Winnipeg clamp pass through the bracket and through springs. One end of a spring is fixed to the tube and the other rests on the bracket in compression. Forcing the tubes up or down in the locking block of the clamp allows adjustment of the spring tension and hence the force with which the friction drive wheel impinges on the driven wheel. Between the flat surface of the bracket to which the motor is attached, and the disc-shaped Winnipeg clamp, is a cylinder of plastic with an extending lever using the support tube as axis. The surface which is against the Winnipeg clamp is inclined so that as the cylinder is rotated, the long side of it travels up the clamp elevating the bracket-motor assembly to disengage the drive wheel from the driven wheel. This allows hand operation of the chair.

Color coating, anodizing and the use of stainless steel coupled to interesting shapes and positions contribute attractiveness of the system.

Under the chair in Figure 1 can be seen the batteries and the drivers of the electronic system. The method of attachment offers a degree of optionality few other ways could offer. Taut wires connected to opposing stable elements of the chair are used as the base to which the components are clamped. This method of attachment gives a degree of freedom which we found to be necessary for attachment to highly adjustable wheel chairs in which rigid attachment to single points often leads to clashing between parts when adjustment options are used.

The drive wheel is a modified rubber caster wheel. It functions well because the system is a "low demand" system. Special design of the drive wheel to prevent slippage would extend the capabilities of friction drive somewhat, but such a drive system has inherent limitations and we feel that we have come close to the limits for such a system. Beyond this either gear or belt drive would be considered. An untried option would be to use a track-like drive system which "wrapped" around a segment of the driven wheel. But as it is, friction drive should be avoided except for low demand systems except when drive is directly against the ground, or special design of driver and driven wheel can be allowed.

Two gelcels drive the two motors, windshield wiper motors, flat out for about an hour. When the target group intended is considered, this can offer sufficient energy to match their needs and capabilities, the environmental circle being small.

The motors themselves however are sufficiently robust to drive an adult briskly when adequate power is applied and the drive wheels are loaded sufficiently to secure friction. With the pair of batteries a 100 pound person can travel at a slow walk. Weight of the system under these conditions is 35 pounds over and above wieght of the person and vehicle.

Given this as a suitable starting point, we foresee that such inherent problems as might surface can be worked out on an evolutionary basis as the system is applied for what it will do. In terms of the time-scale established for development, the system is in PHASE II, Step 8 of hypothesis testing and close to optimized. Passage through Phase III, transfer to use, will not be costly for that reason. Then, a light weight, inexpensive, adaptable, portable low activity level system will be available for use as a training system, a toy and functional item in the life of the small person otherwise dependent.

2. THE PORTABLE LIFT MODULE (PLM)

Vertical mobility can most easily be associated with the problem of lifting and transferring heavy persons from one support surface to another. For such transfers to be facilitated, there needs to be a translatory element associated with the vertical displacement system, or it would be well to have the vertical lift system programmable so that the surface could be tilted to facilitate the transfer. Such terminal functions would put more independence into the hands of some disabled people.

We took as our first task the simple problem of vertical displacement. Among criteria governing design were that the device be portable, inexpensive, capable of load bearing up to 400 pounds, safe (pinch, collaps, electrical), suitable for use in combination with various support surfaces and bases, attractive and of dimensions which suited the environments it would be likely to be used in.

Figure 2 shows the PLM in its present form. Planned was a single module (half that shown) which could be ganged in a variety of ways to achieve the required range and required base of support or variations in elevation at different points. The triangular configuration provides rigidity against oscillations under load. A cam arrangement initiates lift at the beginning when the system is compacted down to approximately 6 inches from the ground. At full height the support surface can be 30 inches higher than its starting level.

Besides the rigidity imparted by the triangular configuration, the expectation was that it would improve access to the user and to wash basins or toilets for hygiene.

While the model shown is hand actuated by means of a hydraulic pump, it is anticipated that any drive option would be possible in a final design. The PLM is at Step 6, PHASE II on our time scale and is sufficiently funded to place it at Step 9, test batch production (Phase III) by the end of 1981. By then there will be approximately 10 systems in use for persons who must be lifted frequently and are transfer dependent.

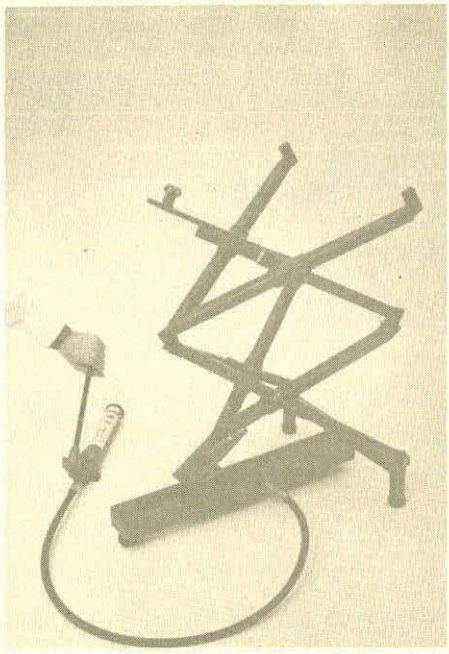

Figure 2 PLM

This means that initially it will be tried in institutions where lifting is a problem. The ultimate aim is for a device that can be used in a great variety of situations by making it easily portable, even industrial . We anticipate making some versions sufficiently small for inclusion in wheelchairs for example. We also anticipate programmable pivots which will allow angulation of the support surface such that transfer or security of the user is enhanced. Such a system could be used in association with other equipment than that used by disabled people. The final outcome expected is a very light weight device in sizes to suit various situations, driven according to what is convenient or available, and suitable for linking to other elements to derive such combinations as suit the needs of particular disabled persons or services. Not in use, it might well hang inconspicuously or even beautifully on the wall, waiting to be called. Called, it should become operational with a minimum of demand on the operator so that direct application in the clinic is possible so that not only the needs of potential users can be directly met, but assessments can be conveniently carried out.

3. A SUPPORT SURFACE

The support surface being developed by S. Cousins for seating severely disabled children illustrates well the approach which must be taken

if the clinic is not to be encumbered with time delays and voluminous specialized items which must be stored if seldom used, or which, because of costs, cannot be stocked for ready use, or even considered at all. Figure 3 shows a seat constructed of cylindrical and spherical beads strung on wires which can be tightened to lock the surface in the required shape. Shape is imparted with the wires semi-slack and then tightening secures it. Adjustments at any location can be made by loosening a few wires, adjusting the shape and retightening the wires to secure it.

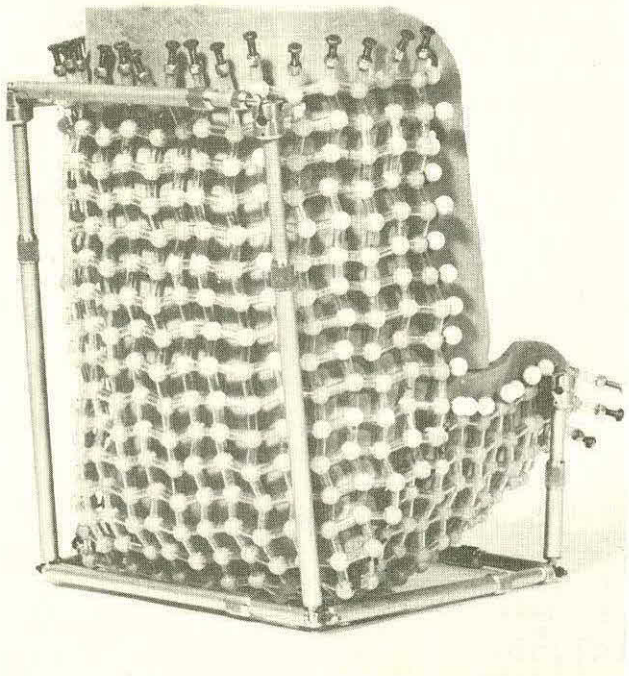

Figure 3 SUPPORT SURFACE OF ELEMENTAL PARTS

This innovation is quite new, and it be appreciated that a great variety of shapes for elemental parts would be possible to derive special effects. Such an approach to support surfaces in general, be they for seating or for the sockets of braces or prostheses, is worthy of extensive effort on the part of designers, design being aided by use of computers. The particular set of components used here to illustrate the principle is of much less importance than the concept itself. What is illustrated is the first step on a new course of design that will greatly facilitate clinical services. While the system shown could be said to be at Step 8, Phase II of our time scale (clinical evaluation), I see it as a very initial step into a new field of action related to the solution of such problems in rehabilitation as are shape-dependent. In the evaluation process, it has been used on eight children so far with results which confirm the viability of the approach. Lacking in the present system is the capacity to make the support surface dynamic - a lack common to all such surfaces currently being designed or used. But at least the prospects for such a capacity can be anticipated when such an elemental component as a bead is used to form the complex surface. One can imagine "sensor beads" or beads of different sizes, shapes or qualities, beads inherently padded, or between which special components fit for padding or adding functions.

4. CONCLUSIONS

While the gaps in the systems presented can be appreciated, the gaps in the whole array of rehabilitation devices related to mobility as defined here may not be so easily appreciated. We see a welter of "solutions" which seem to have limited underlying principles. Certain functions are made central often to cause losses with the gains. Wrong starts are also leaving gaps. To sift out all that has been done to make the disabled mobile, requires that we look at mobility in a very comprehensive way, a way that includes not only wheeled mobility, but vertical mobility and mobility on the smaller scale such as would be present in support surfaces which are responsive to various biological needs. We need to consider the variety of environments within which and between which the user moves. Systems which confine themselves to particular environments, systems which exclude themselves from the clinic or the home, systems which clutter up the environment - home or institution - systems which require special transport or ignore possible use of public or private conveyances are incomplete, even destructive of optimal solutions becuase they exhaust resources which would be applied in other ways to derive required solutions and force design of auxilliary systems which will suppliment them.

Mobility is obviously a prime target for designers and rehabilitation engineers. What is to be avoided in the process is hard focus on design of equipment. Rather, the focus must be on the user in a way that comprehends mobility in a total way and how existing or potential elements can be assembled on the basis of their functional features to deliver to the disabled what they need to free them from us as they free us from them in terms of dependency.

5. AKNOWLEDGEMENTS

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D.Cooper, design and clinical inputs.

POWERED MOBILITY FOR CHILDREN

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ABSTRACT

Traditionally, powered mobility has not been recommended for children. Rationale is presented to encourage a change in attitude based on experience gained in Project TEACH - a three year demonstration grant from B.E.H. Through movement children acquire perceptual skills. Greater choices are available both in terms of powered devices for children and innovative control systems. Experience has shown that children learn to operate and do not abuse powered mobility devices. Folding chairs allow for easier transportation. Finally, powered mobility allows children freedom to explore, to misbehave and to initiate activities.

Ongoing research projects are proposed which will assist making powered technical aids a viable solution to the mobility problems of severely handicapped children.

POWERED MOBILITY FOR CHILDREN

Traditionally, the time to consider powered mobility for persons with physical handicaps was as they entered adulthood.

There were many reasons given to support this time honored stand. Children, they said, did not need a powered wheelchair. They were too expensive and children were not sufficiently mature to duly respect this high cost technical aid. It was felt that the walls of homes and schools were in jeopardy as children would have difficulty learning to direct battery powered devices.

Families were told that transportation of powered chairs was difficult and purchase should only be considered if the family owned a van. Therapists were adamant that children should never be given a powered wheelchair if they had any potential at all to propel a manual chair. The rationale was that the child should be using his own motor abilities so they would remain functional. The only known exception was in the young adults with advanced muscular dystrophy. Finally, it was felt that any child with any degree of mental retardation would never be capable of learning to operate and care for an expensive and complex technical aid.

Rationale for Change

In many parts of the country, therapists, parents, and engineers are now actively persuing funds to provide powered mobility for even young children. They are being met with resistance from third party payers and at times physicians, administrators and collegues. None the less, more children are receiving powered mobility and many of the traditional arguments are no longer valid. In fact, we are observing very positive benefits to children that we had not anticipated. The options of powered mobility available for children have increased. This article will be based on experience gained from the following four devices.

1.) Child-Size Amigo
2.) C.A.P.P. Cart (Developed by Child Amputee Prosthetic Project at UCLA Medical Center)
3.) A BEC - 15
4.) E. & J. - 3P

The children ranged from four years to eleven years of age. All had Cerebral Palsy with motor involvement from the severe to moderate range and with functional intellectual level from trainable to normal. All children were part of Project TEACH*,a three year demonstration grant from the Bureau of Education for the Handicapped.

The premise of Project TEACH, is that through the provision of appropriate technical aids, severely handicapped children can more effectively participate in their educational process. Powered is only one category of aids for which the ten Project children are evaluated. By the end of the third year, a minimum of five children will have powered mobility assigned to them. At the moment, three additional children are being evaluated for and trained in the use of a powered base. Data is being collected and the decision as to whether the children benefit sufficiently to retain the devices will be made in the third year of the Project.

During the training and evaluation stage, many exciting benefits of electric mobility are being noted. Because of the availability of a

*Technical Educational Aids for Children with Handicaps

variety of powered devices, the children are being provided with one which reflects their specific needs. Severely physically handicapped children cannot move at a functional speed either in their manual wheelchair or on the floor. Over the years, this leads to concern in two areas. First some children become quite passive in that movement requires so much effort that the benefits do not outweigh their energy costs. Others become manipulative in order to control their environment but, this is extremely difficult for the severely handicapped child who is also non-verbal.

The other area for concern is the perceptual motor growth of the children. Severely handicapped children cannot move. Therefore, their concepts of body-in-space and spacial relationships are often delayed or distorted. How can a child know how far he is from the wall if he has never been able to travel the distance? How can he learn to direct a light on a scanning communication aid if he has never travelled independently through space? Perceptual development is essential to a child's educational advancement. It is felt that providing a child with the means to propel himself through space will greatly enhance his potential for perceptual growth.

Cost factors must be carefully considered. However, we must include the teacher/aide/parent costs which are required to take the child to the bathroom, to the cafeteria, to the therapy room, to the bus, etc.

Powered wheelchairs have an average life span of about five years. If our children were able to walk with bilateral long leg braces and a pelvic band using a walker, the cost over a five year period, (assuming changes for growth) would cost approximately $4,000 which would include the appliances and training. This does not include any surgical intervention. In comparison, costs of a $2,000 powered wheelchair are not quite so overwhelming.

Transportation of powered bases although still not considered easy, is now more feasible. A variety of folding chairs are now available. As well some of the other bases (Amigo, C.A.P.P.) can be disassembled with relative ease and transported in a car. Finally, most children go to school on buses, most of which have lifts and wheelchair space. This allows at least week day use of the wheelchair at home and school.

Children by nature are inquisitive. They want to explore, misbehave and run away when angry. How can they be children if they sit immobile in a wheelchair they cannot propel? How can we teach them a sense of responsibility when they must always depend on others for their daily tasks.

With advanced technology even many of the most severely physically involved can potentially drive a powered chair. Innovative control systems are necessary and not all bases are appropriate. However, it is the severely physically involved child who most needs the independence of mobility.

Conclusion

Powered mobility should not be withheld from children. By waiting until they are adults, many valuable years of learning are lost. Independence and responsibility must be allowed or taught from an early age. The children learn not to hit walls. They learn to remember to ask someone to charge the batteries. They learn to go in straight lines and how to get out of tight corners. They learn how to be responsible for their own timetable at school. Given mobility, they experience movement and control. They can explore and they can run away. They can be children!

-Device-Amigo-Child Size
-Child - F.N.

F. is a ten year old boy with a diagnosis of spastic cerebral palsy. He can ambulate for short distances using a modified rolator walker and an adduction brace. Energy costs are high and the pattern of gait, though functional, is poor.

Rationale for Choice

F. required a mobility device for long distance travel. He needed to be able to get on and off himself and to transport his walker and books with him.

The Amigo was chosen and modified for his use with the addition of a custom seat, a seat lock which he can operate with one hand, a book receptacle and hooks for his walker.

Results

Since receiving his device, he attends math classes in the adjoining elementary school which is the first step in his mainstreaming process. He continues to walk within his classroom, during therapy periods and at home. No skills have been lost over the six month period since he had his Amigo and many new opportunities have evolved for learning.

C.A.P.P. Cart
Children - A.E.
- S.R.

A. & S. both have diagnosis of Cerebral Palsy. A. was five years of age when he first drove the cart and S. was six. Both children were totally dependent on others for mobility. Both were considered severely physically involved.

Rationale for Choice

The C.A.P.P. Cart was designed low to the ground for use with the Thalidomide children in the 1960s. The base was an ideal height and was modified to accept the MPI (Modular Plastic Insert designed at the REC in Memphis) interface. Each child has their own control tray which contains micro-switches activated by ping pong balls. S. controls are located in the midline while A.'s have been located to the left of the tray indicating his control range.

Results

A. has no fear and enjoys exploring his environment. He can direct the device at will if motivated. However, he also spends a great deal of his time purposely going in circles. It has been hypothesized that he is providing his own vestibular stimulation. Evaluation is ongoing. S. is still quite timid in the device. She has never moved independently and is just learning to enjoy the freedom. Training is continuing.

A BEC
Child - B.K.

Bobby is a thirteen year old boy with a diagnosis of Cerebral Palsy with an intellectual level of trainably retarded. He communicates using a 175 symbol Bliss board.

Rationale

Bobby required primarily indoor mobility. He could operate a joystick with no modifications necessary. A custom seat was provided. The family vehicle is a car and long term objectives did not call for independent community mobility.

Results

Of all the children, Bobby, the most severely retarded, has taken the greatest responsibility for his chair. He literally guards it with his life - perhaps because it has given his life a new dimension. During training, he checked each night to be sure it was being charged and requested immediate service if repairs were needed. After just six weeks, he is in the chair all day and has been allowed to take it home.

Child - D. D.

D. is a twelve year old boy with a diagnosis of severe Cerebral Palsy-Athetoid. He has had his powered chair for two years.

Rationale

The BEC chair was chosen for two reasons. First he required the chair only for indoor use, mainly at school. Second, because of the severity of his physical limitations, the BEC was felt to have the most practical control system to modify. He operates the chair by sliding a hand curser holding a magnet bar over reed switches which are buried under his tray. Being an athetoid, he has his best control in extreme outer range.

Results

D. is completely independent around the school. Arrangements are being made to allow him to take the chair home on weekends and during vacations as a general phase-in begins. The home is small and a ramp entrance is required. D. is nonverbal and investigations for a communication aid require that both aids be operable by the same control mechanism.

Device - E. & J. 24V
Child - W.M.

W. is a nine year old child with the diagnosis of Cerebral Palsy, spastic quadreplegia. He is severely involved and nonverbal.

Rationale for Choice

W. is a child of normal intelligence. Long term goals include independent community mobility and potentially higher education.

Results

W. is totally independent at home and school. He visits nearby friends with his brother. The family keeps a manual chair avialable for outings but W. is in his powered chair at all other times.

W. has a mind of his own. When told to work on a project he didn't like, W. ran away into the adjoining elementary school. He returned at lunch time.

Future Considerations

During the final year of Project TEACH-data will be accumulated to substantiate the claim that powered mobility enables children to more fully participate in their education program. Already we are noting areas where future design and development must occur. A larger choice of reliable folding chairs should become available. Control systems must be developed so that even the severely involved can operate them safely. Variable speed controls, built in safety stops and controls which ignore random motions must be refined. If a child requires mobility as well as a communication aid, one control system should suffice-for simplicity, but also because often only one motor control site is available. This same control should offer access to a school station housing a computerized learning module, a page turner, a calculator or a typewriter. School buses must become equipped with safe standardized tie down systems so that a variety of wheeled bases can be accommodated. More options for powered mobility must be developed for the very young. They need not look like chairs, but they should provide the severely handicapped child greater learning experiences. Work in this regard is already underway at the Rehabilitation Engineering Center at Stanford.

Finally, instrumentation must be developed so that we can accurately measure a child's motor control. In this way, controlsystems of mobility devices will be designed within the child's actual abilities.

Children are not just little adults. Scaled down adult models of powered devices sometimes work. More often devices must be designed specifically for children so they can fully participate in their educational programs and social experiences.

INDOOR WHEELCHAIR

Per Finden, Øivind Lorentsen

Central Institute for Industrial Research

A new indoor wheelchair which requires little space to manage has been devised.

The technical solution provides sideways as well as back- and forward driving. Five units have been produced and have been tested by handicapped people. The results are very satisfying.

A special sitting unit has also been developed with different kinds of functional regulations built in.

INTRODUCTION

Conventional wheelchairs require a great deal of space to be maneuvered, and this makes the use of wheelchairs energy consuming and cumbersome, especially in narrow surroundings.

A project was started for developing a new type of indoor wheelchair. The most important functional specifications for the new product were:

1. Drive sideways as well as forward/backward.
2. Turn around the central axis.
3. Modular construction.
4. Functional and esthetical adaptable.

The project received economic support from the Royal Norwegian Council for Scientific and Industrial Research.

A directing committee was established, consisting of a specialist in physical medicine, a specialist in biomechanics, a representative of the handicapped and a representative of a manufacturing company.

The project started in 1976 and will finish 1980.

TECHNICAL SOLUTION

The construction work was divided in two sub-projects:

I) Frame-module and electric propulsion systems.
   a) Frame construction.
   b) Electronic control-system.

II) Chair-module and regulation systems.
   a) Elevation module.
   b) Chair-frame module.
   c) Cushion module.
   d) Footrest module.

I) Frame-module and Electric Propulsion System

There are many technical solutions that may satisfy the two first specifications mentioned in the introduction, but most of them are complicated and expensive to produce. Among all the solutions we sketched, one in particular was found to be the most simple.

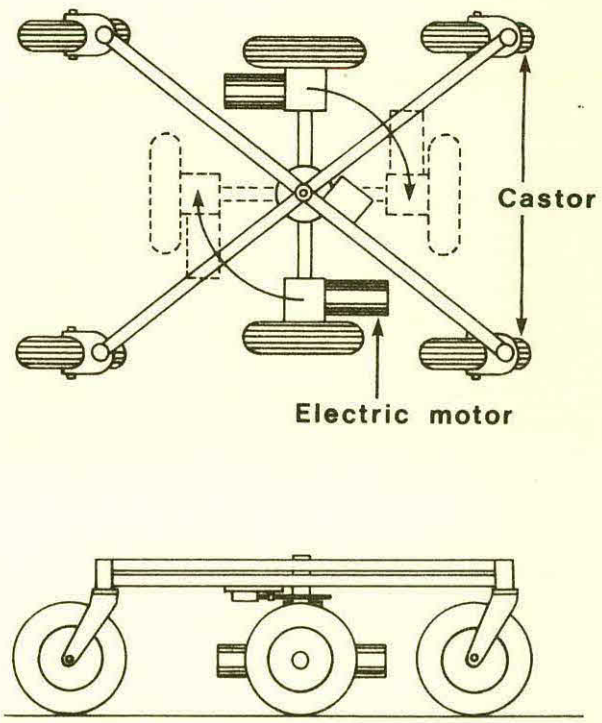

Fig. 1

Six wheels are mounted on the main lower frame; one wheel in each corner and one in each end of a lever that can rotate around the central axis. The lever could be rotated to any position and fixed relative to the main frame. In practice, it is only necessary to fix the lever in two positions, one for driving in the forward/backward mode, and one for the sideways mode. The two positions are 90^{o} relative to each other, and can be seen in Fig. 1.

The wheels on the lever are driven by two independent electric motors. The other four wheels are rotated freely. The lever is fixed in the two mentioned positions by using a specially developed electric-mechanical locking device. The lever is positioned by driving the motors in opposite directions.

Having the lever locked in one of the two positions the wheelchair can be maneuvered by controlling indpendently the speeds of the two motors. Because of the six-wheel construction the suspension of the wheels are critical in order to pass inclinations and doorsteps as smooth as possible.

The electric power system operates by a digital servounit, and the disabled can drive the vehicle by an ordinary joy-stick.

II) Chair-module and regulation system

Upon the lower frame unit we may place mostly any chair that will suit the user. Many of the ordinary chairs on the market we may adjust to the frame. Disabled persons may in this way get an acceptable wheelchair. However, for people with more serious handicap it was found necessary to develop a more specialized unit.

The elevation unit can elevate the seat level from about 450 mm to 750 mm above the floor. The technical solution is based on a very specially forced bars mechanism operated by a small electric motor.

The chair frame has built in two particular functions:

- The back-rest can turn from vertical to horizontal position in such a way that it follows the back of the user. The relative rotation has its centre about the user's hip joint.
- The foot-rest can, co-ordinated with the back-rest, turn from vertical to horizontal level. The mechanical fix-point lies beneath seat-level, but by a chain system it has a rotation point aligned with the user's knee point.

All the movements are operated by a small electric motor.

The seat and back-rest are covered with a cushion system. A complete unit consists of 24 separate cushions, which are made by foam-plastic in three different heights (see pictures). Now we may create many types of seat-profiles by combining different cushions and fix them to the chair frame. The securing is done by metal rails fixed to the frame.

CONCLUTION

Until now five prototypes are produced. They have been tested by handicapped people for about one year, and the results are very promising.

The technical solution of the lower frame module has been patented.

A Norwegian company is going to produce the wheelchair, and the first units will be ready in a few months.

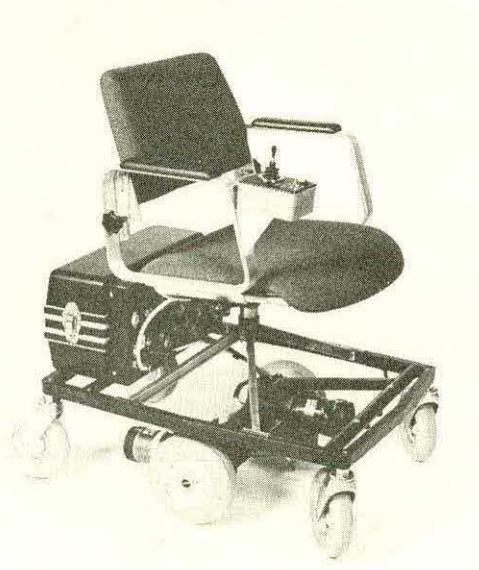

Fig. 2

The frame-module with a standard office-chair as sitting unit.

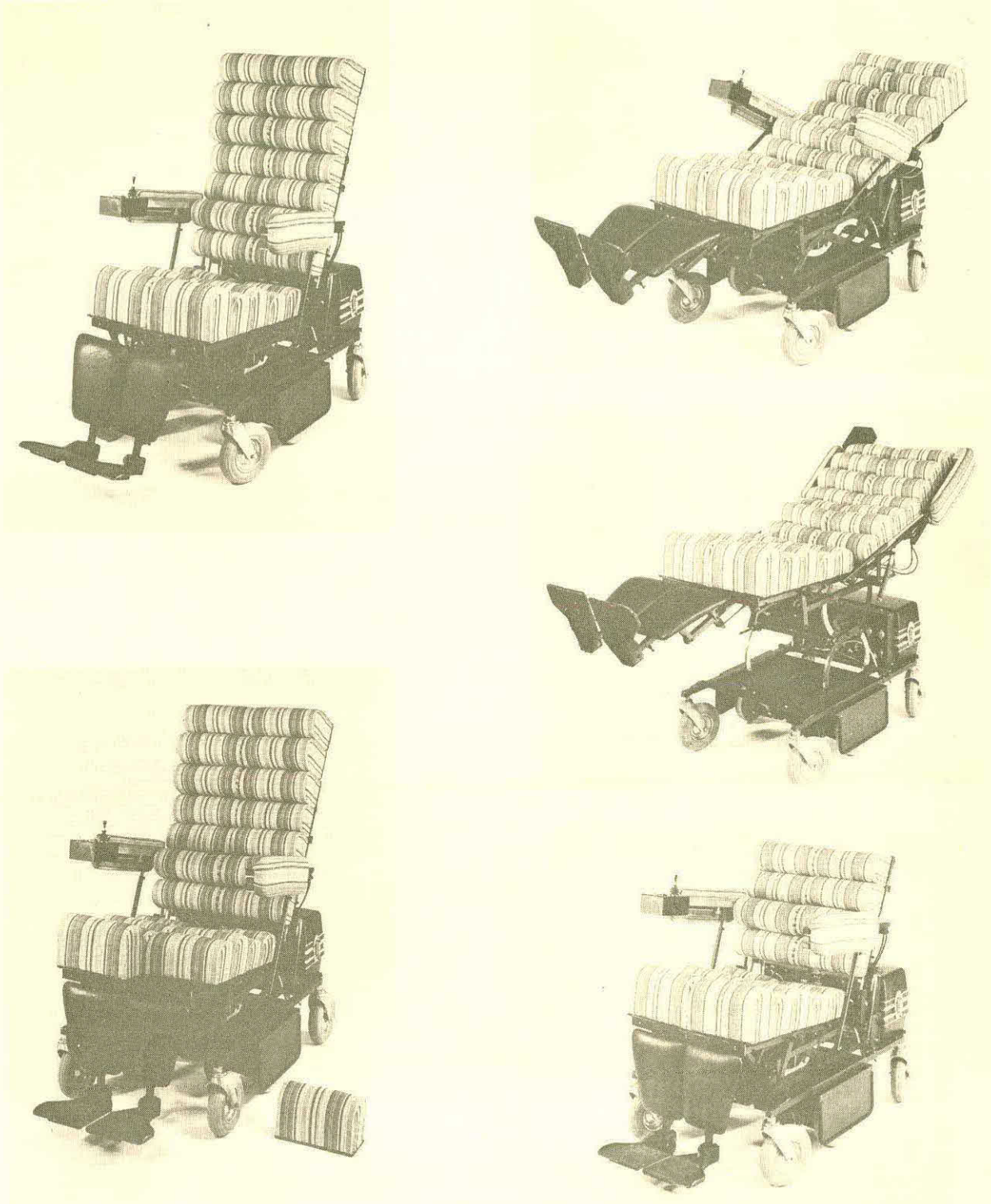

Fig. 2

The frame-module with the more specialized sitting-unit in different positons.

A MULTI-INTERFACE WHEELCHAIR CONTROLLER

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ABSTRACT

This paper presents developments relating to a system for the control of an electrically driven wheelchair by athetoid cerebral palsied children. A major emphasis is placed on the interface between the child and the control system. The system is adaptable to a wide range of interfaces and wheelchairs. It incorporates a ramp controller for linear acceleration, a separate joystick control for parent or attendant use, and the option to operate an external device. The control base is detachable for wheelchair transportation.

INTRODUCTION

Commercially available powered wheelchairs are usually joystick-controlled and hence do not provide a simple means of interfacing the severely involved athetoid cerebral palsied child. In addition, the young first-time user of a powered wheelchair has the problem of gaining sufficient confidence and experience to effectively manipulate the device because of the high starting torque of the electric drive motors. Simplification of the control of these motors is necessary to permit emphasis on the child-controller interface rather than on the wheelchair drive system.

This paper describes a system designed to enable a severely handicapped child with athetoid cerebral palsy to drive a powered wheelchair while also providing him with the option to operate a Blissymbolics communication board using head position control. The control system is designed to be adaptable to a wide range of commercially available powered wheelchairs.

THE INTERFACE

A biofeedback training system has been developed at the Ontario Crippled Children's Centre (OCCC) for providing the cerebral palsied child with auditory and visual information regarding the spatial position of his head(1). The development of adequate head control is seen as a necessary antecedent to further gains in motor and perceptual development(2). As a result of the biofeedback programme at the OCCC, it became possible for a number of severely disabled individuals to use head position to control powered wheelchairs. The device consists of mercury switches with appropriate circuitry mounted on the head by means of an adjustable head-band. The debounced position-dependent switches with logic-level outputs encode 5 functions. These are: right-turn (head tilted to the right); left turn (head tilted to the left); forward (head held upright); reverse (head tilted back); and stop (head tilted forward).

In 1979, a powered wheelchair with head position control was prescribed for A.M., a 12 year old severely handicapped male with athetoid cerebral palsy who participated in a head-position training programme during 1977 and 1978 at the OCCC. He was unable to control the wheelchair because of a panic response induced by the initial jerk of the chair following from the high starting torque of the electric drive motors. A complete redesign of the control system was undertaken. The redesigned control system had to incorporate means for A.M. to use his head controller to operate a Blissymbolics communication board(3), and to permit his parents or attendents to operate the wheelchair from behind by means of a conventional joystick whilst over-riding his controls.

THE CONTROL BOX

A block diagram of the control system is shown in figure 1.

The control base provides the following functions:

(1) A manual control to provide an external means of operating the wheelchair independent of the user. It consists of a regular 4-switch joystick control selected by a manual switch.

(ii) Motor control. This is a ramp controller for linear acceleration from zero to a preset maximum speed. It consists of a ramp generator whose output is compared to a voltage on a capacitor charge circuit. The comparator produces a pulsed waveform used to drive the motor through a set of Darlington transistors. The motor control board also decodes logic ciruits to enable the Darlington drivers and the motor reversing relays thus controlling the direction of the drivers to the motors.

(iii) Manual speed control. This provides a means of setting the maximum speed of the motor control by adjusting a potentiometer which

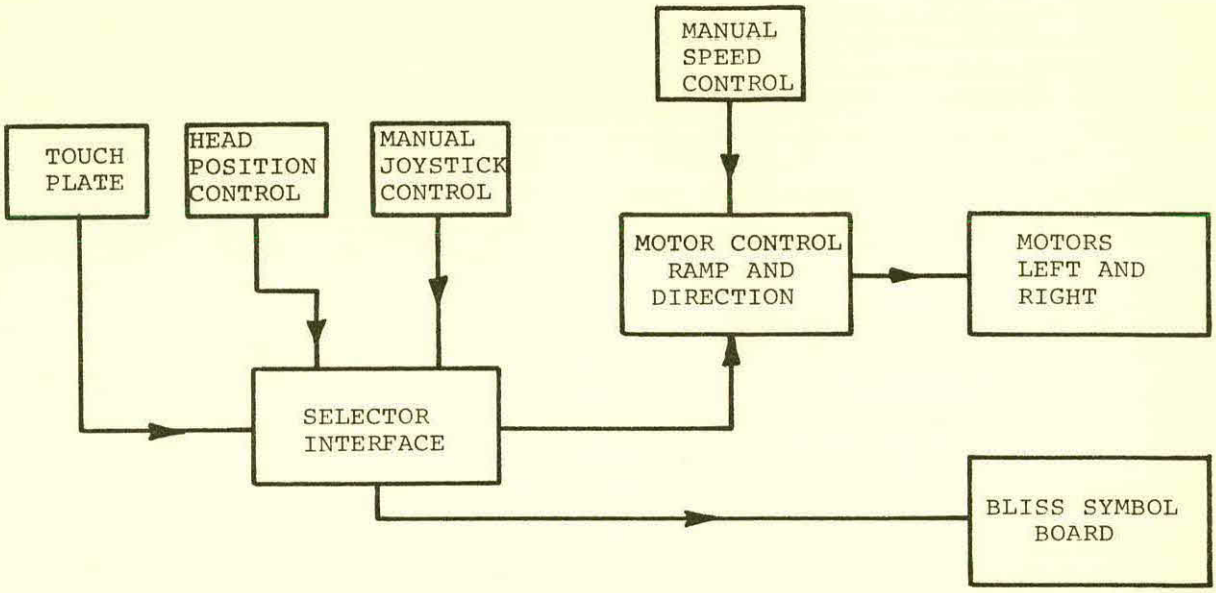

FIGURE 1 BLOCK DIAGRAM OF CONTROL SYSTEM

determines the comparator level in the motor control board.

(iv) Selector interface. This circuit directs the control signal to either the motor control or an external device. It also decodes the selection of the manual or wheelchair-user control.

The circuitry is modular in form, each section of the controller system being seperate. The function of the interface is selected by changing only the decoding of the selector interface board.

In addition to the head control unit and the communication board which was mounted on the wheelchair, a touch plate switch was provided. It consists of a capacitive-enabled Schmitt trigger latched by a toggle flip-flop. The touch plate switch is used to enable the communication board and also serves as an emergency stop for the wheelchair. The selector circuit thus directs output signals from the head position control-unit to the motors or to the Blissymbolics board, or from the manual joystick to the motors.

The manual speed control enables a maximum speed to be selected such that the user feels confident to drive under head position control. As the level of skill increases, the maximum rate can be increased. The speed control also enables the parents to increase the speed of the wheelchair when the manual joystick is used.

CONCLUSIONS

A wheelchair controller has been designed which permits control of a powered wheelchair or an external device. The control system is sufficiently simple and flexible to be used with a variety of interfaces and is adaptable to commercially available chairs. The control system is modular in form thus facilitating ease of maintenance. The control box is easily detached, the characteristics of the wheelchair to which it is fitted being virtually undisturbed e.g. a folding type of wheelchair can still be folded. Detailed circuit diagrams are available(4).

ACKNOWLEDGEMENTS

The authors would like to acknowledge the efforts of Sam Chung, Scott Walder, and Damien Wiechula in developing the touch-plate system and the linear accelerator.

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BRAKING SYSTEMS FOR POWERED WHEELCHAIRS FOR THE SEVERELY DISABLED

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Powered wheelchairs for use by persons with quadriplegia and other severe neuromuscular disorders must have a reliable braking system which is easily operated, and whose operation is readily understood by the driver. The system should utilize standard parts to facilitate repair or replacement when necessary and should be adaptable to any style of powered wheelchair. We have developed mechanical and hydraulic brake systems for powered wheelchairs which utilize standard brake components and can be operated by mechanical switch or from a pneumatic ("puff" and "sip") transducer mounted on the wheelchair. The brakes are applied as required by the driver in the same manner as he would brake if he were driving an automobile. Our electric motor driven hydraulic brake system provides proportional braking as well as a convenient parking/transfer brake and does not require a separate manual disconnect for manual pushing of the chair.

INTRODUCTION

Powered wheelchairs for use by persons with quadriplegia and other severe neuromuscular disorders must have a reliable braking system which is easily operated, and whose operation is readily understood by the driver. The system should utilize standard parts to facilitate repair or replacement when necessary and should be adaptable to any style of powered wheelchair. The need for a suitable braking system has become even more compelling because of the great increase in the number of spinal cord injuries in the U.S. Spinal cord injury affects approximately 175,000 in the U.S. About 6,500 new injuries occur yearly. Most patients are under 30 years of age and 60% are quadriplegic (1).

A multidisciplinary team consisting of physiatrists, occupational therapists, rehabilitation engineer, driver training instructor, behavioral scientist, and disabled consumers at IRM has concluded that the optimum braking system is one which operates in a manner analogous to the braking system in an automobile, especially since almost all of our patients were drivers prior to their disabling accident and are thoroughly familiar with the operation of such a system. Thus we would like our braking system to perform as follows:

1) Brakes are applied as required by the driver in the same manner as he would brake if he were driving an automobile.

2) Brakes may function as auxiliary parking/transfer brakes provided they can be released without the need for a separate manual disconnect.

3) Brakes operate independently of the wheelchair drive motors.

4) Braking action should be proportional.

5) Brakes are not interconnected to the drive controller and are controlled from a separate switch (or transducer).

Our system differs from what Stout (2) considers to be a "good braking system". Stout specifies that "The switch which sets the brakes also interrupts power to the motors". Such a switching arrangement would not enable a wheelchair to stop on upward sloping terrain and then resume upward travel. For example: the brakes must be applied to hold the chair on the slope. It would then not be possible to apply power to the motors to enable the chair to continue its travel without first releasing the brakes. Our experience with an E & J Model 3P chair on a 10-12° sloping surface which also slopes slightly sideways (see Figure 1) has been

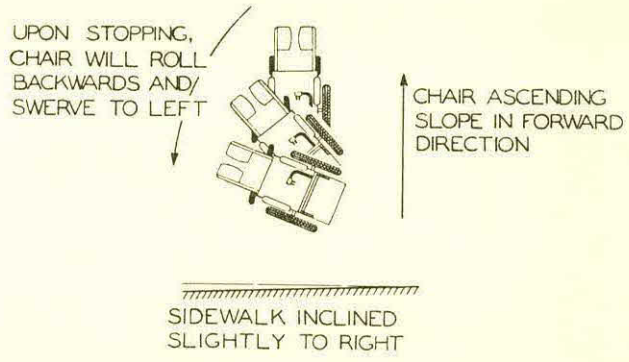

Figure 1: Backward and swerving motion of a powered wheelchair when stopped on a sideways tilted inclined terrain.

that when the brakes are released the chair will roll backwards and will also spin or swerve to one side. This action can easily be avoided when forward power is applied prior to releasing the brakes. For this reason we do not automatically set the brakes when the motors are not energized. Strout also specifies that "Application of brakes occurs upon interruption of power to brakes". This arrangement requires the addition of still another system: "...a manual disconnect to allow manual pushing of the chair". There are reports (3) in which disabled users of powered wheelchairs (in which brakes set upon interruption of power to the brakes) were seriously injured when the chair lost power, stalled in a dangerous location, and could not be moved because passersby could not operate the manual disconnect.

It has been our observation that the usual reasons for loss of power is that batteries have not been adequately recharged. To warn the driver of impending failure there is now a battery voltage indicator light on the control box of most powered chairs which begins to flash when the battery voltage drops below the preset level. This is an indication that the batteries should be charged as soon as possible.

METHODOLOGY

We have designed and fabricated three prototype braking systems:

Mechanical Brake System

This system utilizes drum type friction brakes in which the friction brake bands are connected to a pair of actuating solenoids through flexible cables. Two separate drums are required - one for each of the rear drive wheels. The solenoids utilize an efficient two coil design: one coil to develop the required high initial pull, the other to hold the load in position. After the solenoid plunger moves through the required stroke distance, it actuates a switch which cuts the high power coil out of the circuit leaving a separate low current drain coil to hold the load in position. Details of this system are shown in Figures 2, 3 and 4.

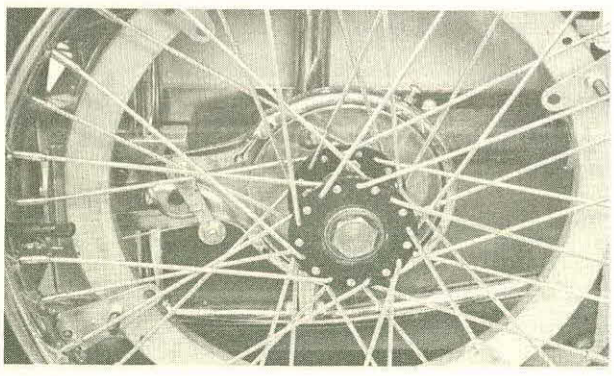

Figure 2: View through spokes of drive wheel of powered wheelchair showing mechanical brake system.

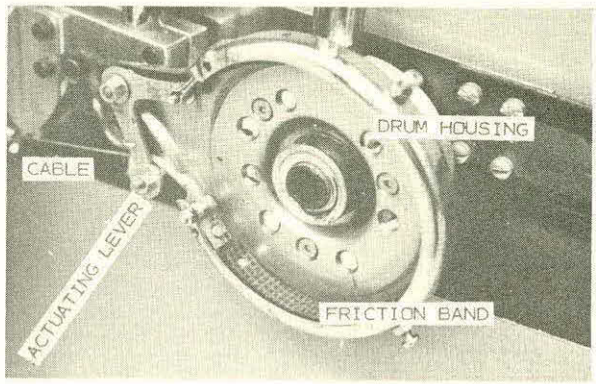

Figure 3: Disassembled mechanical brake showing the friction band, drum housing, actuating lever and cable.

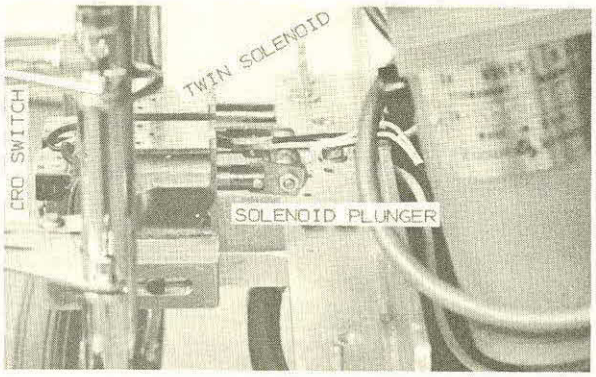

Figure 4: Twin solenoid assembly showing position of microswitch and solenoid plunger. After brakes are applied, high power coil is disconnected and low current drain coil is connected to hold brakes in position.

Hydraulic Brake System

This system utilizes caliper disc brakes which are actuated hydraulically from a master cylinder. The master cylinder force is produced by the solenoids previously described. Two separate sets of brakes are required as for the drum brakes. A schematic drawing of the system is shown in Figure 5. Figure 6 shows the actual assembly. Figures 7 and 8 are front and rear views of an E & J 3P powered wheelchair equipped with the IRM hydraulic brake system.

Motor Driven Hydraulic Brake System

This system utilizes the caliper disc brakes and hydraulic system described previously. However, the solenoids have been eliminated and a small geared electric motor is used to generate the input force to the master cylinder. The electric motor drive provides:

1. Proportional braking action.

2. A transfer and emergency brake re-

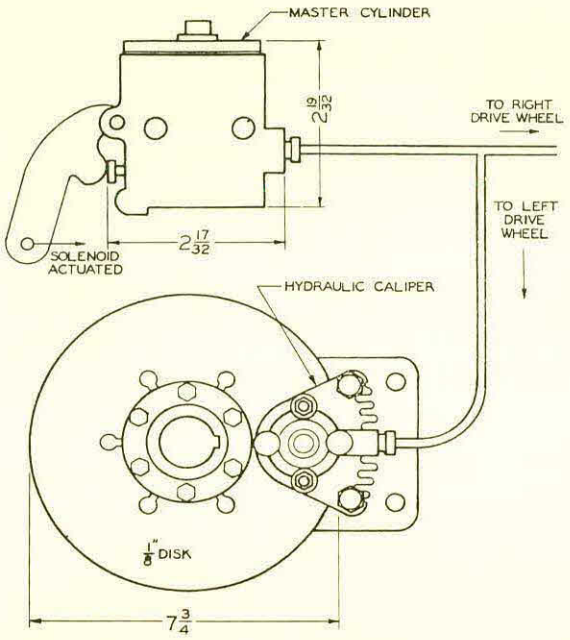

Figure 5: Schematic of hydraulic brake system.

quiring no drain from the battery.

3. A convenient means of releasing the brakes to allow manual pushing of the chair.

EVALUATION

All three systems have been evaluated utilizing an E & J Model 3P powered wheelchair carrying a 175 pound driver. The braking systems can be operated from a mechanically actuated switch or from a pneumatic ("puff" and "sip") transducer. The mechanical brake system is the least expensive of the three. However, we found that braking in the forward direction was more effective than in the backward direction. This is a characteristic of the particular drum-band configuration being utilized. The hydraulic brake system performed without flaw on upward and downward terrain with slope of 15° - 2.7 feet rise (or fall) per 10 feet of horizontal travel. When travelling down slope at 2-3 miles/hour, it was possible to come to a smooth stop within 1-2 feet of travel after engaging the brakes. No backward slippage was experienced when stopping on the upward sloping grade. A disadvantage of both of these systems is that a separate parking (or transfer) brake is required. Truly proportional braking is provided by the motor-cam driven hydraulic brake system. This system permits the driver to modulate the brakes. In the pneumatically actuated version, the brake motor operates only when pressure is maintained in the pneumatic tube. This system also provides parking brake action because the brakes will remain in the set position after actuation. In order to release the brakes the braking motor must be reversed.

It is hoped that continued development of these systems will result in a packaged set of brakes that can easily be adapted to powered wheelchairs for the severely disabled. Technical details of any of the described systems will be furnished upon request.

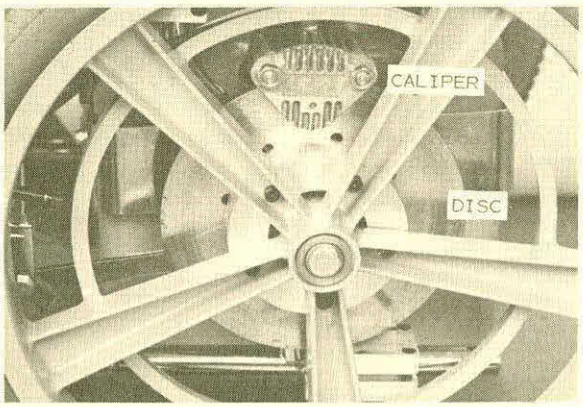

Figure 6: View through spokes of drive wheel of powered wheelchair showing hydraulic brake assembly.

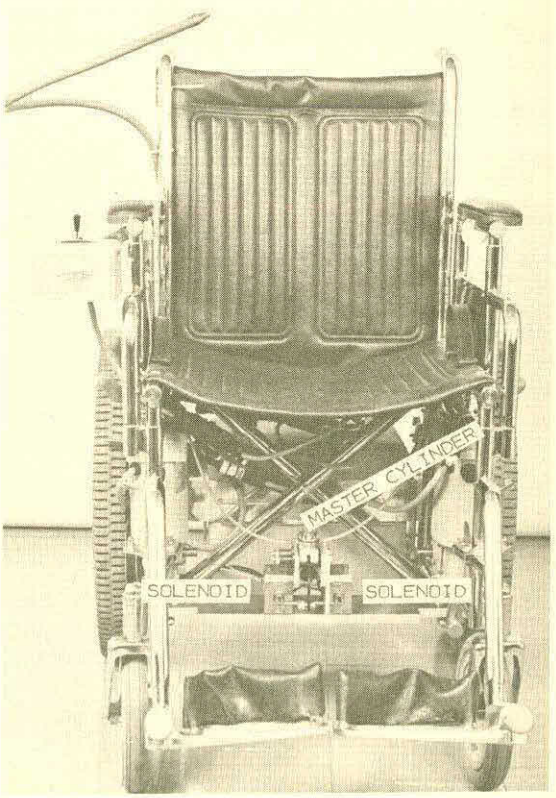

Figure 7: E & J Model 3P powered wheelchair equipped with IRM hydraulic brake system. Brakes can be operated from a mechanical switch or pneumatically ("puff" and "sip").

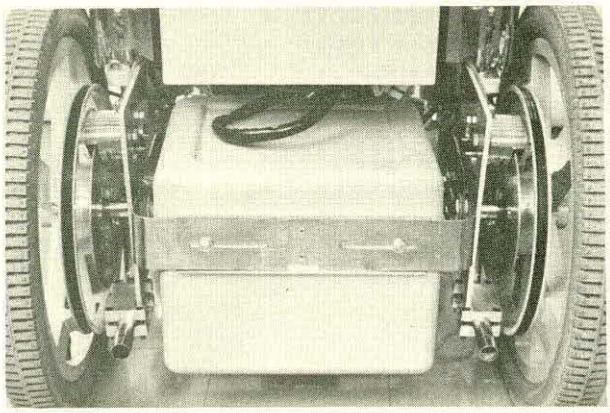

Figure 8: Rear view of powered wheelchair equipped with hydraulic brakes showing method of mounting.

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POWERED WHEELCHAIR SIDEWAYS TIP-OVER AND CURB CLIMBING

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ABSTRACT

The method of testing and results for the determination of sideways tip-over and curb height limitations for five powered wheelchairs are presented. A method for calculating the center of gravity of a seated wheelchair user is also described.

INTRODUCTION

The purpose of the powered wheelchair testing program at the University of Virginia Rehabilitation Engineering Center is to evaluate the operating characteristics of commercially available powered wheelchairs. This program will eventually involve tests concerning all aspects of chair operation from battery life to tip-over angles. Two tests, sideways tip-over and curb climbing, are reported in this paper plus an analysis for a center of gravity calculation technique for a seated wheelchair subject. This report includes tests on powered wheelchairs currently on hand at the Rehabilitation Engineering Center. Additional powered wheelchairs will be tested when they become available.

SIDEWAYS TIP-OVER

Several powered wheelchairs were tipped to the side, first unoccupied and then with seated subjects, and the sideways angle of tip-over was measured to within one degree with a protractor attached to the chair. The casters were allowed to turn freely. The subject was not allowed to correct for the tip, but allowed his/her body to be tipped to the side with the chair, as if strapped into the chair.

The tip-over angle measured at the unstable equilibrium position is that angle at which the center of gravity of the chair, or chair and subject, passes vertically over the point of pivot on the ground. Therefore, the higher the combined center of gravity of the chair and subject, the smaller the angle causing tip-over will be. The centers of gravity of two subjects of significantly different body size, one 5'2" 110 lb female and one 6'4" 180 lb male, were measured by the method described in the Appendix. For the male the coordinates for the center of gravity, $\bar{X}$ and $\bar{Y}$, were 2.8" and 11.3" respectively, and for the female, 3.2" and 8.9". Zero is at the intersection of the center of the thigh and trunk.

The results of these measurements are shown in Table I. There is a significant decrease in the angle required for tip-over when a subject is seated in the chair than when the chair is empty because of the rise in combined center of gravity. However, the difference in sideways tip-over angle caused by extreme human torso variations is not appreciable.

CURB HEIGHT LIMITATIONS

In order to determine the maximum curb height that a wheelchair is able to negotiate, a 4' wide adjustable curb was constructed from wood. The initial height was one-half inch and it was raised by half-inches until it reached three and a half inches. The curb was blocked so it would not move when in use.

Five powered indoor/outdoor wheelchairs were run up onto the curb, forward and backward, low and high speed, from both a running and a standing start. In this way, the heighest possible curb each wheelchair was able to climb was determined. The wheelchair batteries were fully charged and the tires inflated to their specified pressure. For a standing start the wheels touched the curb. For a running start the wheelchair was started several feet from the curb and was at full speed when hitting the curb. In all the tests the user was a 5'2" 110 lb female. The results of these tests are shown in Table II.

TABLE I

Sideways Tip-over

Chair	Chair Unoccupied	Chair with 5'2" 110 lb female	Chair with 6'2" 180 lb male
General Teleoperators Inc. Model MHWC1	43°	28°	25°
Bec Model No. 12 (Front Wheel Drive)	58°	collapses before tipping	collapses before tipping
Everest & Jennings Model 3P	40°	30°	23°
Bec Model No. 14 (Rear Wheel Drive)	47°	30°	collapses before tipping
Everest & Jennings Model 3M	42°	25°	21°
National	52°	31°	28°

TABLE II

Curb Height Limitations

Chair	Standing Start				Running Start			
	Low speed		High speed		Low speed		High speed	
	For.	Back.	For.	Back.	For.	Back.	For.	Back.
General Teleoperators Inc. Model MHWC1	1"	2"	1"	>3.5"[a]	1"	3"	1.5"	>3.5"[a]
Everest & Jennings Model 3P	1"	1.5"	1"	2"	1.5"	2.5"	1.5"	>3.5"[a]
Everest & Jennings Model 3M	.5"	.5"	b	b	1.5"	1.5"	b	b
Bec Model No. 14 (Rear Wheel Drive)	.5"	.5"	1"	c	1"	1.5"	1"	c
National	1"	1"	1.5"	2"	1.5"	3"[d]	2.5"	3"[d]

a 3.5" was the heighest curb tested.
b the Everest & Jennings Model 3M has only one speed.
c the Bec Model No.14 only reverses in low speed.
d the battery pack prevents the National Chair from climbing a higher curb than 3" in reverse.

APPENDIX: Determination of the Center of Gravity of a Seated Human Body

The composite body method of center of gravity determination is used based on a volume analysis. Following the method of Contini [1], the total volume of the human body is calculated by dividing the body into six segments, four of these having symmetric duplicates. The segments and volumes are given in Table III and shown in Figure 1. The location of the centers of volume of the segments 3,4,5, and 6 as a percentage of the length of each segment, measured down from the top point of the segments, was taken from Contini's paper [1]. The dimensions from which the percentage distance to the center of volume of the combined segments 1 and 2 were taken from the book by Williams and Lissner [2]. These distances are also given in Table III and illustrated in Figure 1.

Assuming the density of each body segment to be the same, the center of gravity is coincident with the center of volume, and the total center of volume/gravity for the body is found using the equation:

$$C_v = \frac{\sum V_i d_i}{\sum V_i} \qquad (1)$$

where V_i is the volume of the i^{th} segment and d_i is the distance from the top of the i^{th} segment to its center of volume/gravity. The coordinate system of Figure 2 has been placed such that the y-axis is coincident with the center line of the head, trunk, and upperarm of a seated subject, and the x-axis is coincident with the center line of the thigh. The z-axis, shown in Figure 3, is directed out of the page in Figure 2 at the intersection of the x and y axes. The distances from the coordinate axes to the center of gravity of each segment are:

$$\begin{aligned}
&\bar{x}_1=\bar{x}_2=\bar{x}_3=0\\
&\bar{x}_4=(.42)L_4\\
&\bar{x}_5=(.427)L_5\\
&\bar{x}_6=L_5\\
&\bar{y}_{1,2}=(1-.603)(L_1+L_2)\\
&\bar{y}_3=(L_2-L_3)+(1-.461)L_3\\
&\bar{y}_4=L_2-L_3\\
&\bar{y}_5=0\\
&\bar{y}_6=-(.467)L_6\\
&\bar{z}_1=\bar{z}_2=0\\
&\bar{z}_3=\bar{z}_4=-B\\
&\bar{z}_5=\bar{z}_6=-A\\
&\bar{z}_{3'}=\bar{z}_{4'}=B\\
&\bar{z}_{5'}=\bar{z}_{6'}=A
\end{aligned} \qquad (2)$$

where A is the distance from the xy-plane to the center line of the leg, and B is the distance from the xy-plane to the center line of the arm, along which the center of gravity lies. The values for $\bar{X}_{total}$, $\bar{Y}_{total}$, and $\bar{Z}_{total}$ are calculated from:

$$\bar{X}_{total}=\frac{\sum V_i x_i}{\sum V_i} \qquad (3)$$

$$\bar{Y}_{total}=\frac{\sum V_i y_i}{\sum V_i} \qquad (4)$$

$$\bar{Z}_{total}=\frac{\sum V_i z_i}{\sum V_i} \qquad (5)$$

For a complete body, with no segments missing, $V_3=V_{3'}$, $V_4=V_{4'}$, $V_5=V_{5'}$, and $V_6=V_{6'}$, and the value for $\bar{Z}_{total}$ is equal to zero. $\bar{Z}_{total}$ will have to be taken into account for a body with any segment missing.

These equations can easily be put into a computer program to aid calculation. The measurements for C_1 through C_{10} and L_1 through L_6 were taken on four subjects and $\bar{X}_{total}$ and $\bar{Y}_{total}$ for each were calculated. These values were also measured by tipping the subjects in a chair and calculating the location of their centers of gravity using the tip-over angles. The results are shown in Table IV.

The measured and calculated coordinates for the center of gravity of subject A in Table III are in good agreement. But, the coordinates for subjects B, C, and D differ by as much as 35%. The errors are mainly due to the assumption that the density of each body segment is the same. These results show that it is more accurate to measure the center of gravity by tipping than by using a center of volume approximation. If the subject is unable to tolerate the tipping measurement procedure, however, this body segment volume method may be a useful first approximation. By including the average density of body segments it is expected that the calculated value for center of gravity will be improved.

TABLE III

Segment	Volume	Location of Center of Volume as % of segment Length measured from Top Joint
1. Head and neck (above collar bone)	$V_1 = \frac{(C_1+C_2)^2}{4}(L_1\pi)$	$d_{1,2}=(.603)(L_1+L_2)$
2. Trunk (collar bone to crotch)	$V_2 = \frac{(C_3+C_4+C_5)^2}{9}(L_2\pi)$	
3. Upperarm (two) (shoulder to elbow)	$V_3=C_6{}^2L_3\pi$	$d_3=(.461)L_3$
4. Forearm and hand (two) (elbow to fingertips)	$V_4=C_7{}^2L_4\pi$	$d_4=(.42)L_4$
5. Thigh (two) (crotch to knee)	$V_5 = \frac{(C_8+C_9)^2}{4}(L_5\pi)$	$d_5=(.427)L_5$
6. Shank and foot (two) (knee to ankle)	$V_6=C_{10}{}^2L_6\pi$	$d_6=(.467)L_6$

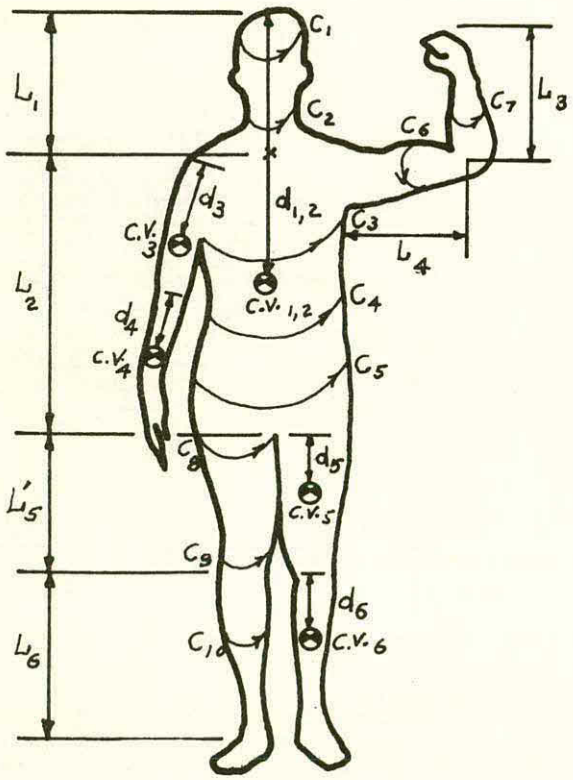

Figure 1

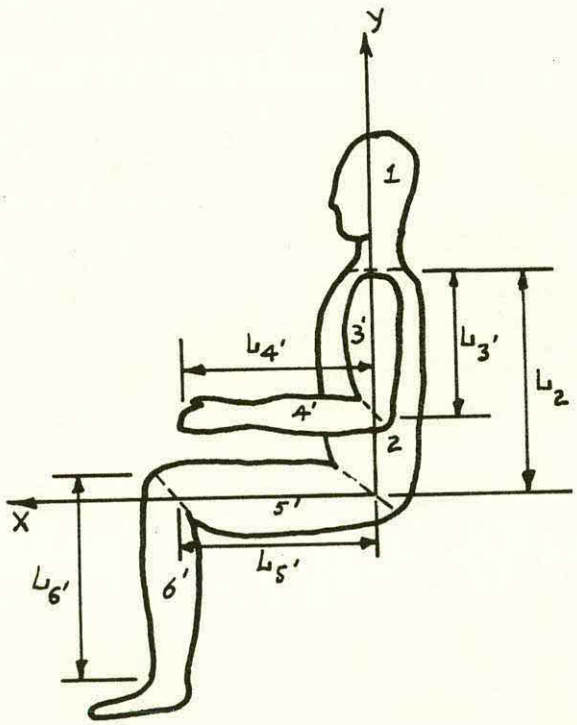

Figure 2

TABLE IV

Comparison of Center of Gravity Determination Methods:

Subject	Calculated $\bar{X}$	Measured $\bar{X}$	Calculated $\bar{Y}$	Measured $\bar{Y}$
A. 5'2" 110 lb female	3.22"	3.32"	8.85"	7.59"
B. 6'5" 185 lb male	3.41"	5.26"	10.46"	10.67"
C. 5'7" 130 lb female	2.97"	5.60"	9.30"	7.02"
D. 6'4" 180 lb male	2.79"	2.36"	11.31"	9.5"

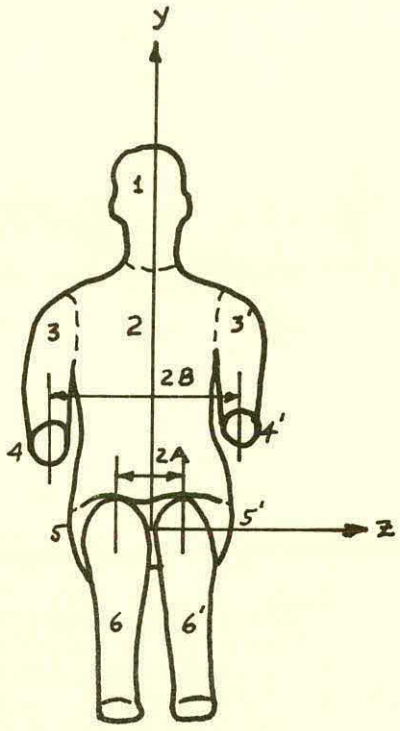

Figure 3

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(1) Contini, Renato. Body Segment Parameters (Pathological), 69,86, New York University Research Division, 1970.

(2) Williams, Marian, and Herbert Lissner. Biomechanics of Human Motion, 134, W. B. Saunders Company, 1962.

WHEELCHAIR BATTERY TESTING

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ABSTRACT

A method of testing has been developed for the evaluation of batteries with repect to wheelchair load conditions. This method allows for considerable flexibility in control over loading variables as well as monitoring techniques, representing a significant improvement over field testing in terms of its ability to apply consistant loading patterns. Preliminary test results have verified that capacities both lead/acid and nickel/cadmium batteries are relatively unaffected by high frequency pulse width modulation.

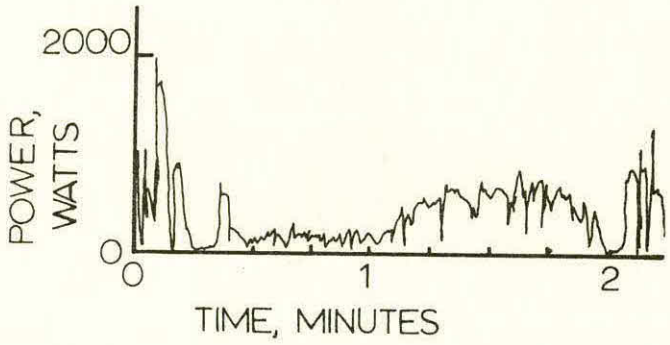
INTRODUCTION

The objective of this project is to evaluate recent technology, high energy density, storage batteries for poslible powered wheelchair application. The useful energy capacity of these new batteries exhibits signs of adverse effects when subjected to highly repetitious, pulsed loading cycles. Tests on Ni/Zn batteries conducted by the NASA/Lewis Research Center showed considerable loss of energy capacity when subjected to a simulated postal vehicle driving cycle (1 sec at 30A, 4 sec at 200A, 7 sec. rest)[1] Since wheelchair power requirements are very unsteady (See Fig. 1), the Ni/Zn combination may well exhibit this loss of capacity when subjected to wheelchair loading patterns. Testing of Ni/Zn batteries has been delayed by battery availability problems, but the project is ongoing in terms of test equipment development and testing of Pb/acid and Ni/Cd batteries.

TEST UNIT DESCRIPTION

A battery tester has been designed and constructed. It can best be described by separating the test unit into two systems: the loading system and the monitoring system. This section presents them in that order.

FIGURE 1
INSTRUMENTED WHEELCHAIR DATA



The loading system consists primarily of a set of transistor controlled resistor banks. The controls allow the operator to select either a direct current or a pulse width modulated power drain. Direct current loads are available at present values depending on which resistors are activated by the operator. Presently, power drains of 72, 144, 288, and 576 watts (at 24V) are available as well as any combination of the above. Pulse width modulated loading is infinitely adjustable from 0 to 100 percent duty cycle. Pulse frequency may be selected from the continuous range of 20 to 2,000 Hz. Peak current and power drain is adjustable, but a peak drain of 50 amps/1200 watts was used for these preliminary tests. Feedback systems are available to hold the desired variables independently constant. Ultimately, the load will be controlled by a microcomputer program based on electronically recorded data from an instrumented wheelchair.[2]

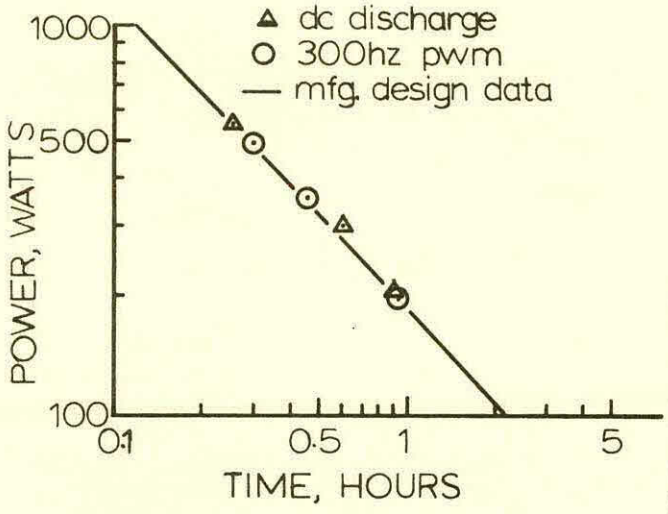
The monitoring system operates concurrently with the loading system to provide data on battery voltage, current, power, and cell temperature. Data is electronically averaged and reduced to direct current voltages. Instantaneous values for all four variables are available to the operator via L.E.D. panel meters. Optional oscilloscope monitoring may be easily used to observe the unaveraged as well as the averaged values.

Permanent real time records are strip chart recorded and also sampled at 1 Hz and stored on floppy disk by computer. Computer software has been developed to perform power curve integration, averaging, and other forms of data manipulation.

TEST RESULTS

The results of the initial tests on Pb/acid batteries (Fig. 2) have shown that their energy capacity is independent of pulsed loading. These tests also provided confidence in the testing method. In Figure 2 we see that the energy capacity for a pulse width modulated discharge is the same as D.C. discharge.

FIGURE 2
BATTERY CAPACITY
GLOBE / UNION U-128 GEL/CELL ⓣ

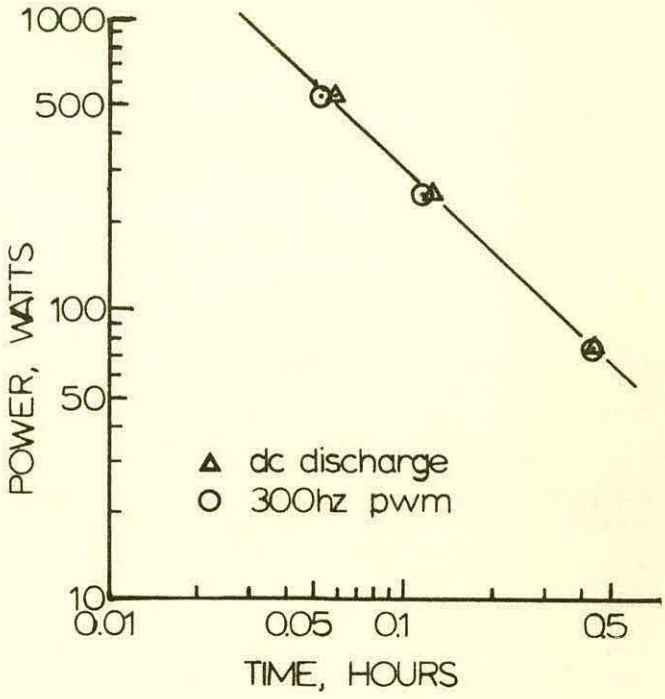


Initial tests have also been conducted on Ni/Cd batteries. Like the lead/acid batteries, preliminary test results (Fig. 3) indicate that the Ni/Cd combination is unnoticiably affected by high frequency, pulsed discharge. This does not necessarily imply that effects will not become apparent over greater numbers of charge-discharge cycles.

CONCLUSION

Tests of Pb/acid and Ni/Cd batteries show that battery capacity is not affected by the high frequency pulse width modulated circuitry usually found in wheelchair controls. In the future more varied types of loading will be applied and different battery types tested.

FIGURE 3
BATTERY CAPACITY
NICKEL / CADMIUM



1. "300 AH Nickel Zinc Cell", NASA/Lewis Research Center, Cleveland, Ohio, Public Relations Document Release, 1978.

2. 1979 Annual Report, Univ. of Va. Rehab. Eng. Center, Charlottesville, Va., pge 41.

EFFECT OF SEAT POSITION ON WHEELCHAIR PERFORMANCE

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ABSTRACT

Efficiency and stroke frequency and duration were determined for four subjects for simulated wheelchair propulsion on a dynamometer in nine seat positions. It was found that the conventional position and a position with the seat moved backward eight inches were the most favorable in terms of efficiency. Efficiency decreased considerably when the seat was moved further back an additional eight inches and also when it was raised and lowered two inches from the standard position.

INTRODUCTION

The universally accepted wheelchair with rear mounted main wheels is not compatible with horizontal changes of seat position without loss of stability. Comparisons have been made between wheelchairs with front mounted main wheels and the now conventional rear-wheeled chair in terms of efficiency and other performance characteristics. Brattgaard, et al (1) compared front-and rear-wheeled designs and found the rear mounted configuration superior. Engel, et al (2) compared forward, middle and rear seat configurations for lever propulsion and found that less energy was required in the middle position than for either front or rear. The absence of an effective lever propulsion system adaptable to the conventional wheelchair has made the latter finding of relatively academic significance. It is therefore not particularly surprising to find only modest interest in the effects of seat position on wheelchair performance.

The advent of a new generation of center of gravity (cg) wheelchairs with the seat positioned over the main wheels equipped with both front and rear casters has awakened interest in the effect of seat position on performance.

METHODS

Four able bodied males, aged 26-33 served as subjects. They each participated in nine randomly assigned bouts of wheelchair propulsion of five minutes duration. Three tests were conducted on each of three days. Although none were wheelchair users all worked in rehabilitation related activities and were familiar with propelling a wheelchair. The tests were conducted on a motor driven wheelchair dynamometer with electric brakes which were used to set the work rate. A wheelchair seat adjustable in both horizontal and vertical directions was mounted on the dynamometer. Hand rims were mounted to the wheels by means of a connecting torque tube with strain gauges bonded to it. The dynamometer and torque rims were calibrated with standard weights. The output voltages from the dynamometer and torque rims were fed into an LSI-11 microcomputer. A nominal work rate of 0.25 watt/kg. was used for each subject and the output of the dynamometer and torque rims was sampled during the last ten seconds of each minute of work. Oxygen consumption and related ventilation measurements were obtained with a Beckman Metabolic Measurement Cart which provided print outs of these measurements and heart rate each minute.

Tests were conducted with the rear most part of the seat at heights of five, seven and nine inches above the axle and in horizontal positions with the rear of the seat above the axle, eight inches, and sixteen inches behind the axle, respectively.

Anthropometric measurements included upper extremity length, height of the shoulder above the seat, and weight.

Approximations of the points where the subjects grabbed and released the rim were made visually during each bout and recorded on a scaled drawing of the seat and wheels.

RESULTS

The power output was nominally set at 0.25 w/kg body weight. In one series of tests the nominal setting was inadvertantly doubled. For this reason the mean values for all tests at the middle seat height are reported with all subjects represented and again with the data at the higher setting excluded as noted in Table 1. The actual power outputs recorded from the dynamometer based on brake voltage and speed were quite consistent, but considerably lower then the pre-selected setting.

The higher values of efficiencies were generally obtained for the middle height positions with respect to vertical deviations and for the forward seat position with respect to horizontal displacement. The highest values were 7.23% and 7.21% obtained in the middle-middle and middle-forward positions. The lowest values for efficiency were the high-rear and low-rear seat positions. While there was considerable variation in the values obtained from the different

subjects, the order of efficiency according to positions was quite consistent for all subjects and closely approximated the order for mean efficiencies by position. The heart rate responses are generally consistent with efficiencies and the small differences which exist may be due to variation in workload.

The approximate positions at the beginning and end of the stroke and the magnitude of the stroke arc by position are shown in Figure 1.

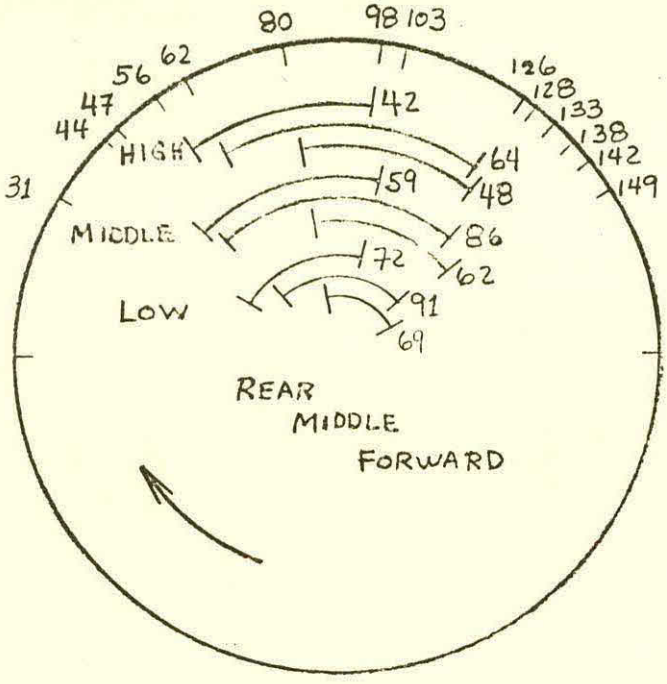

Figure 1
Grab and release positions and stroke arcs for nine positions

The stroke arc is lengthened by lowering the seat and shortened by raising it. The rear seat position was relatively more affected by the change in seat height.

The longest arc for all heights was found in the middle position. This is consistent with the lower stroke frequency and greater duration at this seat height for the forward and middle positions which were also the most efficient. While an approximation of optimum seat position is tenable on the basis of efficiency it would appear that an understanding of the relationship of other factors such as upper extremity length, sitting height, strength, and functional spinal cord level would be extremely important. The fact that efficiency is known to vary with power output, and the combination of speed and torque would also be factors for consideration.

On the basis of the results of this investigation it would appear that there are favorable compromises among the effects of greater stroke length, changes in muscle length, and working against gravity in the middle positions although not for the middle-rear position, as can be seen in Table 1. Also, evident from Table 1 is a slightly higher stroke to recovery ratio for the middle height positions. The peak torque values are lower, corresponding to the greater duration of the stroke cycle for the middle-forward and middle-middle positions.

DISCUSSION

The factors affecting wheelchair propulsion efficiency certainly include seat position, however, it would be hazardous to specify a particular optimum seat position as upper extremity length, utilization of gravity, trunk motion and other factors make it difficult to isolate the limiting factor or factors which are the determinants of wheelchair performance.

The advantage of greater stroke length in the middle-middle position would appear to be offset by the more advantageous continuous application of force along a downward trajectory in middle-forward position and possibly by a greater range of extension's at the elbow joint. The latter is by no means certain as the possibility also exists that accomplishing the stroke with less change in muscle length may be more efficient. It is quite clear that muscle function changes, in some positions (i.e. high-rear and low-rear) rather dramatically. The importance of variation in muscle involvement and function relative to seat position when prescribing a wheelchair for a user with limited upper extremity function is apparent.

While it has not been possible to answer most of the interesting questions with regard to the effect of seat positions on wheelchair performance, the need for biomechanical and electromyographic analysis of this problem has been documented. This need will be accentuated by new designs especially for c.g. wheelchair and lever driven vehicles.

1. Brattgaard, S.O., G. Grimbly, O. Hook: "Energy Expenditure and Heart-Rate in Driving a Wheelchair Ergometer." Scand. J. Rehab. Med. 2: 143, 1970.

2. Engel, P., M. Neikes, K. Bennedik, G. Hildebrandt, F.W. Rode: Arbeitsphysiologische Untersuchungen zur Oplimierung des Sitzanordnung beim handhebelbetriebenew Rollstukl Rehabilitation 15: 217-228, 1976.

Table 1. Mean values for stroke frequency, stroke duration, stroke to recovery ratio, power output, energy cost, efficiency and heart rate for wheelchair propulsion in nine different seat positions

Seat Position	Frequency (strokes/min)	Duration (Seconds)	Stroke to Recovery ratio	Power Output W/kg)	Energy Cost W/kg	Heart Rate (beat/min)	Efficiency %		
							Mean	High	Low
High Forward	14.22	60.5	0.45	45/55	0.167	2.56	87.0	6.52	7.78	5.45
High Middle	14.43	61.6	0.45	46/54	0.162	2.60	88.3	6.23	7.65	5.57
High Rear	14.81	61.9	0.46	47/53	0.162	3.02	95.7	5.36	6.30	4.67
Middle Forward	14.80 12.66[y]	53.1 46.8[y]	0.54 0.60[y]	46/54 48/52[y]	0.189 0.158[y]	2.62 2.28[y]	87.0 85.0[y]	7.21 6.93[y]	8.60	6.04
Middle Middle	14.71 12.58[y]	54.9 49.2[y]	0.54 0.59[y]	48/52 49/51[y]	0.193 0.162[y]	2.67 2.22[y]	85.3 83.3[y]	7.23 7.30[y]	8.50	6.39
Middle Rear	16.31 13.93[y]	63.7 61.0[y]	0.48 0.52[y]	53/47 51/49[y]	0.188 0.157[y]	3.21 2.68[y]	91.0 91.0[y]	5.86 5.86[y]	6.90	5.33
Low Forward	14.56	56.4	0.49	46/54	0.161	2.48	86.3	6.49	7.35	5.55
Low Middle	14.57	57.6	0.48	46/54	0.159	2.55	88.3	6.24	7.94	5.55
Low Rear	13.43	60.3	0.46	46/54	0.164	2.86	92.0	5.73	6.23	5.21

x N=4

y N=3 One subject was given a higher work load for the tests in the middle height positions. Additional resultes are present without his data.

AZIMUTHAL CONTROL FOR A POWERED WHEELCHAIR

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ABSTRACT

Due to variances in the components of the drive systems of powered wheelchairs and the effect of terrain on the front casters, powered wheelchairs tend to depart from an intended straight path. The problem becomes more acute when traversing inclined surfaces such as sidewalks sloped for drainage. The frequent corrections required of the operator may cause an individual with limited control capabilities to tire and become less effective in their activities. A microcomputer-based system has been developed to control the azimuth of a powered wheelchair by individually adjusting the torques of the drive motors. Performance of a wheelchair with and without the azimuthal controller is compared.

INTRODUCTION

Limitations in the matching of the electromechanical drives of the rear wheels of a powered wheelchair and variations in the terrain over which the chair is driven combine to divert the chair from an intended straight path. Sidewalks sloped toward the street for drainage make it particularly difficult to maintain the chair's direction as the casters tend to turn with the slope and direct the chair toward the street. To maintain a desired straight path, i.e., azimuth, the operator must periodically reorient the chair. The constant vigilance and effort required of the operator under such conditions may prove tiring or annoying and restrict the operator's effectiveness. This problem is particularly acute for individuals with very limited control capabilities.

A microprocessor-based system is described which automatically compensates for deviations of the chair from an intended azimuth. The system is capable of countering deflections due to mismatching of the drives and sloping or unevenness of the terrain being crossed. Performance of a powered wheelchair with and without the azimuthal controller is compared.

CAUSES OF AZIMUTHAL DEVIATION

Deviations of the path of a powered wheelchair from a straight line can be attributed to two primary factors: (1) the electromechanical drive systems of the rear wheels and (2) the effect of terrain on the alignment of the front casters.

In the first case, variances in the components of the drive units generally result in one drive wheel rotating slightly faster than the other drive wheel. The faster wheel will attempt to traverse a greater distance than the slower wheel for a given increment of time, thus exerting a torque on the chair to divert it from a straight path. If the front casters are free to pivot, as they normally are, the torque on the chair due to the imbalanced drives will align the casters so that the chair will travel an arc. Differences in the torque produced by the two wheels may result from differences in the efficiencies of the motors, variations in the wear and adjustment of the linkages between the motors and the wheels, and in the case of belt linkages, slippage between the belts and pulleys.

Another factor in the differences between the drive systems are the drive wheels themselves. If the tires are unevenly worn or, with pneumatic tires, if the inflation pressure is different, the effective diameters of the rear wheels will be different. Assuming the wheels are rotating at the same rate, the wheel with the greater effective diameter will travel a greater distance than the wheel of lesser effective diameter, and the path of the chair will again be an arc.

If the wheels are not rotating at the same rate, the effect of the difference in wheel diameters may subtract from or add to the effect of the differences in the drive units. The overall effect on a powered chair's path will consequently vary over a wide range from chair to chair. Some chairs of a particular design will require only an occasional correction to maintain a straight path; whereas, other chairs of the same design will require frequent corrections.

Because the degree of deviation due to drive and tire differences varies greatly from chair to chair, some powered chair users may not consider this a serious problem. Of greater concern is the second factor which produces path deviations; the effect of terrain on the alignment of the front casters.

Figure 1 shows a schematic representation of a chair moving perpendicular to the slope of an incline. The chair's center of mass has been transferred to point O for purposes of illustration. Assuming perfectly matched drive units and identical drive wheel diameters, the torque of the drive units produces a force F_D which propels the chair forward parallel to the edge of the incline. The gravitational force vector F_G is at an angle to the chair's perpendicular axis equal

to the angle of the incline. F_G can be resolved into a component $F_{G\parallel}$ which is aligned with the perpendicular axis through the chair and a component $F_{G\perp}$ which is perpendicular to the axis and points in the direction of the incline's slant.

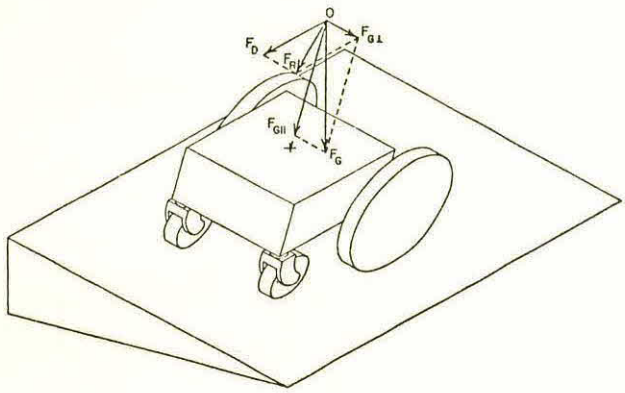

Fig. 1. Schematic representation of forces acting on a wheelchair traversing an inclined surface.

The perpendicular gravity component $F_{G\perp}$ tends to pull the chair down the incline. If the front casters are locked so that they cannot pivot, the friction between the wheels and the inclined surface prevents the chair from sliding sideways down the slope. However, the mechanical arrangement of the front casters places their point of attachment to the chair in front of their point of contact with the surface. Because the casters are free to pivot, the force vector F_R (resulting from the forward pointing force of the drive units F_D and the perpendicular gravitational force component $F_{G\perp}$) causes a torque to be exerted on the casters at their point of attachment to the chair. This torque forces the casters to pivot on their point of contact with the surface and turn with the slope, leading the chair down the incline. The casters continue to turn with the slope until F_D and $F_{G\perp}$ are aligned and the torque diverting the casters vanishes. At that point, the chair is moving forward down the incline.

Traversing inclined surfaces is a serious concern for individuals with very limited control of their powered chairs. Inclined surfaces sufficiently sloped to affect a chair's direction of travel are common. Sidewalks are generally sloped toward the street to facilitate drainage, and driveways crossing sidewalks produce local increases in the slope of the surface and a greater deflection of the chair. Such conditions are potentially dangerous because they divert the chair toward the street, curb, and traffic. Local fluctuations in surfaces also produce random changes in a chair's direction and must be countered if a straight path is to be maintained.

CONTROL HYPOTHESIS

Whether a powered chair deviates from an intended straight path due to differences in the electromechanical drive systems or due to the effect of terrain on the casters, it does so by turning, with one rear wheel traveling a greater distance than the other. A control scheme which forces the wheels to travel equal distances when the operator wishes to move along a straight path will maintain the intended direction of travel regardless of either of the sources of deviation described. This hypothesis formed the basis of the azimuthal control system.

IMPLEMENTATION

An E and J, type 33, 12-volt powered wheelchair modified with a Northwestern University controller [1] was selected for implementation of the azimuthal control scheme. (See Fig. 2). The chair's power unit supplies 400 Hz pulse width modulated armature current to each of the two permanent magnet drive motors. The width of the current pulses controls the torque produced by each drive system and determines the velocity of the chair. By selectively controlling the width of the pulses to each motor, the torque on each drive wheel could be adjusted to produce a resultant torque on the chair. With monitoring of the displacements of each wheel, this resultant torque could be regulated so as to compensate for diverting torques due to differences in the drives and tires and the slope of the terrain, and thus keep the circumferential displacements of each wheel equal.

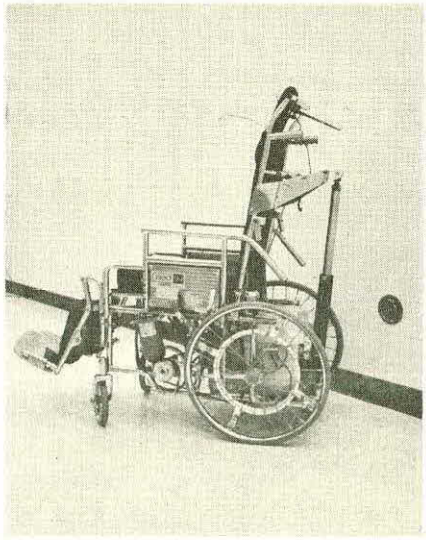

Fig. 2 E and J, type 33, 12-volt powered wheelchair with Northwestern University controller

A Motorola 6800 microprocessor-based system was selected to perform the wheel displacement detection and pulse width control of the motor current. As the project progressed, the greater flexibility of the microprocessor became apparent and all control operations including decoding of user inputs and acceleration limiting [1] were implemented with the processor.

To detect wheel displacement, two friction-mounted bicycle generators were modified by replacing their armatures and stators with rotating perforated discs and electro-optical sensors. The sensors were attached to the frame of the chair in contact with the rear wheels. (See Fig. 3) They were designed to produce one pulse for every 0.04 inch of circumferential displacement (0.227 degree of angular displacement) of the rear wheels.

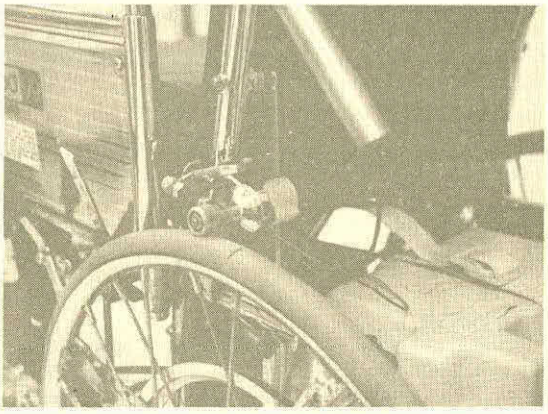

Fig. 3. Wheel displacement sensor modified from a bicycle generator and friction-mounted on rear wheel

As a wheel rotates, its sensor produces a train of pulses. During a given increment of time, the number of pulses is proportional to the circumferential displacement of the wheel. If during a time increment, the pulse trains from both sensors have an equal number of pulses, the wheels have traveled equal distances. If the pulse train from one sensor has more pulses than the train from the other sensor, the wheel which produced the greater number of pulses traversed a greater distance than the other wheel, and the chair has turned from the intended azimuth.

The quantity of interest is the difference between the number of pulses in each train. To extract the difference, one pulse train is used to increment a counter while the second pulse train decrements it. As long as the wheels travel equal distances over a given period of time, the number of pulses in each train are equal and the resulting value of the counter is zero. If one wheel travels a greater or lesser distance than the other, the counter registers a positive or negative value. The microprocessor periodically (once every 2.5 milliseconds) checks the value of the counter and adjusts the pulse width of the drive of one of the motors so as to return the counter value to zero. As a result, the chair is realigned to the intended direction of travel.

The correction in the drive pulse width is applied to increase the torque of the motor that counters the diverting torque on the chair, if the width of the pulses to that motor is not already set at its maximum. If both motors are receiving pulses of the maximum width, the pulse width is reduced to the motor which produces a torque on the chair that contributes to the diverting torque.

The amount of change in the drive pulse width for a given difference between the number of pulses produced by the sensors was determined empirically. The correction factors were selected to provide adequate control of the azimuth without significantly reducing the mean velocity of the chair and without inducing oscillations because of overcorrection.

Figure 4 shows a block diagram of the wheelchair control system. It should be noted that the azimuthal controller is not operational during the initial acceleration phase of the chair's drive and whenever a turning command is given by the operator.

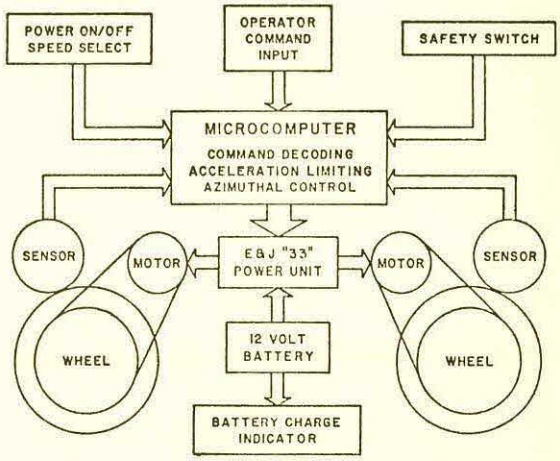

Fig. 4. Block diagram of microcomputer-based wheelchair control system

RESULTS

The sensitivity of the azimuthal controller was determined by locking one wheel while rotating the other rear wheel until the counter registered a difference of one. The change in the orientation of the chair was the amount of deviation to which the controller would respond. Given that the distance between the rear wheels was 22 inches, a change in azimuth of 0.10 degree from its original orientation could be detected and would have resulted in a corrective response had the controller been engaged.

To obtain a quantitative measure of the efficacy of the azimuthal controller, six trials were run with the wheelchair driven along a laboratory corridor with and without the controller operating. The chair was driven without corrective turns until it encountered one of the

corridor's walls or traveled a maximum distance of 100 feet. As a measure of performance, the perpendicular distance from the ideal straight path to the final position of the chair was measured and divided by the projected distance of travel along the ideal path.

For six trials without the azimuthal controller, the mean ratio of the deviation to the projected distance traveled was 0.110. With the azimuthal controller, the mean ratio was 0.002. Without the controller engaged, the wheelchair could not traverse more than 30 to 40 feet (deviations of 3.3 to 4.4 feet) before a wall was encountered. With the controller engaged, the wheelchair traversed the entire 100 feet without significant deviation. If the travel distance without the azimuthal controller was extrapolated to 100 feet, the amount of deviation would be 11 feet, compared to a mean value of 0.2 feet (2.4 inches) with the controller.

The corridor tests indicated that the control scheme was adequate to correct for deviations due to differences in the drive systems and due to slightly uneven surfaces. To verify that the chair with the controller could traverse an inclined surface, the chair was driven along sloped sidewalks in the vicinity of the laboratory. Qualitatively, the azimuthal controller was able to compensate for slight to moderate sidewalk pitches.

On significantly pitched sections of the walk, however, such as driveways leading into the street, the perpendicular component of the gravitational vector overpowered the torque produced by the drive system and the chair would turn with the slope. Observations revealed that the azimuthal controller was providing the maximum correction factor; one motor fully off and dynamically braked, the other motor fully on driving against the slope. In spite of the azimuthal controller, gravity was forcing both wheels to rotate and the state of the drive currents to the motors was no longer relevant. Because the controller was still operating under this extreme condition but the motors were ineffectual, use of a more powerful drive system would probably produce significantly better results.

CONCLUSIONS AND RECOMMENDATIONS

The described azimuthal control scheme was shown to be effective in maintaining the intended direction of travel over uneven surfaces and over slightly to moderately pitched sidewalks. The system was also able to compensate for differences in the electromechanical drives to the rear wheels and the effective diameters of the tires.

Although adequate for this study, the friction mounted sensors were determined to be impractical in a final design. For a given increment of time and angular velocity of the wheel, the number of pulses produced by the sensor was dependent on the angle between the longitudinal axis through the sensor and the tangent in the plane of the wheel at the point of contact between the wheel and the sensor. A sensor mounted in contact with the outside surface of the tire is also susceptible to error due to materials picked up by the tire, such as mud, snow, gum, etc.

Present work with the azimuthal controller has been involved with the application of the concept to the faster and considerably more powerful model 3P E & J, 24 volt, powered wheelchair. The wheel displacement sensors on the 3P chair are commercially available electro-optical velocity encoders mounted on the chair's frame. To eliminate the alignment problem of the friction mounted sensors, the newer sensors are geared to a large ring gear mounted on each wheel. This approach also prevents materials picked up by the tires from interfering with the sensors' operation.

The improved performance characteristics of the model 3P chair over the type 33 chair are expected to enable the user to traverse steeper inclines without gravity overpowering the chair's drive systems.

Based on this project, an upgraded version of a powered wheelchair with azimuthal control could remove some of the difficulty in operating a powered chair in a non-ideal environment, and, as a result, increase the effectiveness and range of the chair's operator. It is possible that such a system may also make powered wheelchairs available to potential users whose degree of disability makes control of a powered chair impractical at this time.

ACKNOWLEDGEMENT

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AN IMPROVED HEAD CONTROL FOR ELECTRIC WHEELCHAIRS

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ABSTRACT

The paper describes a modification to the head control system previously developed at the Rehabilitation Institute of Montreal for electric wheelchairs using potentiometric proportional control. The main advantage is that a standard electronic controller module may be used because all the necessary electronics are contained in the head control unit. A single switch enables the quadriplegic to select Forward, Off or Reverse. The head control has only one cable, simplifying its attachment or removal from the wheelchair.

INTRODUCTION

The well-known head control developed by Lozac'h (1) at the Rehabilitation Institute of Montreal uses a modification of the standard proportional control potentiometric joystick. This head control has benefited many quadriplegics by providing them with independent mobility. The system has been employed by Everest & Jennings for head control of the model 33B electric wheelchair (2). This head control system requires a specially modified electronic controller module in addition to the specially constructed control box.

In a wheelchair program which includes both head control and hand control wheelchairs from one manufacturer, it is desirable for all of these wheelchairs to make use of the same standard electronic controller module. This makes interchange of modules between different wheelchairs easier and avoids having to stock more than one type of module.

The head control system described in this paper makes use of the standard electronic controller module provided for potentiometric joystick-control wheelchairs. The only changes required are in the control box which not only has mechanical modifications similar to the Montreal head control but also incorporates all the necessary electronics.

CONTROL BOX MODIFICATIONS

The mechanical changes to the joystick box are virtually identical to those described by Lozac'h (1). However, the Montreal head control uses two separate shoulder switches for Forward-Reverse and On-Off operations. In the design described here, only one switch is required and successive operations of the switch cause the chair to step through the sequence ...Forward, Off, Reverse, Off ...

The electronics has to be flat and compact to fit inside the control box and therefore uses a low profile 12-volt four pole double-throw relay (Potter & Brumfield T10-E2-Y4) to achieve forward and reverse selection by interchanging the end connections of the potentiometers. This relay and the remaining electronics are mounted on a small circuit board which is attached to the inside of the control box housing.

ELECTRONIC CIRCUIT DESCRIPTION

A circuit diagram of the electronics for a 24-volt system is given in figure 1. The circuit for a 12-volt system would omit the Zener diode D2. The pushbutton switch operated by the quadriplegic is of the latching double-pole double-throw type. One pole (P1) is in series with the main on-off switch of the control box and successive operations of the pushbutton cause the wheelchair to follow the sequence ... On, Off, On, Off .... The other pole (P2) provides clock signals for a dual D-type flip-flop circuit contained in a single CD4013 integrated circuit package. This dual circuit behaves like a single bistable flip-flop without the need for a contact bounce eliminator at the clock input. Successive operations of the pushbutton cause the output Q2 to follow the binary sequence ...0110... and the relay K to follow the sequence ... Off, On, On, Off ... The direction of the motor drive thus follows the sequence ... Forward, Reverse, Reverse, Forward ... The combined effect of the two poles operating simultaneously is that successive pushbutton operations cause the wheelchair to step through the sequence ... Forward, Off, Reverse, Off ...

When the Off state is selected, the electronics is still energized but the current drain is negligible due to the use of CMOS circuitry. In the Reverse state, the current drain is about 60mA, due mainly to the relay. The main On-Off switch of the control box deenergizes the electronics when switched to Off.

The non-polarized capacitors C1 and C2 are necessary across each potentiometer to provide electronic damping. This prevents the wheelchair and its occupant from going into a

sustained mechanical oscillation in reverse and also prevents the wheelchair from exhibiting a sudden jolt when switched on.

DISCUSSION

The head control system described has a number of advantages:

1. It enables a standard electronic controller module to be used for both head control and hand control wheelchairs. This reduces the number of different types of module to be kept in stock in a wheelchair repair program and simplifies servicing. A wheelchair may be quickly converted from head control to hand control without a change of module.

2. Only one shoulder switch is required to obtain Forward, Reverse and Off functions. Also the problem of protection against motion reversal when the wheelchair is travelling at speed (in the Montreal system) does not arise because it is necessary to pass through the Off state in switching between Forward and Reverse.

3. Only one cable connects the head control box to the module so that attachment and removal of the head control is simplified.

The switch for control of Forward, Off and Reverse may be placed at any suitable location to harness an appropriate movement. In the case of a polio quadriplegic in a Winnipeg hospital, lateral motion of the left knee was used to select Forward, Off and Reverse in a model 3P wheelchair.

The modifications described in this paper have been applied to two commercially available electric wheelchairs viz. the Everest & Jennings 33B and 3P wheelchairs and, in each case, a trial of the system by a quadriplegic over a period of several months was successful.

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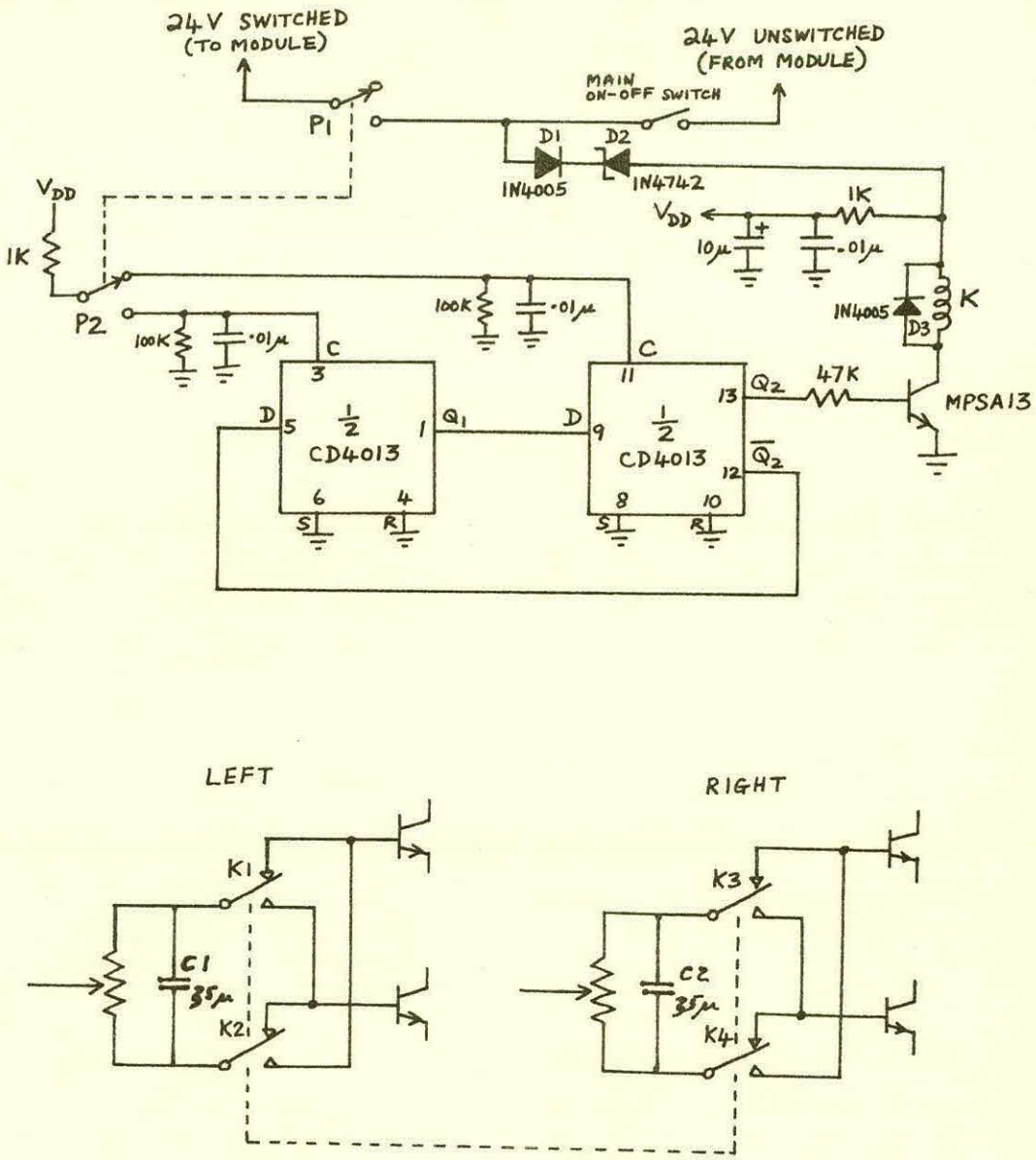

NOTE : TRANSISTORS IN THIS DIAGRAM ARE LOCATED IN MODULE.
POTENTIOMETERS AND RELAY CONTACTS ARE IN CONTROL BOX.

FIG. 1 ELECTRONIC CIRCUIT DIAGRAM
FOR IMPROVED HEAD CONTROL
(24 VOLT SYSTEM)

Individually Customized Postural Support System

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When presently available postural support systems were grouped together for evaluation, it was felt that commercially available individually customized wheelchair inserts did not adequately represent current fabrication techniques.

It was decided that we could develop a customized seat incorporating the customization into standard industrial fabrication techniques. The seating process consists of a fitting technique utilizing a bead bag, allowing for positioning and functional assessment of the client's proposed position. A plaster cast of the seat's exact shape is made, and a layer of soft foam, backed with hard plastic, is then thermally formed to match the contours of the plaster mold. The resulting insert, which faithfully reproduces the shape of the original seat, is then finished with Naugahyde and fitted to a standardized ABS rigid shell which attaches to a wheeled base (standard wheelchair frame, Pogon buggy, etc.). The entire seat unit can be easily detached from the wheeled frame for convenient transportation.

Tufts-New England Medical Center's Biomedical Engineering Center has been involved with client positioning for several years. There is a need for a Bioengineering solution to the many seating problems. Enhanced use of extremities, improved communication, minimization of medical complications of skin and tissue breakdown, and scoliosis and deformity prevention are all involved with a postural support system. Therefore, an analysis is necessary of the various possible systems to evaluate their function against a variety of goals.

The postural support systems currently available were studied and it was noted that there was no cost efficient fully customized contoured seat available, and that this would be an essential addition to the range of postural support systems. We determined to develop a seat which would accurately reproduce the body contours with a high resolution while also striving for a fast, low cost fabrication technique minimizing individualized custom labor. The customizing is best accomplished if it is incorporated in the fabrication technique.

The Technique

The client is assessed for appropriateness of a fully contoured seat. Primary seating objectives are established and the client, and his or her clinician, and a Tufts Occupational Therapist meet for a fitting session. A cloth-reinforced 4 mil polyurethane bean bag was fabricated and 1/2 filled with 1/16" dia. polystrene beads. When fitted with a seal, the air can be evacuated using a standard hospital vacuum pump which causes the beads to crush together, reducing the volume only slightly causing frictional forces between beads and a rigid structure. The property of the bag can be changed by using a larger, "foamier" bead, causing some cushioning, and a greater amount of volume reduction when the air is evacuated. A smaller firmer bead will result in a firmer bag, with little volumne reduction - obviously an aggregate alters the characteristics differently. The bag is supported by an adjustable seat back which attaches to a common work table allowing the entire assembly to be inclined for different situations.

At fitting time, the client is shown the chair and given an explanation of its workings, and the client's clinician and an attending occupational therapist assist in positioning. The beads are arranged to an approximation of the position and the client placed in the chair. With much pulling and tugging of the bag, the position will begin to take place - starting with the hips, and upward onto the trunk. Conformity is achieved, providing lateral support and other necessary contours.

Care is taken to see that the small of the back and other concave contours are recorded fully. Gross changes, inclinations, re-orientation to vertical, or correctional forces involving large areas -- are accomplished by manipulating the exterior of the bag, by blocking and pushing, to achieve the desired position. The vacuum is finally pulled as tight as possible and the patient is allowed to experience what the finished seat will be like. This is a time for functional

assessment of arms and legs, balance capability, etc. Straps are available to provide proper restraint, and the realistic expectations of the client are on trial. What is possible in this fitting chair will be possible in the seat - nothing more.

When it is felt that everything is satisfactory, the client is removed from the chair and the chair is examined for small bumps and folds that might bring discomfort or lead to eventual skin breakdown. The patient is often not sensitive enough to notice small irregularities that could cause trouble later.

The client is replaced in the chair and is allowed to sit for 1/2 hour or more, to be sure of comfort, position, etc.

When the client is removed, signs of skin redness can be checked as an indication of trouble spots. A small thumb pressure to the bead bag will leave a slight indentation in the seat, so that the finished seat will apply less pressure at this spot. Spina Bifida can be accommodated in this way. Orientation of the acceptable seat is now measured so that it may be reduced with respect to the floor. A vertical rod is placed in the seat (with the aid of bubble level placed on the top.) describing a reproducible axis. The rod forms a small dent in the bead bag which will be locateable on the finished seat. Measurements may be taken from this axis to specific points on the chair (identified by more tiny dents). These measurements are simplified by using small rods held perpendicular to the axis, measuring their lengths and relationship to one another. A line pointing straight forward is drawn on the cast with a colored drawing pencil, which will transfer to the moist plaster casting material.

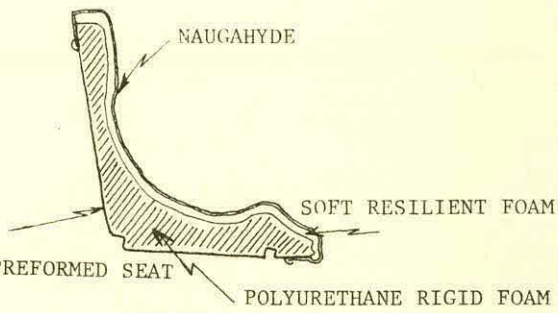
Wrinkles in the seated region are taped flat with masking tape, bridging any gaps smoothly. Plaster casting cloth is then applied to the seat, covering it entirely, with 2 or 3 layers to provide enough strength in 20 minutes to allow cast to be transported without breaking. The vacuum is broken at that time and the cast is pulled free -- a very close copy of the original body.

Photographs are taken of the positions for references and greater comprehension in the fabrication stage. Measurements are also taken of the wheelchair or buggy to which the seat must adapt.

In the fabricating shop the case is inverted and decisions are made as to how to mate the body cast with the outer shell which has straight lines and flat surfaces. Cardboard is taped to the cast, forming an enclosure between the inverted cast and the table. The inside of the body cast is then filled with a 2-pound destiny, two-part polyurethane foam, the type used in flotation for boats. This strengthens the cast for the heat-vacuum forming process.

Normal manufacturing safety precautions are observed in these steps. All the components are present in an automobile interior and present no known health hazards.

The cast is then set upon a vacuum-thermo-forming machine. Some materials, when heated, exhibit amazing flexibility and stretchability, and upon cooling, revert to a stable, normal state in their new positions. ABS (a hard plastic sheet), a foam cushion sheet of ethylene vinyl acetate foam (often called evazote or ibilite), and Naughahyde (a vinyl) exhibit this behavior and lend to this technique precision and accuracy. The soft foam cushion comes in 1/4" sheets and is laminated to the desired thickness. All sheets but the outer sheet are perforated to allow the vacuum to pull evenly on the outer sheet. When heated, these sheets bond completely together and conform to the plaster cast providing the total cushioning in the seat. A 1/8" sheet of ABS is formed over this foam, which heat-bonds to the plastic, holding the seat contours firmly. This shell is trimmed according to lines drawn on the plaster cast. (It is found that a wax marking pen will transfer a line from the plastic cast to the heated, soft foam, allowing lines to be carried from the bean bag seat to the contoured shell.)



The shell is then fitted to a pre-formed ABS bucket seat which will be the finished exterior of the seat, to which mounting hardware has been attached. The orientation measurements are used and the seat shell is taped to the bucket and 2 pound-density polyurethane foam is poured in to fill the large void between the two shells. After this foam has set to form a lightweight and rigid material, excess foam is trimmed off and covered with resilient seating material so that there are no hard edges to abrade the Naughayde covering.

The entire chair is placed in a box with the portion to be covered with Naugahyde facing upward. This surface can then be thermally vacuum formed. Several holes are drilled in the deeper seat pockets so that the heated sheet will be drawn into these cavities by the vacuum.

Glue is brushed onto the entire surface and when tacky, the heated sheet is formed over it. The edges of this sheet are trimmed and fastened to the outer seat shell, using a trim strip, gluing down loose edges.

The seat and its hardware fasten onto a baseplate which is engineered to fit the desired vehicle, a wheelchair or buggy, etc.

We are in the process of researching proper seating positioning to develop a means of evaluating the effectiveness of an individual's postural support. In the process of this research, we found a need to augment the existing technology of seating by developing a manufacturing technique that can produce a cost effective customized support. The manufacturing technique that has been described in this paper will now serve as a tool in the ongoing research project.

A COMPARISON OF THREE CUSTOM SEATING TECHNIQUES

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Seating patients with severe and fixed deformities requires a good technique for making a custom cushion especially for that patient. At the Rehabilitation Centre for Children we have been using three different methods for producing such cushions. This paper was written to convey our experience with each of these techniques and to weigh them against each other.

INTRODUCTION

Through our clinic and others (1, 6) we have seen patients for whom a modular seat or a simple wheelchair insert is not the answer. Usually they are severely involved patients exhibiting severe spasticity, high and fluctuating tone and joint contractures. Frequently, we see advanced scoliosis with hip and spinal deformities and accompanying rib hump. As well patients may demonstrate very little communication poor balance and, poor head and trunk control.

Such patients seen at the Special Devices Clinic are being supplied with custom-shaped seat cushions made by three different techniques the foam in place method, the foam in box method and the Moire method of anatomical shape sensing Natural strengths and weaknesses of each technique have become evident. We hope that by outlining each method and comparing them we can familiarize others with each method.

THERAPEUTIC OBJECTIVES

Much has been written on the objectives of good seating (1,2,3,4,5) and an overview of the criteria will be enough for our purposes. The primary concern for seating severely involved patients is patient comfort. This is achieved through a proper fitting cushion which provides total support. As well the cushion should be constructed to maintain good optimal sitting posture. The cushion material and fabric itself should be durable enough to withstand attack by perspiration. At the same time it must not encourage heat or moisture build-up close to the skin. These are the main characteristics of a comfortable seat.

Proper seating should allow easier maintenance of the patient including transportation, transfer, education and nursing care. It will provide the patient with protection and safety.

As far as musculo-skeletal deformities are concerned the best that can be reasonably hoped is to arrest the progression of any existing deformities and prohibit any additional ones from occurring. It is not reasonable to expect correction of contractures and deformities however it is not ruled out. With this in mind it is evident that patient comfort becomes the outstanding concern. A comfortable seat and cushion will allow the patient to sit functionally, which would otherwise be prohibited due to pain. To enable the patient with severe involvement to sit comfortably frees them from confinement in bed.

TECHNICAL SEATING OBJECTIVES

The process and materials must satisfy all of the appropriate therapeutic objectives and at the same time be technically feasible. A laborious and uneconomical process which frequently produces poor results is not acceptable. Before comparing the merits of the three techniques in question we should first examine the criteria on which we shall judge them.

Because patient comfort is the utmost concern, convenient positioning of the patient and maintenance of that position during assessment and fabrication of the seat is the most important objective. Because all three of these techniques record the patient's position and posture at one instant and produce a seat duplicating that, the clinic must be sure that, at that instant, the

patient is seated to their satisfaction. Any difficulties and uncertainties associated with the process will make it harder to get the patient seated satisfactorily before fabrication. The longer it takes to get proper seating the harder it is to achieve and maintain throughout the process.

The overriding factor in the economy of the process is the time it takes to complete. For each of these processes a trial cushion is first produced and then a final finished cushion, modified as appropriate. The time to complete each of these stages determines the cost.

The simplest process, i.e. simplest equipment and less technical procedure, has advantages. Because the equipment on hand and the manpower available limit the scope of the seating, a process must be within the realm of available equipment and manpower. Otherwise equipment and staff will have to be obtained at added expense.

Obviously, the process that produces the most predictable results with the best reliability is preferred. From our clinical experience it has become evident that the cushion which "wraps" around the patients sides between the hips and rib cage the most will provide most support. How well the cushion comes around the sides is the first indication of a successful seat. The more frequently we can get this result the more preferable the process.

The material itself rather than the process will determine the lifetime of the cushion. A durable material and fabric should outlast the patient's need for that cushion. Therapeutic considerations should be the indication that a new chair or cushion is required.

Finally the procedure must be safe and provide no hazard to patient or staff.

THE THREE METHODS UNDER CONSIDERATION

1. Foam-in-Place(FIP). The foam-in-place process consists of direct casting of the patient while seated. The patient is suspended on latex rubber over a cavity in a mould box. The cavity is then filled with ready mixed flexible polyurethane foam which sets up to form a cushion. The cushion is demoulded and fitted to the patient. If it proves comfortable a finished cushion and seat are produced.

The equipment necessary for FIP fabrication is an adjustable seat base, moulding boxes and clamps, a foam mixer and containers. The materials used are the two part polyurethane foam (9), 3 to 6 mil latex rubber sheet and a release agent.

Coat the inside of the mould box and one side of the latex sheet with the release agent and cover the side of the box with the latex sheet (coated side in). Fit the clamp onto the mould box holding the latex under vertically, but leaving some slack horizontally. Place a bath towel

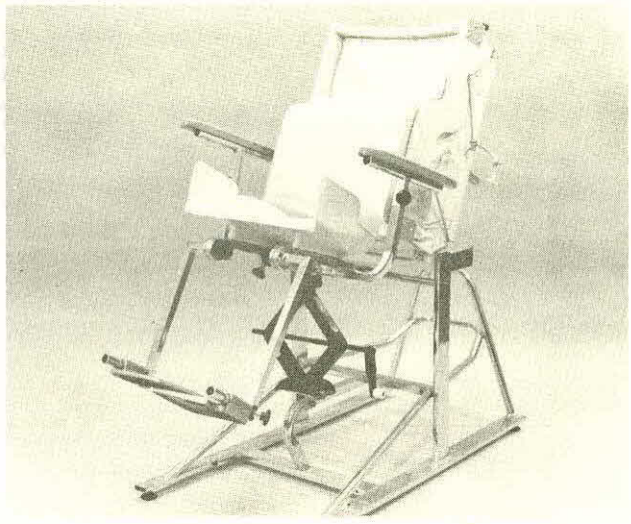

Fig. 1 Foam-in-Place equipment ready for patient

over the latex to protect the child from heat generated during the exothermic reaction of the foam. Seat the patient in the chair. He should be seated as far into the cahir as possible without bottoming. Pause at this stage to be satisfied that all of the therapeutic seating objectives are satisfied.

Mix the two foam components according to instructions. Mixing time is usually about 15 seconds, but may vary depending on the freshness of the components. Immediately pour the foam into the mould box and replace bung. Maintain the patient's posture as established for 3 - 5 minutes before removing him. Allow the cushion to cure for about one half hour.

Demould the cushion and make the initial fitting. If it proves suitable an ABS plastic shell is made to suit wheelchair dimensions and it is covered with fabric (double stretch terrycloth or bathing suit material). This description is adequate for our purposes here, however more detailed descriptions have been published. (2,6).

2. Foam in Box(FIB). The foam in box process consists of an idirect casting of the patient. It is a cross between the FIP technique and orthotic practice for producing a body jacket. The casting technique is borrowed from orthotic practice and the foaming and finishing are from FIP.

The equipment required consists of a casting table, wedge, sectional mould box, foam mixer and containers. The materials used are plaster bandage, laboratory plaster, one quarter inch polyurethane foam, 3 - 6 mil latex sheet, release agent, and two component polyurethane foam.

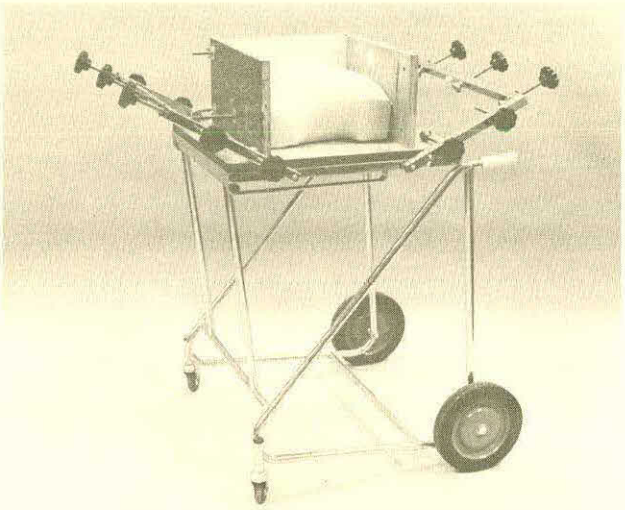

Fig. 2 Foam-in-Box equipment ready for pouring foam.

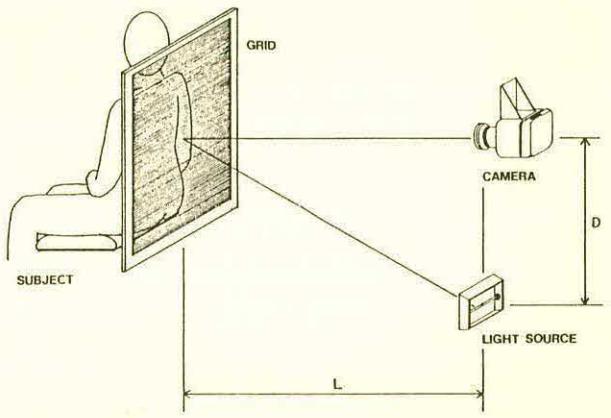

Fig. 3 Schematic of Moire photography equipment

With the wedge placed on the casting table, place the patient prone on the wedge. Position the patient as predetermined by the therapeutic seating objectives. Cast the patients back from the sacrum to the axillae and as far around the trunk as the cushion must extend with the plaster bandage strips. When these have hardened, fit an ABS pipe through the longitudinal center of the cast. Pour the positive pattern by fitting the casting with laboratory plaster. The pattern is lined with one quarter inch polyurethane foam to provide relief and avoid too much sanding. Now the pattern is covered with latex foam and placed into the mould box.

Coat the interior surfaces of the mould box and the pattern with release agent. Close the box and pour the foam as described for FIP. When the cushion has cured it is demoulded and finished as for the FIP cushion.

3. Moire Method. This technique employs a Shadow Moire photograph in conjunction with a replicator to produce a custom fit cushion. The Shadow Moire photogrammetry has been accurately described (2,7,8). Light from a point or linear source passes through a tightly spaced grid close to the patient and casts a series of shadows on his back. Viewing through the grid the eye or camera perceives its own series of shadows. The two sets of shadows intersect to form alternating light and dark planes parallel to the grid and at a distance characteristic to the geometry of the equipment. A photograph shows a topographical relief of the back with each fringe evenly spaced from the grid.

The equipment required consists of a Shadow Moire photogrammetry apparatus, (fig.3) and a replicator, (fig.5) to cut the shape from a foam block.

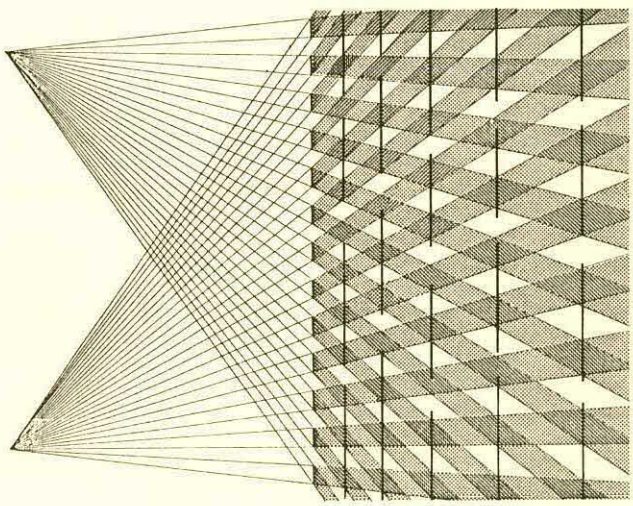

Fig.4 Generation of Moire fringes, by eye (camera) and light source, (both on right side of picture) and grid (dashed line).

Darkroom facilities are advisable. The materials used are the photographic materials, 8 inch thick polyurethane blocks, ABS for the shell, and covering fabric.

The patient is seated behind the grid and positioned in accordance with the therapeutic goals. When proper seating is achieved a photograph is taken. The photograph is enlarged to slightly larger than full scale to provide relief

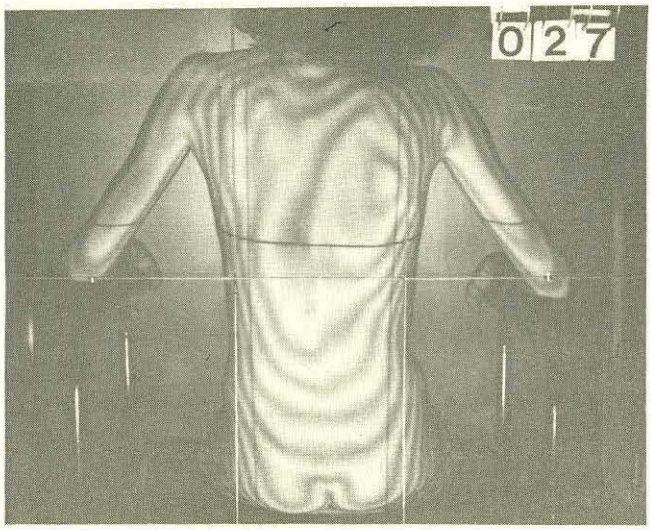

Fig. 5 (a)

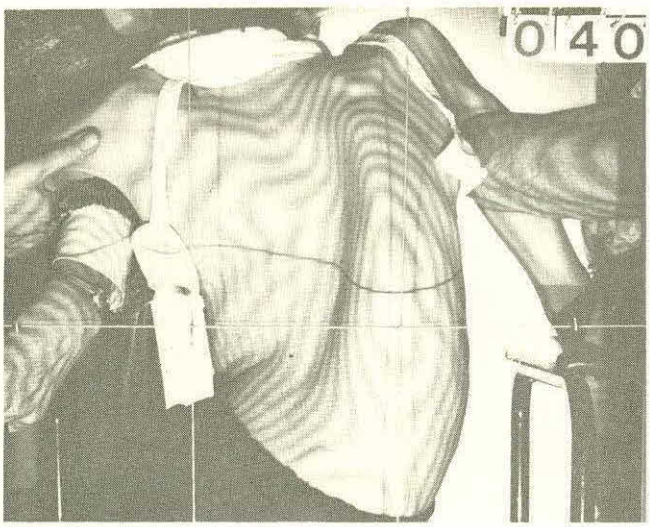

Fig. 5. (b) Moire photographs of two different patients.

relief in the finished cushion. Using the replicator (essentially a tracing machine) and knowing the distance between fringes a cushion can be cut from foam block. This produces a trial cushion which is tested and completed as for the FIP and FIB.

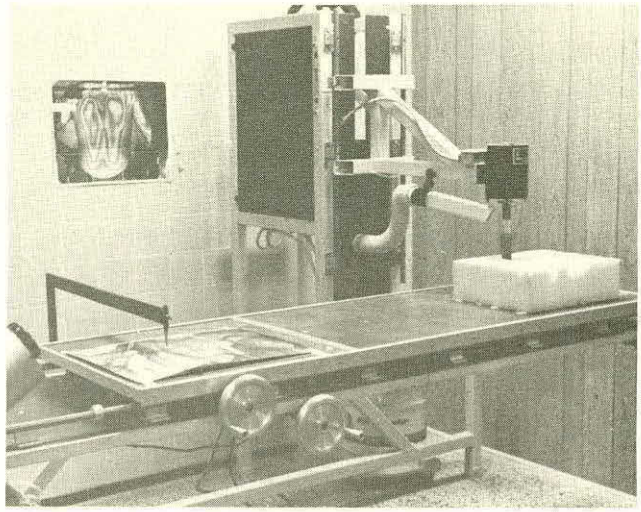

Fig. 6 Replicator

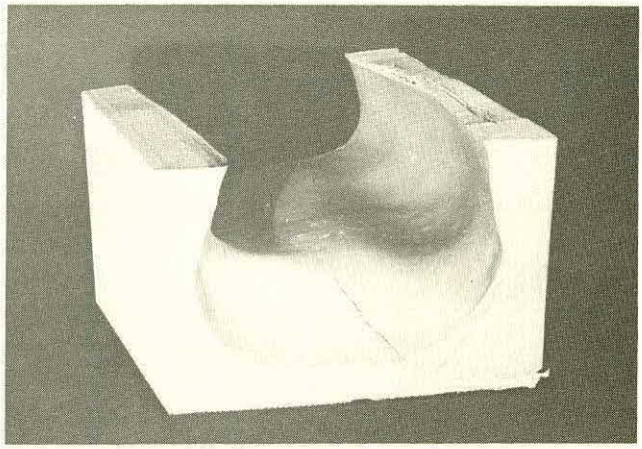

Fig. 7 Foam-in-Box Cushion.

COMPARISON

1. Recalling that the qualities of a cushion which enhance patient comfort the most are the wrap of the cushion and the patient position during fabrication we should weigh our comparison strongly on these.

Because of high control over casting the FIB cushion, the process offers the best wrap and is not restricted by undercuts. Due to the line of sight orientation of the Moire photograph and hammocking effect of the latex rubber in FIP, wrap beyond the midline is not expected. As well, the Moire photographs may be poorly defined close to the patient's outline.

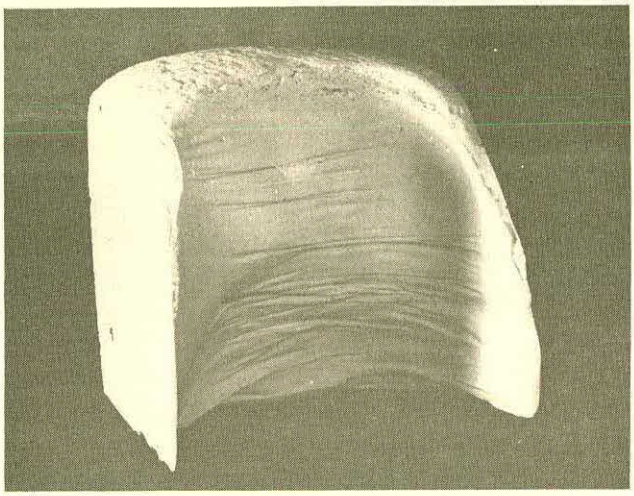

Fig. 8 Foam-in-Place cushion.

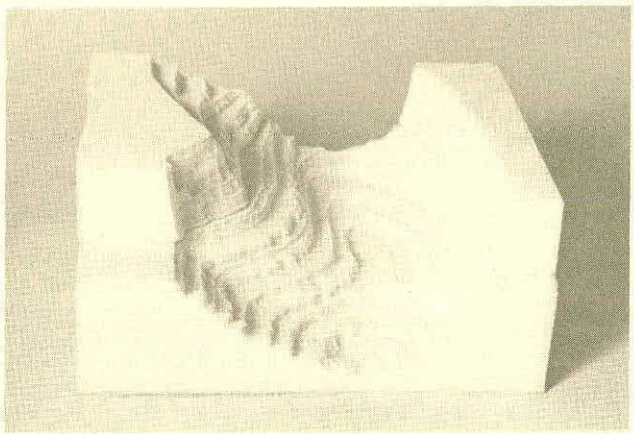

Fig. 9 Cushion made from Moire photograph.

2. The Moire photograph and FIP casting essentially freeze the patient's position at an instant in time. The patient's position at the moment of casting or photographing dictates the suitability of the finished product and therefore control of patient posture during this is critical. In the Moire technique the patient is positioned from the front. His back is unobstructed to allow easy assessment. However, the supporting hands may block part of the picture.

In the FIP technique assessment and positioning of the seating must be done from the front because the back is recessed into the mould box. Positioning is aided by the use of the adjustable chair which can duplicate the final insert prescription. During this process it is necessary that the patient maintain his position for up to five minutes when the foam is poured and sets up.

The FIB technique allows for the easiest patient positioning and manoeuvering as he is lying on his front and does not need to be supported. As well this allows a clear view of the back during casting. It may not provide a true to life situation as the effects of gravity are ignored.

3. A high quality cushion of even density and true to shape is desired. By using a closed mould during FIB a higher quality cushion of good density and self-skinning is achieved. The cast itslef can be modified to provide relief at critical areas noteably the axillae and iliac crests, prior to foaming. The cushion produced by FIP tends to be of lower density and have a less durable skin because the foaming pressure cannot be as high or the patient positioning is sacrificed. Air voids and surface blemishes are more likely a problem with FIP cushions. The cushion produced by the Moire method is the lowest quality of the three. It is characteristically terraced because of the cutting operation. Relief or over-sizing is more difficult with the Moire method.

It would seem that the FIB, because of the density control and self-skinning, provides the most uniform and durable cushion. To date clinical experience has not been enough to determine if the FIB cushion has a longer lifetime although it could be expected to outlast the others. The foam used in the Moire cutting is a medium density open cell polyurethane foam and is not as tough or resilient as the closed cell polyurethane of higher density achieved with FIP and FIB.

4. Because the Moire method and the FIB are remote from the patient there is excellent control over the cushion forming process itself. Pouring the FIP cushion requires more accurate assessment of the volume of foam required than does the FIB. Underestimating causes air voids and insufficient density while overestimating forces the patient out of the mould box and his position is lost.

5. The FIB method is the most time consuming because of the cast and mould box preparation. However, the cushion produced in this manner is easiest to finish. The FIP and the Moire methods involve about the same number of man hours. However, FIP produces results sooner because it does not require photographic processing and foam cutting. The requirements to produce a finished cushion are comparable in all three methods. Because of the lower control during foaming the FIP method requires slightly more cushion preparation than the FIB.

6. Of the three processes the Moire method certainly requires the most expensive equipment and facilities. This equipment has application in anatomical sensing and scoliosis screening which offset some the its cost (10). The FIB method uses existing orthotic skills and facilities with the addition of a simple mould box. The FIP equipment is more expensive than that for FIB, but still much less than Moire. The levels of expertise and familiarity for the FIB process and the Moire are cmparable and lower than FIP.

7. The safety of all three methods is good. The FIP and FIB foaming stages must be carried out in a well ventilated area (11,12,13).

	FIP	FIB	MOIRE
Wrap	Moderate	Excellent	Moderate
Patient Positioning	Moderate	Good	Good
Cushion Quality	Good	Excellent	Good
Process Control and Reliability	Moderate	Excellent	Good
Time Involved	Minimal	Greatest	Moderate
Facilities and Expertise	Moderate	Low	High
Safety	Good	Good	High

CONCLUSION

Some general statements can be made from this comparison. The FIB method provides the most supportive cushion with adequate ease of patient positioning. As well the FIB process will provide more predicable resluts. The FIP method produces a trial cushion sooner than either of the other methods.

Technically speaking each facility must make their own decisions based on their circumstances and the relative strengths and weaknesses of each method.

All methods are weak in that they produce a cushion which is based on the patient's position at one instant. There is little room for trial and error and the success relies heavily on the initial assessment of the patient. It is precisely this capacity for trial and error, and refinement of the seating before final foaming or photography that is the strength of the vacuum consolidated technique (1). It would appear that a combination of vacuum consolidation and foam in box would provide a powerful seating technique.

ACKNOWLEDGEMENT

The authors gratefully acknowledge the support given this work by Health and Welfare Canada under research grant 6607-1181-4.

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REUSABLE CASTING TECHNIQUE FOR CUSTOMIZED TOTAL CONTACT SEATING

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Virginia

ABSTRACT

A technique has been developed that reduces the fabrication time and cost for a custom made, total contact seating system. This new method substitutes a reusable mold made of crushed walnut shells for the plaster mold used in the dilatency casting approach. A vacuum is drawn on a negative and a positive impression of the person to be fitted with the special seating system. The positive walnut shell impression is then used as the mold over which vacuum formable materials are draped.

INTRODUCTION

Over the years, total contact seating for the severely disabled has evolved from a trial and error, carved from foam approach of seating fabrication, to a technique commonly referred to as the dilatency casting approach (1, 2, 3, 4). This effort involves taking an impression of an individual to be seated by placing them in a large air-tight, plastic bag filled with expanded polystyrene beads. After the person is positioned properly, air is evacuated from the bag and a firm impression of the seating surface for the individual remains. A plaster cast is made in the impression and a customized seat is fabricated on the plaster mold. This casting technique eliminates a significant amount of fabrication time over the upholstered cut and try method of foam inserts (5). There are, however, still a few areas that can be improved upon.

The plaster mold taken over a rigidized plastic bag often has unwanted grooves due to the wrinkles in the casting bag. Furthermore, obtaining a large plaster cast and the subsequent time for total drying significantly affects the turnaround time for a customized seat. This plaster impression method is labor intensive and hence it is also quite costly to the consumer. In his discussion of molded seats for the handicapped, Nelham (6) states "it requires some further development and alternative designs, capable of rapid, economic manufacture...". A new technique has been developed which eliminates the need for the plaster impression and problems associated with it. This new technique also offers additional variables for manipulation to obtain an improved structural support system.

METHODS AND MATERIAL

To eliminate the unwanted wrinkles and the subsequent time to touch up the plaster cast, a 5 millimeter thick latex sheet was used to construct a flexible membrane for a mold box. The mold box is constructed out of plywood that has been sealed with epoxy laminating resin to be air-tight and the latex is stretched over an open side. A valve is added to a side of the box to control the vacuum in the box. The mold box is filled with crushed and washed walnut shells. These shells are commercially available as filler for solid molds. The walnut shells are used to give a firmer impression than available by a polystyrene bean impression. The irregular side of the shells provides a locking action and since the shell particles are solid wood the final impression is very firm. The mold box also offers the advantage of predictable sizing for the finished seating system to interface with either a wheelchair or an automobile seat. The person to be fitted can then be placed in the mold box (Fig. 1) and an impression can be taken in a similar manner to the polystyrene bean bag method.

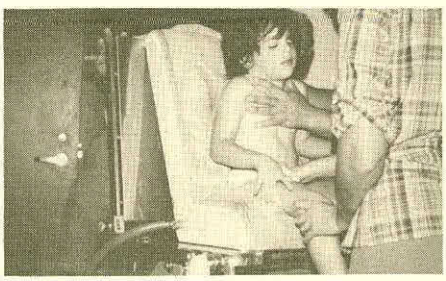

Figure 1
Client being positioned while negative walnut shell impression is taken.

The casting box was placed on an old operating table in order to get easy tilt adjustment. This tilt adjustment was probably not a necessity, however it did make optimum positioning of the individuals easier.

Another latex bag filled with walnut shells and attached to a board is them placed in the negative impression. Air is evacuated from this mold to obtain a positive impression of the

client. This impression, because of its firmness, can then be used directly as a mold for a vacuum formed, total contact seat. By careful adjustment of the amount of vacuum on the seat, the walnut shells become a working consistency of wet sand and therefore can be hand molded through the stretchy latex. This allows special features such as thoracic support or retention humps, to be accented or shifted for more effectiveness.

With the vacuum assured by an air-tight mold box and valve arrangement, the positive mold (Fig. 2) is then transported to the vacuum forming table where the actual seat will be shaped around it.

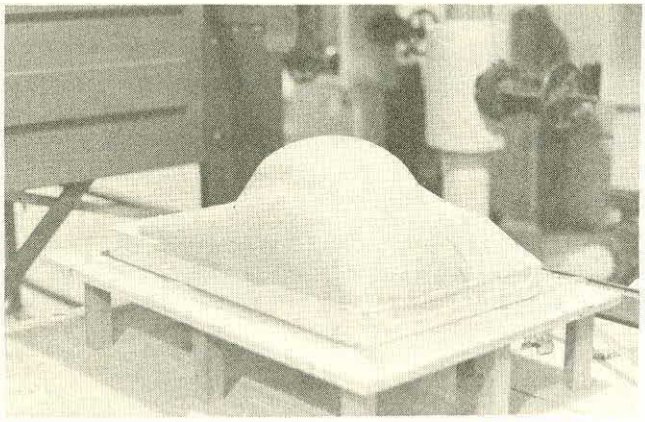

Figure 2
Hard, positive impression shown ready for vinyl to be vacuum formed over it

The total contact seat can be fabricated by vacuum forming urine-resistant, vacuum formable vinyl over the positive impression and then vacuum forming a selected thickness of polyethylene foam over the vinyl. An ABS (rigid plastic) outer shell is vacuum formed over the polyethylene foam. The vacuum used on the vacuum forming machine is adjusted to be 80 percent of the vacuum drawn on the walnut shell impression.

After the positive mold has been placed on the vacuum forming bed, nylon bathing suit material (stretchable in two directions) is thumb-tacked to the mold. The bathing suit material allows an air space for the vacuum to pull the first layer of vinyl over the positive mold for total contact. The vinyl has an expanded polyester backing on it already, and this backing permits the polyethylene foam to be subsequently drawn down over the vinyl. The foam can be directly bonded to the vinyl with spray adhesive (Fig. 3).

Bathing suit material is then placed over the foam to allow total contact draw of the ABS over the foam, because once again the material gives an evacuation air space. All three pieces

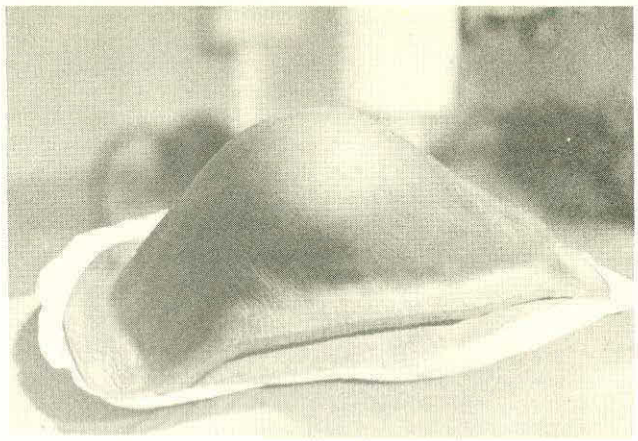

Figure 3
Polyethylene foam and vinyl shown formed over walnut shell mold.

can be removed from the mold and then bathing suit material removed from the layer between the ABS and the foam. The foam can then be spray-glued to the properly trimmed ABS seat insert. If special interfacing bars are required for a wheelchair or some other mobile frame (child stroller) they can be attached to the ABS shell before the vinyl-foam sandwich is adhered to the ABS (Fig. 4).

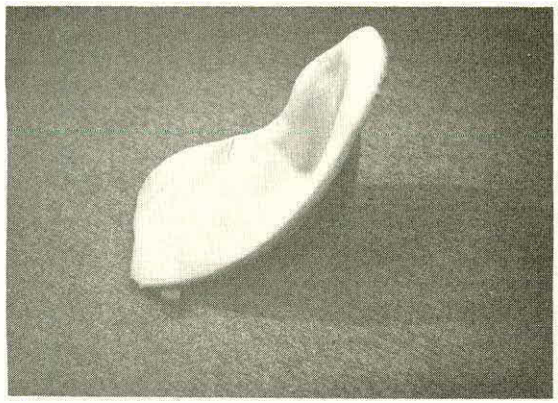

Figure 4
Final seat shown with wheelchair interface bar attached.

After the negative impression has been taken, the whole process for seat fabrication takes approximately only an hour. This is a substantial time savings and cost reduction over the plaster impression method. After the vacuum has been released on the positive walnut shell impression, it is reusable many times.

SUMMARY

The final laminated product has all the advantages of the conventional total contact seat with the added advantage of cost reduction and time savings.

As in any other method used for vacuum forming total contact seating, this method involves the outlay necessary for vacuum forming equipment. There is an initial set-up time involved with making the reusable mold boxes for optimized interfacing. Once the preliminary work has been made in terms of mold preparation, the processing moves very quickly and produces an excellent body support system. This method can be used on adults or children with the main constraint being the platen size available on the vacuum forming machine. The walnut shells are relatively inexpensive as is the latex sheeting, therefore, a large mold size is not prohibitively costly.

This technique has been mainly utilized for fabrication of seating and positioning devices. However, the reusable casting technique should have many applications, such as taking impressions of various body extremities or their remaining stumps. Walnut shells were used because they were inexpensive, easily available, and interlocking, however other material such as aluminum spheres, firm plastic beads, or other small, firm media might be used for the positive cast, depending on the texture, amount of detail, or firmness desired in the positive impression. This method also eliminates some of the mess associated with plaster casting as well as the disposal of outdated total body casts. This method does not eliminate the possibility for taking a plaster impression, if for some reason a long-term record was needed.

Overall, this technique represents a quick, neat, inexpensive way to obtain a solid impression for subsequent draping or vacuum forming fabrication.

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THE COZY SEAT

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The Cozy Seat is a multi-purpose seat made to support small children, of approximately 1 - 3 years of age, who need special seating. Their diagnoses may include cerebral palsy, seizure disorders, and developmental delay. The seat can be used as an insert in strollers, tiny tot wheelchairs, high chairs, car seats and toboggans and may also stand free on the floor. It is easily transferred from one base to another and therefore supports the child in most of his daily activities.

This paper describes the development of the Cozy Seat and its application; patient selection, fitting, use, and follow-up.

INTRODUCTION

To meet the various seating needs for young handicapped children has been a special concern to us over the last few years. Earlier referral to therapy programs and Special Devices Clinics resulted in the demand for technical aids to help these children in their development, give them support and protection and help the people caring for them. A young child uses several seats daily for different activities. We were asked to adapt highchairs, strollers, car seats, and other infant seats for those children who could not use commercially available seats. A "normal" child spends a lot of time sitting on the floor, a handicapped child cannot do so without special support. In trying to serve these children and to offer them proper postural support for the activities expected from them, our design objectives became evident. Using the information we received from the children's parents, therapists and other people caring for them, we defined seating goals and design objectives. A team, headed by a physician and including engineers, therapists, parents, social workers and nurses, started to collect measurements and other pertinent information for the development of one seat which could meet all the seating needs of a young handicapped child.

Design Objectives

Our aim was to produce a seat which:

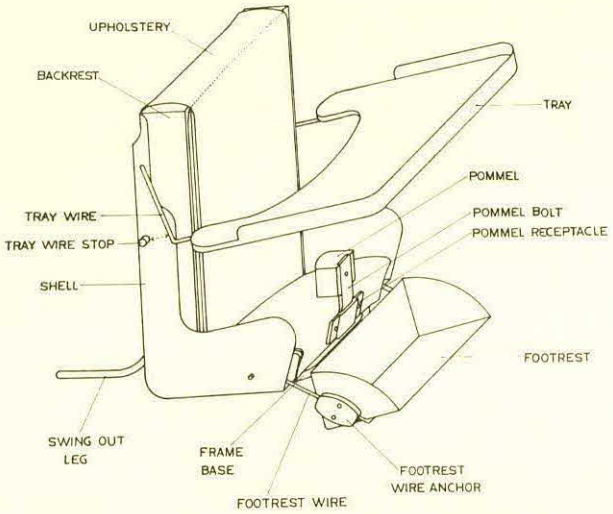
1. is multi-functional, meaning it can be used in different bases, mainly a highchair, stroller, car seat. It should also be safe standing on the floor.
2. is adjustable reclineable from 90^{o} upright sitting position to 45^{o} reclined.
3. is easy to handle, easy to clean, attractive, and durable.
4. provides good postural support and comfort.
5. allows for growth.
6. allows for the addition of bolsters and accessories, such as: neck rests, head rests, pommels, foot rests, and a tray.
7. is easily transferred from one base to another.

Therapeutic Considerations

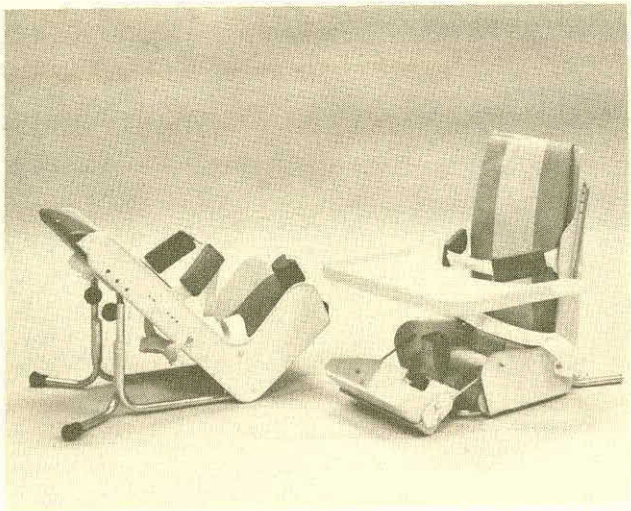
The seat should support the child without restricting him, give him good midline orientation, aid the development of head and trunk control, and put him in a position to facilitate hand function. To place the child in the optimum position for feeding skills was also a consideration.

In March 1978, the first Cozy Seat was fitted and supplied to a multi-handicapped child. He used it in his highchair, stroller, on the floor and later in a Tiny Tot wheelchair. The Cozy Seat enabled this particular child to sit and was a great help to his mother as it freed her hands for feeding him without having to hold and support him. She could also take him for walks which previously had been impossible, because he could not sit in any stroller or carriage.

Since that time some changes were made to the original design relying on feedback from parents and therapists, who worked with children using a Cozy Seat.

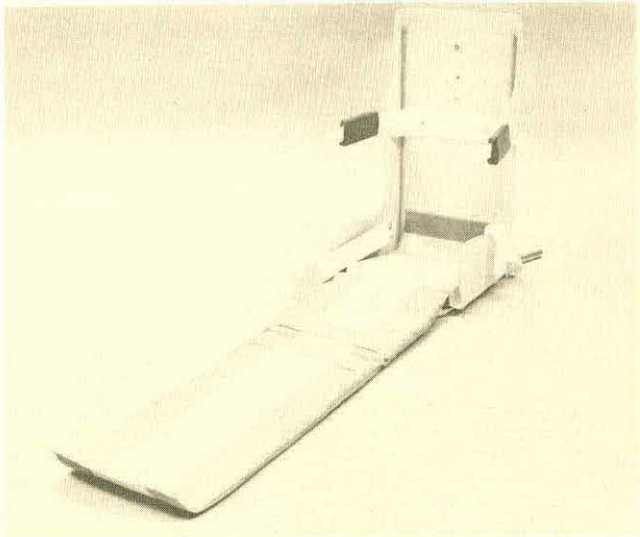


The sketch illustrates the components of the seat as we now use it, and some accessories. The seat shell and back insert (backrest) are vacuum formed from ABS plastic. The upholstery consists of a Velcro attached liner padded with 3/4 inch foam. The liner can be made from Vinyl or cloth; it is detachable for ease in cleaning. A telescoping steel tubing frame supports the seat, it allows the seat to be reclined as desired. Swing out legs give the seat stability for free-standing. Foot rest and tray are also made from ABS plastic.



PATIENT SELECTION AND FITTING

Most of our patients receiving a Cozy Seat are referred to the Special Devices Clinic by their doctor, therapist or social workers. They should be prepared for sitting by a regular therapy program and show some sign of readiness to sit. The physician in charge of the Clinic examines the child and prescribes the seat if appropriate. A thorough therapy assessment is presented at this time. The Clinic team, including the child's parents, discuss the seating needs of the child and define the goals. The child is consequently called in for fitting with the Cozy Seat.



Fitting

1. The child is placed in the seat and the required seat length is determined. Knees are bent and should rest comfortably over the front seat edge. If necessary, the back insert (backrest) is trimmed to size or removed.

2. Hip flexion angle is determined, the seat allows for 90^{o} hip flexion. If more or less is required the backrest is trimmed accordingly, or a wedge is inserted behind the back upholstery.

3. If a lumbar bar is desired this is inserted behind the back upholstery at this time.

4. A seat belt is attached at 45^{o} to the seat.

5. If the child needs a pommel for abduction and/or to prevent sliding forward, this is attached.

Note. If at this time more hip flexion is desired, a foam roll or wedge can be put under the upholstery in front of the seat.

6. Position and type of lateral trunk bolsters are determined. These bolsters are custom made or are pre-foamed from polyurethane with either a mild steel (more rigid) or polyethylene (more flexible) core. They attach to the sides of the seat shell and are connected by a small belt.

The seat is then put into the position in which the child will use it most of the time (upright or reclined). Tray height and position are then determined and the attachment points are marked. The tray adjusts in and out, and allowances for small changes of tray position are built in.

This concludes a fitting session for mildly to moderately involved children. However, the following accessories are available and can be attached to the Cozy Seat, if necessary:

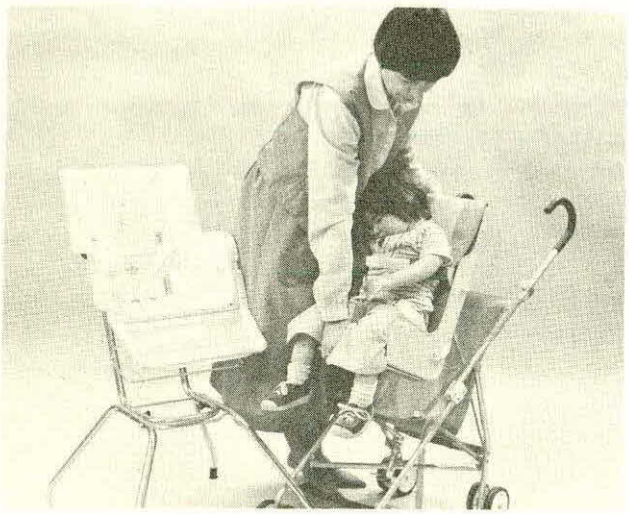
1. pre-foamed headrests.

2. pre-foamed neckrests.

3. small wedges may be cut from foam, and placed behind the back upholstery in the head region to aid midline orientation.

4. removeable footrests.

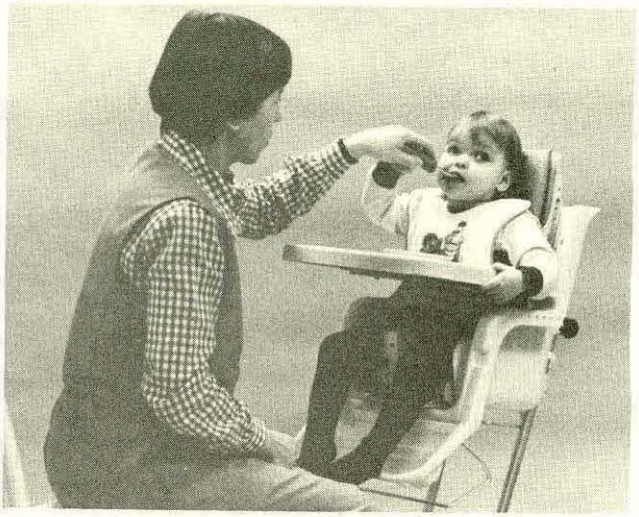
Note referring to growth allowance. More seat length can be achieved by trimming and eventually removing the back insert (backrest). More back height can be achieved by adding a headrest to the top of the back part. After fitting, the child should be observed in the seat for a while, and changes made, if necessary.



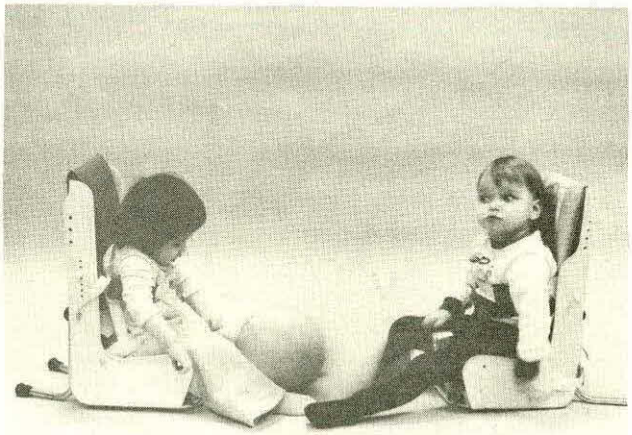
To conclude the fitting session the Cozy Seat is tried in the different bases it is going to be used in. A belt with Velcro closing is attached to the back of the shell. This is usually all that is necessary to hold the seat in the different bases. If the base is a car seat, the original car seat harnessing is used over the Cozy Seat.

USE

The use of the Cozy Seat is demonstrated to the parents and/or the persons caring for the child. The parents are given simple written instructions regarding the proper use of the seat.



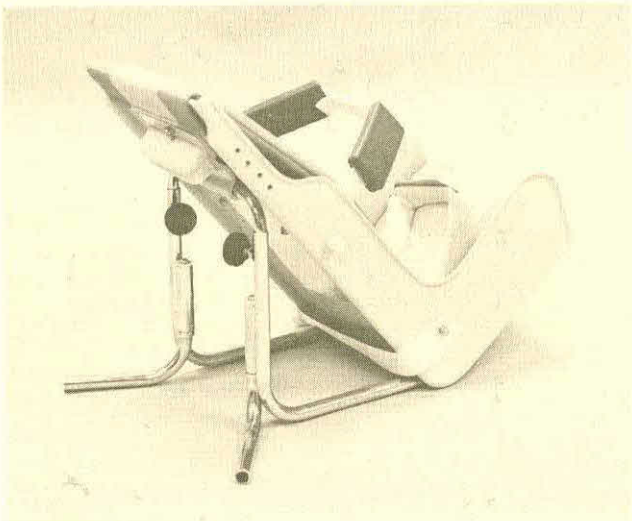
Parents are encouraged to use the seat for feeding. They can adjust it to the position best suited for this purpose. They then have their hands free to either feed the child or teach him self-feeding skills. The seat and tray are easy to clean. ABS plastic is waterproof. The liner can be removed for cleaning. The tray has no sharp corners for food to get stuck in.



The use of the seat on the floor is encouraged, to enable the child to play at the same level as his peers. In a few cases the seat has tipped forwards on the floor. To remedy this tendency, swing out legs can be attached to the front of the seat to give more stability.

The Cozy Seat is a convenient aid in transportation. As mentioned above, it fits into strollers, wheelchairs, car seats, and toboggans. One father reported that he took his child for bicycle rides, placing the child in the Cozy Seat, then into a carrier on his bicycle.

Therapy programs and parent education should continue. The seat itself should not be considered the final solution to the child's problems. It should be used as prescribed and reviewed at regular intervals.



Follow up

Approximately every 3 to six months the child using a Cozy Seat is re-assessed regarding his progress, and the fit of the seat. Changes and modifications are made if necessary. The parents are encouraged to make suggestions and a questionnaire is filled out. As a result of these follow-up interviews, we have incorporated some extras into the seat, e.g. a carrying handle on the back of the seat shell which makes lifting the seat with the child in it much easier. This again emphasizes the importance of the team approach when dealing with handicapped children.

ECONOMIC CONSIDERATIONS

Relative to the cost of equipment for non-handicapped children, the Cozy Seat is fairly expensive. However, it can be recycled, and we feel that the price is reasonable considering that the seat aids handicapped children in most daily living situations. It is definitely less expensive than altering and modifying several commercially available bases. Commercial manufacturing of the Cozy Seat is a future consideration.

In our experience a child can use his Cozy Seat for about one and one-half to two years. This, of course, depends on several factors, for example at what age the child receives his seat.

If special seating is still required after the child outgrows the Cozy Seat, the smallest size seat and back combination of the modular seating system, also developed at the Rehabilitation Centre for Children, Winnipeg, is usually appropriate.

CONCLUSION

Thirty-four Cozy Seats have been supplied by the Rehabilitation Centre for Children, Winnipeg, in the time from March 1978 till February 1980. In addition to these, ten units are being field-tested in Ontario, Saskatchewan, and Alberta since March 1979. In evaluating the input from the field-testing centres and looking at our follow-up assessments, we feel that we met our design objectives. The Cozy Seat is easily transferable between several commercially available bases and safe in free standing on the floor. Reports from parents are favorable as the seat make life easier for the child and themselves. It is an aid to the parents, mainly in feeding and transporting the child. It ensures a good seated positions for the child and gives him the opportunity to engage in activities without having to use all his energy to remain sitting. Communication and good co-operation between professionals and parents have contributed a great deal to the successful development and use of the Cozy Seat.

ACKNOWLEDGEMENT

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FACTORS INFLUENCING THE DESIGN OF A MODULAR INSERT SYSTEM FOR DISABLED CHILDREN

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REHABILITATION CENTRE FOR CHILDREN, WINNIPEG, MANITOBA

A systematic approach in the design of a comprehensive modular seating system for the physically handicapped child has enabled us to offer a more efficient seating program to our patients. The use of plastic moulded components has eliminated the repetative fabrication of desired features in a seat or back. A system designed to make maximum use of the conventional wheelchair and provide the patient with postural support from childhood to maturity.

INTRODUCTION

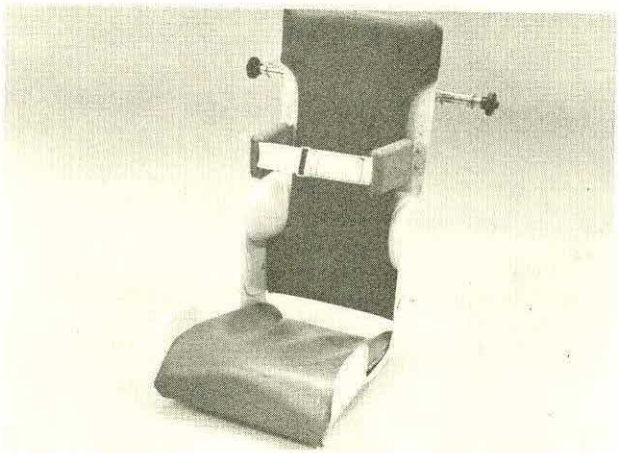
The Rehabilitation Centre for Children is typical of many clinically-oriented service delivery centres in that increasing attention and significance are being given to the postural seating requirements of disabled children and young adults. As the demand for postural seating increased, so too did the need for a systematic approach to the delivery of services.

Reviewing work we had done previously, we noticed that many inserts were similar in style and size, and this lead to the belief that prefabrication of some insert components was possible. It also showed that there was a sizable group of children requiring unique supportive cushions who were not candidates for a modular approach, characteristically, they had severe fixed skeletal deformities, especially scoliosis with accompanying spinal rotation and rib hump, and often as a consequence of severe spasticity and lack of voluntary limb movement.

Thus our approach to improved clinical seating services was two fold: synthesize a versatile modular seating system for the majority of the children, and develop techniques for special components. This paper explores the factors influencing the design of inserts in general, using the modular seating system to illustrate how some of the design problems were overcome.

GENERAL APPROACH

All inserts we provide are based on a common approach. The designed insert must be responsive to the unique individual needs and abilities of the child in total. It is part of a treatment program, and as such is goal-oriented and dynamic. Technical overkill should be avoided.



GOALS FOR SEATING

The reasons for providing a special seating aid for any given child will usually embrace some of the following goals:

- enable the child to sit
- improve posture, control, function, and/or comfort
- provide protection, and/or stimulation
- facilitate easier management of the child
- delay the onset or reduce the severity of orthopedic problems

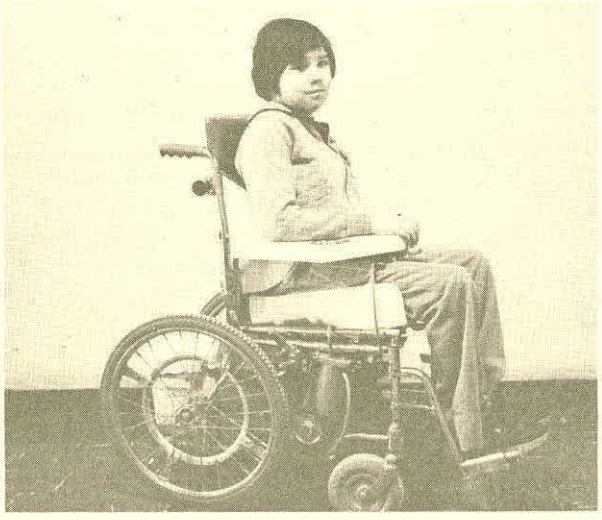
If some of these goals are met, quite often secondary benefits will also result: increased awareness and independence; and improved communication and educational performance by the child.

Priorities for goals will be decided on an individual basis reflecting the child's needs, abilities, and circumstances. Frequently, all goals cannot be achieved fully, and some compromise will be necessary.

INSERT DESIGN FACTORS

1. First and foremost, the insert should be designed and constructed so as to meet the seating objectives established for the child. For most goals, this requires manipulating the child's posture, and the insert must supply appropriate support/restraint to permit the child to maintain the improved posture.

As an example, Clifford C. is a 12 year old boy with Duchenne's-type M.D. At a clinical review it was observed that Clifford would be spending an increasing amount of time in his wheelchair, and that his strength was diminishing. The clinic decided that the principle goals were to maintain his independence by recommending the supply of an electric wheelchair, and to guard against advancement of his scoliosis by providing a thoraco-lumbar lordotic insert. The insert consisted of: an insert to suit the wheelchair, and a lordotic back cushion to accomodate Clifford. A headrest was fitted to give adequate support during transportation.



2. Several secondary factors arise from the introduction of an insert. The insert should avoid additional medical complications, such as chafing at restraint sites, decubitus ulcers, muscular atrophy or contractures, and reduced breathing capacity. It should not bring a diminuation of the child's abilities, such as in transferring or wheeling the chair. The insert should be well-accepted by the child, which will involve factors such as comfort, appearance, and sense of security.

The second case study centres around Betsy R., a 14 year old girl with moderate spastic quadriplegia. Betsy walks with the aid of a special walker, but usually uses a wheelchair in the integrated school she attends. While in her wheelchair Betsy complained of a sore back, resulting from her attempt to compensate for poor balance. The goals in providing an insert for Betsy were to supplement her balance in sitting, and thus relieve the discomfort she was experiencing. A modular back with a single bolster augmented her balance. For a seat, Betsy started with a pommel seat, sized to fit her, because she sat better with the slight pommel. However, she was not able to transfer as well, and was given a transfer seat, which she rejected after a couple of weeks, in favour of the more comfortable and secure pommel-style seat.

3. The insert itself is another piece of equipment to be managed by the parents or attendants of the child. Their acceptance of the insert will be influenced by its size, weight, convenience of use and storage, aesthetics, ease of cleaning and maintenance, as well as its usefulness and the child's reaction. The insert should not be an encumbrance to the handling of the child, as during assisted transfers, feeding, and so forth.

There are several features of the modular inserts aimed at simplifying their management. The cushions are attached by Velcro, and remove easily for cleaning. There are very few places for dirt and grime to collect.

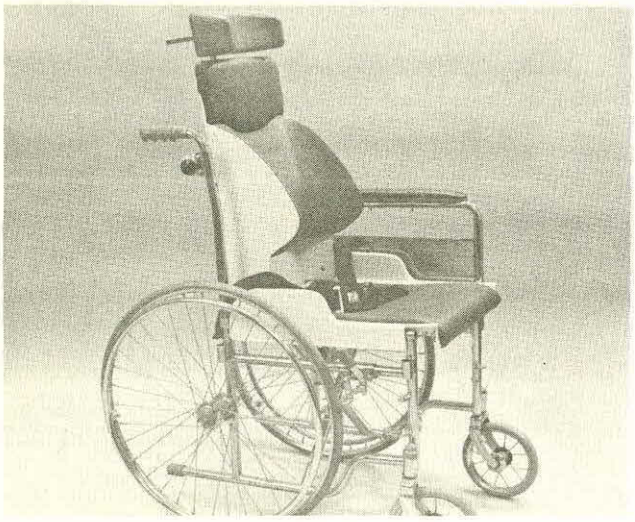
Clifford's seat was covered with a Tee foam cushion (for comfort), wrapped in a plastic bag (for waterproofing), and placed inside a zippered terry cloth cover, (which can be laundered, looks nice, is a breathing material, and low surface tension does not detract from Tee foam qualities.

The patient support shells in the modular insert system are only slightly wider than the patient himself. This efficient use of space reduces the bulk of the insert, and avoids the necessity of an overly large wheelchair which may be too wide for the child to wheel. Parents are instructed in the use of the insert.

4. Safety is another design criterion. The insert should be mechanically sound, non allergenic, chemically inert, fireproof, waterproof, hygienic and cleanable. Note that many of these factors are integral properties of the materials used in construction.

BASES AND INTERFACES

All inserts will require a base of some description. Selection of an appropriate base should consider the child's abilities, the family's lifestyle, and availability of the bases. Economics often dictate the base which will be used, as when an agency or institution has an existing supply of wheelchairs and components. The insert should be easy to mount on the wheeled base, for interfacing, and should not detract from any good features inherent in the base such as smooth ride, foldability, and general mobility. It may be necessary for the insert to interface with several different bases, as the child moves from home to school for example. The entire insert may have to recline on the base, as during transportation or feeding.



Betsy's insert sits on the unmodified wheelchair upholstery, and is secured by a belt around the handle uprights. In Clifford's case, it was not possible to obtain the correct position for the back using only a belt, and cross-tube was used which is bolted to the insert back and snaps into latch blocks on the handle uprights. Clifford's seat was chosen to fit directly over the seat tubes of his wheelchair, from which all upholstering was removed.

CONCLUSION

As a result of working with numerous patients such as Betsy and Clifford, we have developed over twenty four months a range of seat and back molds, with complimentary hardware for field testing. During the course of the field test several small mold modifications were made, and on conclusion of the test in two months time a total review of the system will take place.

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ACKNOWLEDGEMENT

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IDENTIFICATION AND CONTROL OF BIOPHYSICAL FACTORS RESPONSIBLE FOR SOFT TISSUE BREAKDOWN

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ABSTRACT: A major research project aimed at prevention of pressure sores is in progress at Helen Hayes Hospital. These investigations include laboratory and clinical measurements together with computer modeling to identify quantitatively the key factors responsible for soft tissue breakdown. Two approaches, "Detection" and "Prediction", are being developed toward the goal of providing early clinical warnings of incipient or actual tissue damage. Detection methods will rely on non-invasive, direct measurements of changes in properties of superficial and deep patient tissues. In one case, mechanical measurement devices will be employed, in another determinations will be made by ultrasonic backscatter and attenuation. Prediction involves studies leading to computer modeling of stresses in patient skin, fat and muscle during seating and other loading conditions. Dangerous stresses will be identified with respect to magnitude-duration curves for tissue damage with direct and shear stresses which are being established in a newly developed animal model.

INTRODUCTION

Clinical problems related to the effects of mechanical pressure on human tissue represent one of the greatest challenges in rehabilitation medicine. Patients with spinal cord discontinuity and dysfunction, peripheral vascular disease, diabetes, amputations, leprosy, collagen disorders, loss of consciousness, debilitating diseases, geriatric problems, and other conditions all are susceptible to soft tissue damage and breakdown from excessive mechanical pressure. Based on the classic work of Kosiak (1), Lindan (2), and others, it is assumed that damage is done when external pressure causes ischemia to an area for an excessive period, but there are many unknown factors with regard to both the mechanics of injury and to tissue factors which predispose to breakdown.

In the Biomechanics Research Unit at Helen Hayes Hospital, a Soft Tissue Mechanics Research Group has been formed in association with the Center for Biomedical Engineering at Rensselaer Polytechnic Institute. The purpose is to investigate the role of biophysical factors in soft tissue breakdown with the long range goal of devising practical clinical tests to provide early warning of incipient tissue damage.

In the initial research plan, two approaches leading toward this goal were visualized. The first approach utilized measurement of physical parameters, including environmental conditions at the interface between skin and supporting surface, to detect when dangerous conditions predisposing to pressure sores occur. The second involved use of computer modeling techniques to predict when dangerous conditions occur. This paper will describe progress, modifications and findings in the investigation of these approaches during the past two years.

METHODS/RESULTS

Detection Approach

Environmental Factors. At the skin/cushion or other interface, it is possible to measure temperature, humidity, heat flux and pressure as studied in detail on a sample of 24 wheelchair cushions by the Soft Tissue Mechanics Research Group (3,4). To date, no satisfactory method of measuring shear has been devised although the authors have reviewed this problem extensively. While clear differences in the measured parameters were seen among the various classes of cushion (i.e. foam, gel, etc.), no cushions were satisfactory in all aspects. Each type has its own major advantages and many commercial units are poorly designed, particularly with respect to covers (3,4). The general impression of this work was that for patients with critical problems, no cushion can be selected off the shelf, a presciption based on known cushion properties appropriate for the patient should be made, then individual fitting should be done ideally while monitoring the above parameters to obtain optimal levels. To take advantage of the best attributes of several types of materials, a multilayer combination cushion within a special cover has been proposed by the authors (3).

To aid in cushion fitting procedures, in particular the modification of foam cushions, members of the Soft Tissue Mechanics Research Group have devised a portable "wheelchair barograph". Patients sitting in this device produce a light pattern of high pressure areas that can be marked directly on a plastic sheet for use as an overlay to locate areas for modification on the cushion. This is a "low-technology", non-electronic device

intended for clinical use (Fig. 1).

Pressure measurements are useful primarily in these custom fitting procedures when direct comparison can be made between cushions of the same type. For foam cushions, measurements are of particular value during the various wedging and cut-out procedures (5). Diaphragm pressure transducers are expensive, difficult to use and subject to a variety of errors (6). The authors have found that the "air-cell" type of transducer is more consistent and satisfactory in clinical use (7), and these transducers can be used to provide data at multiple sites during one "sitting" (8,9).

In an attempt to determine the accuracy of interface pressure measurements in absolute terms, the authors have applied controlled pressures to various transducers between layers of different foam and gel materials. The results have been discouraging as wide errors are present. The causative factors are related to the degree of envelopment of the transducer as well as diaphragm contact and shear in the case of electronic transducers. Consequently, while the measurements are useful for relative pressures, it is difficult to define the actual pressure at the skin/cushion interface. It is best to use pressure measurements only as an indicator of relative changes in condition.

In an effort to identify what interface pressure means in terms of tissue pressure, the authors are employing subcutaneous wick-catheter measurements in the posterior thighs of human subjects, while sitting on a firm surface. Control measurements beneath the same area are made with air cell transducers. Initial results suggest that the pressure measured by the air cell reflects the pressure as measured by a wick catheter through which a small amount of fluid has been injected. Some question remains as to whether the wick catheter measures the interstitial or total tissue pressure under these conditions (10) but the technique was validated in animal experiments in this laboratory (11).

In summary, it appears that measurement of environmental factors at the skin surface can indicate when certain undesirable conditions exist and can aid in modifications to produce improvements, but there are many variables and it may be impractical to translate these measurements into a system capable of detecting incipient tissue damage.

Tissue Factors. As a result of the initial investigation, it seems that a more promising approach would be to monitor changes in patient tissue directly for comparison to normals, and/or changes detected in control patients prior to a breakdown. Two non-invasive mechanical devices have been developed for this purpose and are currently being tested. One device is a sensitive skin extensometer suggested by the work of Alexander and Cook (12). The other device provides a light indenting force over a bony prominence to produce a characteristic of the underlying tissue which reflects compressive and viscoelastic properties. Results of similar tests are known to change with age and tests are in progress to determine if changes in these measurements occur over time in spinal cord injury and other patients subject to pressure sores.

A third, more indirect method under development by this Group will utilize ultrasonic attenuation and backscatter measurements to detect early changes in deeper tissue layers that result in slight alterations in mechanical properties. Already utilized by others to characterize and differentiate normal and pathologic tissue in myocardial infarction (13), this sophisticated, but simply applied technique probably offers the best chance of obtaining a direct, clinical, early warning system to detect incipient decubitus ulcers in time to take corrective action.

Prediction Approach

At present, the authors are engaged in work leading to development of a 3-D finite element-model of buttock/cushion interaction to provide information on stress states in the skin, fat and muscle in specific subjects. Simultaneously, animal work in progress is aimed at defining, for the first time, the sensitivity of normal tissue to actual internal stresses in the three tissue layers. Prior work on stress-duration curves (1,2) relates only to the externally applied pressure, not to calculated stresses within tissue.

Animal Experiments. The Soft Tissue Mechanics Research Group has developed a reliable animal model which facilitates the investigation of the effects of various parameters on decubitus ulcer formation. The animal model is based on indentation of selected sites of anesthetized Yorkshire pigs weighing 20-30 kg using a specially designed six head loading system with two different shapes of indentors. Loads are applied through pneumatically actuated pistons connected to the indenting heads. Precision pressure regulators are used to supply compressed air to the cylinders. Although Dinsdale (13) used pneumatically actuated pistons, he did not regulate the pressure precisely, and pressure fluctuated by 50-150% from the beginning to the end of the experiment. Reliability of the present system was established by electronically monitoring the regulated air pressure and by calibrating the applied load with a load cell. After experimentation at a number of sites, the following sites have been found to provide good underlying bone support and yield reliable results bilaterally: a) infraspinous fossa of the scapula, b) transverse process and 12th ribs 4.5 cm lateral to the midline and c) greater trochanter.

The animal model is being used to study the threshold stress-duration relationships in pressure induced tissue damage. Two cm diameter hemispherical or flat circular indentors are used to indent at the above sites bilaterally for periods up to 12 hours. The two types of indentors are designed to create different stress fields in terms of direct and shear components: the hemispherical

indentor provides relatively large shear stresses whereas the flat circular indentor causes relatively large compressive stresses in the tissue. Animals are sacrificed on the eighth day following indentation and tissue damage assessed histologically. Load duration curves for each site for each of the types of indentors are being established in relation to tissue damage. Early results indicate that the damage produced by hemispherical indentors is often initiated in the muscle layer adjacent to bone. The damage produced by flat circular indentor is often initiated in the skin. Simultaneously, stress fields beneath each type of indentors are being determined by computer modeling taking into account finite deformation and non-linear characteristics. Computer results will be validated by dimensional studies on frozen segments of the indented tissue. The ultimate result of this work will be stress magnitude-duration curves for direct and shear stresses for skin fat and muscle for normal pig tissue. This data will provide a starting point to develop indexes for tolerance levels for normal and abnormal human tissue.

Modeling of Buttock-Cushion Interactions. Computer modeling of stresses in human buttock represents another complex task with the same problems of large deformations and non-linear characteristics found in the pig indentation. Chow (14, 15) has accomplished initial physical and mathematical modeling in this area. In the present study, a 2-D physical (gel) model of the buttock was developed and used to measure direct and shear strains in the tissue as a function of various types and thicknesses of cushion support surfaces under vertical loading (Fig. 2). It was found that maximum compressive stresses, for all cushion cases, tend to concentrate near the "bone". Maximum shear stresses occur near the skin for all cushion cases except with stiff foams. On a stiffer foam, maximum shear stress location becomes deeper. Gel cushions produce high compressive and shear stress under direct loading. A Medium density foam cushion causes lower compressive and shear stresses on the buttock. In addition, it was found that thick cushions distribute the stresses more uniformly. The present results confirm and extend Chow's findings (14) that maximum compressive stress in general occurs along the load axis adjacent to bone and that the compressive stress decreases to a lower level near the buttock-cushion interface. In realistic situations, horizontal loads are often encountered, and gel cushions are thought to reduce shear under these conditions. Work to study this effect in the 2-D model is planned.

In order to facilitate studying of the complex load conditions, a finite element model of buttock cushion interaction is being developed. As the first step, the 2-D physical model will be simulated on the computer to validate the model and check programming, then the work will be extended to a 3-D network, as another step toward 3-D patient modeling. For patient determinations, actual data on buttock dimensions and tissue thicknesses (determined ultrasonically) will be inputed to the computer together with information on cushion characteristics. In this manner, it may be possible to predict the presence of dangerous stresses in advance and plan appropriate corrective action.

SIGNIFICANCE

The current project is resulting in several R & D products of immediate or potential benefit to handicapped persons susceptible to pressure induced damage of soft tissues.

(1) improved guidelines and techniques for measurement of parameters that aid in prescription and fitting of wheelchair cushions.
(2) a set of three noninvasive clinical tests to detect when mechanical changes that precede pressure sores are present in patient tissues.
(3) a measurement system and computer program capable of calculating stresses in patient tissues to predict when dangerous stresses may be present (based on animal studies) and aid in prescribing steps to reduce the stresses,
(4) to further research, a new animal (porcine) model for production of experimental pressure sores, and stress-duration curves for damage to normal tissues has been developed. (tests with the same model using spinal cord injured animals are feasible but beyond the scope of this project).

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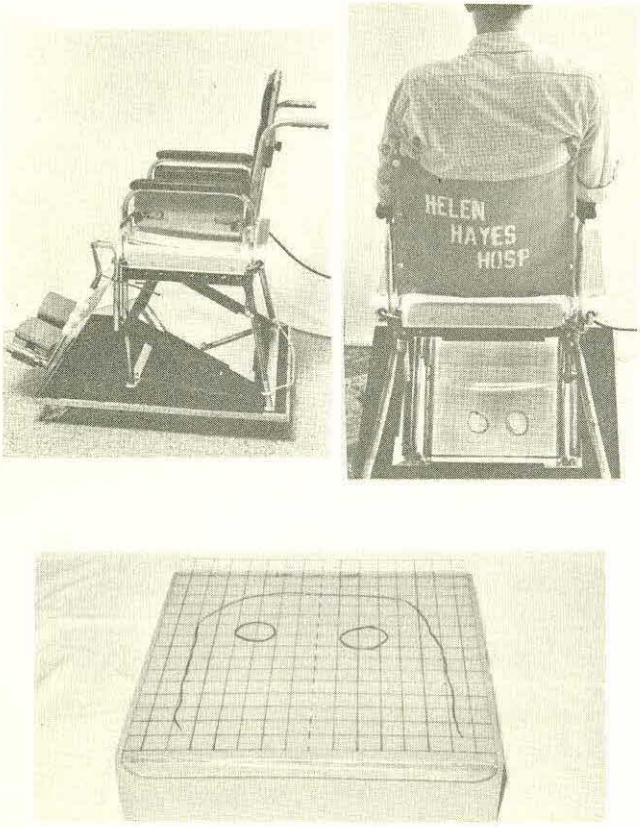

Figure 1. Portable Wheelchair Barograph with Overlay Used for Marking Cushion

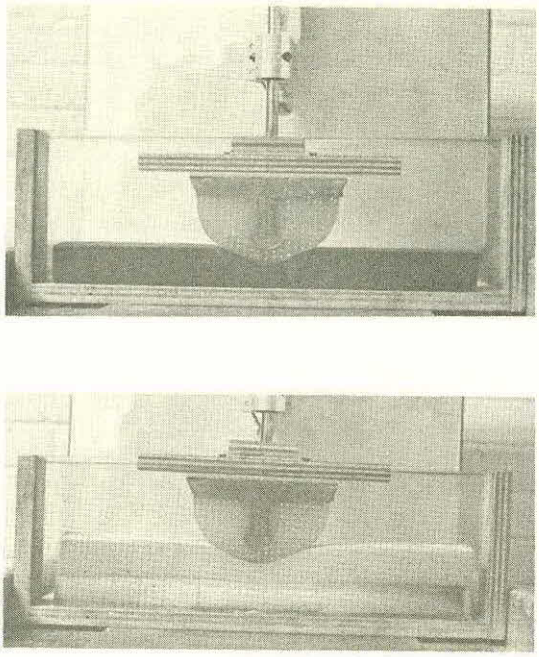

Figure 2. 2-D Models of Buttock-Cushion Interactions Under Vertical Loading

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PRESSURE SORE PREVENTION FOR THE WHEELCHAIR USER

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ABSTRACT

The concept of a wheelchair cushion fitting clinic for patients vulnerable to pressure sores was developed initially by Rogers at Rancho Los Amigos Hospital. The authors have developed the concept further and have monitored results for a group of approximately 150 spinal injury patients over a period of 4 years. Details of patient assessment, fitting, cushion manufacturing techniques, associated documentation, and pressure sore incidence are presented.

The cushion fitting clinic has been extended recently in order to encompass other diagnostic groups and support situations. A Pressure Sore Prevention Unit provides a focal point for information, advice and expertise for patients and personnel responsible for their care. The organisation and methods used in running this unit are described.

INTRODUCTION

Two surveys (1, 2, 3, 4) of pressure sore prevalence conducted in the Greater Glasgow and Borders Area Health Boards of Scotland have indicated that on the day surveyed, between 9% and 11% of patients in hospital and in the community receiving professional nursing care had pressure sores. Of these patients, approximately 15% had pressure sores in tissues overlying the ischial tuberosities. It may be assumed that the majority of these ischial pressure sores were developed by patients when seated in rest chairs, wheelchairs or in car-seats.

Spinal injury patients are a group particularly vulnerable to developing pressure sores, especially over the ischial tuberosities due to the prolonged periods of time they spend sitting with absent sensation, and paralysis below the level of spinal lesion. For the majority of these patients the wheelchair cushion is a vitally important means of modifying the conditions which produce pressure sores.

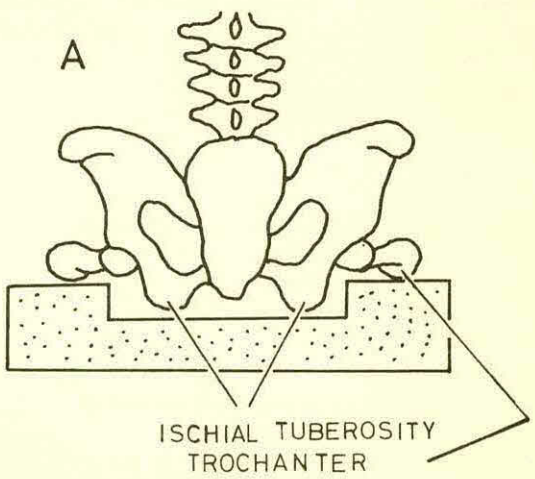
Most commercial wheelchair cushions are designed for patients with complete sensation. They are designed for comfort, and as a work surface from which the seated patient can operate. In addition to comfort, ease of transfer, dynamic properties, and ease of storage when the chair is collapsed, are major considerations in the design. A number of cushions are now available for patients whose primary requirement is for a support surface that helps them to prevent pressure sores from developing. Although many patients successfully avoid developing pressure sores on these cushions, an unacceptably large group continue to develop these distressing, expensive, and dangerous lesions.

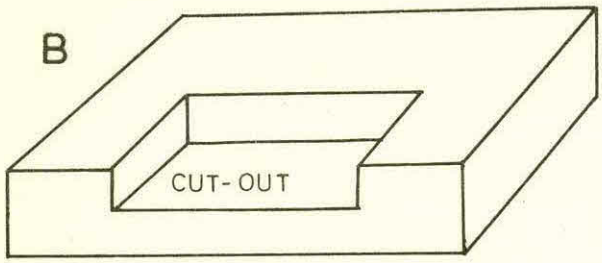
THE PRESSURE CLINIC

An approach to the problem, developed initially by Rogers (5), emphasises the need to accommodate patients' individual support requirements. Roger's 'Pressure Clinic' provides the facilities to supply and monitor wheelchair cushions shaped to reduce the pressure beneath the ischial tuberosities to a 'safe' level. This is achieved by removing a section from a suitable block of high density expanded polyurethane foam (figure 1).

Figure 1

Diagrammatic representation of pressure relief provided by a cut-out of foam beneath the ischial tuberosities of the patient. (A) rear view, (B) oblique view.





The majority of the patient's body weight that is normally borne by the ischial tuberosities is transferred to the trochanteric shelf and proximal thighs. These regions of the body have a relatively large surface area and are therefore capable of distributing the transferred body weight with only a modest increase in pressure.

Measurements of pressure are made using a simple-to-use 'Pressure Evaluator' (5) which is now available commercially (Talley, U.K.; Scimedics, U.S.A.). One of the cushion fitting criteria established by Rogers (5) was that for the cushion to be effective in preventing pressure sores over the ischial tuberosities, the local pressure should not exceed 30 mm Hg (4.0 kPa). This requirement was determined on the basis that pressures in excess of capillary pressure (6) are likely to affect skin nutrition and render it vulnerable to breakdown.

The majority of wheelchair cushions have limited durability. Many patients sit for over 12 hours each day and consequently cushion materials fatigue, and covers deteriorate. In addition the patient's requirements may change if his wheelchair is replaced or there are changes in his posture or body weight. In order to ensure that the wheelchair cushion provides adequate support in the long term, it is necessary for the patient to return to the clinic regularly. The majority of the patients vulnerable to pressure sores require protection from adverse support conditions for life. The pressure clinic is therefore seen to be a long term commitment requiring a reliable allocation of resources.

Results from Rancho Pressure Clinic

In collaboration with the Rehabilitation Engineering Centre at Rancho Los Amigos Hospital the authors have analysed results from 657 patients' records collected between 1973 and 1975 from the Rancho Pressure Clinic (7, 8, 9). A correlation of the pressure beneath the ischial tuberosities and the patient's skin condition at this site (Table 1) indicates a break-point at about 40 mm Hg (5.3 kPa) where the proportion of patients with ischial pressure sores or skin redness begins to increase rapidly. Patients with ischial pressures greater than 70 mm Hg (9.3 kPa) appear to be 5 times more likely to have an ischial pressure sore than those with pressures less than 40 mm Hg (5.3 kPa).

Table 1

Correlation of skin condition versus pressure beneath the ischial tuberosities of spinal injury patients attending the Rancho Pressure Clinic. Numerals in parentheses are percentage of total.

Pressure mm Hg	Patients with pressure sores	Patients with skin redness	Total number of patients measured
0-10	0 (0)	1 (-)	2
11-20	1 (3.3)	2 (6.7)	30
21-30	2 (3.2)	3 (4.8)	63
31-40	4 (3.9)	4 (3.9)	103
41-50	7 (5.7)	18 (14.7)	122
51-60	5 (5.7)	18 (20.4)	88
61-70	6 (7.1)	22 (25.9)	85
70	32 (19.5)	32 (19.5)	164
Total	57 (8.7)	100 (15.2)	657

The average pressure measured beneath the ischial tuberosities of patients fitted with the Pressure Clinic cushion was significantly lower ($P < 0.001$) than that developed by patients using a variety of commercially produced cushions, some of which were specifically designed to prevent pressure sores. There was a correspondingly lower proportion of patients with ischial pressure sores using the Pressure Clinic cushion at their first clinic check-up compared with patients sitting on alternative cushions (8). These results were sufficiently encouraging for the authors to establish a similar pressure clinic for spinal injury patients at Philipshill Hospital, Glasgow, U.K.

DEVELOPMENT OF PRESSURE CLINIC

Early attempts by the authors to duplicate the procedures outlined by Rogers (10) were complicated by local factors, in particular the availability of suitable materials. Experience has also identified the need for a number of design options in addition to those of the 'ischial cut-out' and 'pre-ischial bar' recommended by Rogers (10). These additional options have necessitated accurate documentation of patient details and cushion design. It has also proved necessary to develop a systematic patient assessment procedure in order to conserve materials and limit the number of cushion changes during assessment and to prevent the patient from becoming excessively tired.

Cushion Design Options

Expanded polyurethane foam, cut to the size of the wheelchair seat, 75 mm or 100 mm thick, and of alternative densities (29-31 kgm^{-3} and 38-42 kgm^{-3}) are used. In general, patients with some sensation in the gluteal region prefer the lower density foam. Considerable difficulty has been encountered due to sag in the sling seat of the wheelchair. The sag caused the top surface of the cushion to become concave resulting in knee adduction and poor location of the ischial tuberosities in the cut-out. A correspondingly convex base formed from 'Plastazote' fitted to the bottom of the cushion has reduced these

problems considerably.

Patients who have difficulty in transferring from their wheelchair to other supports often find a cushion with an anterior slope reduces their difficulties. Patients with poor catheter drainage when sitting occasionally find this modification an advantage.

The dimensions of the cut-out in the foam beneath the ischial tuberosities are important if maximum load transfer to the trochanters and proximal thighs is to be achieved. The inter-ischial distance may be measured using a thin latex bag filled with small polystyrene beads. The bag is placed beneath the patient and located accurately against the back of the wheelchair thus providing a reference position. The air in the bag is removed using a bicycle pump, a firm mould thereby being formed. The indentation of the ischii can usually be seen without difficulty once the mould has been removed from beneath the patient. The cut-out in the foam is normally made 30 mm to either side of the ischii and 40 mm in front of them. A typical cut-out would be 200 mm wide and 180 mm long. The depth of the cut-out is not permitted to exceed half the thickness of the cushion. If the patient has vulnerable tissue covering the trochanters, or has a severe lordosis, an under-cut cut-out may be preferable to one on the top surface of the cushion.

A cut-out is normally made in the Plastazote base of the cushion, its dimensions corresponding to those of the cut-out in the foam. Occasionally if only modest pressure relief is required this cut-out may be the only one required.

The authors are usually reluctant to provide a pre-ischial bar of foam anterior to the cut-out as described by Rogers (10) as this may result in catheter blockage and possibly affect venous drainage of the legs. If ischial pressure relief cannot be achieved satisfactorily using other techniques or if the patient requires a fulcrum to rock forward on, then a pre-ischial bar 100 mm deep x 50 mm thick placed across the width of the cushion may be supplied.

Cushion Cover Material

Two materials are currently being used by the authors, the choice depending upon the reflex sweating and incontinence containment status of the patient. Both materials are biaxially stretchable and thereby maintain the advantageous mechanical properties of the foam component of the cushion (11, 12, 13). One material is water permeable (Suprima Textiles U.K. No. MNL 1228), the alternative being a waterproof woven nylon fabric (Clutsom-Penn, U.K. No. 31902) coated with polyurethane. The majority of the patients are fitted with cushions covered by the waterproof material but with the non-coated surface closest to the skin. Some patients prefer the smooth coated surface uppermost in order to lower the coefficient of friction which aids transferring. For patients who present a negligible urine spillage risk, or who endure substantial reflex sweating, the non-waterproof material is preferred.

Two strips of 'Velcro' are attached to the cover, on the underside of the cushion. Corresponding strips are attached to the wheelchair sling-seat, and thereby ensure accurate location of the cushion in the chair. Patients with severe spasm or lordosis often displace their cushion forwards. If the precaution of fitting 'Velcro' is not taken, accurate positioning of the ischial tuberosities relative to the cut-out cannot be guaranteed.

Patient Assessment

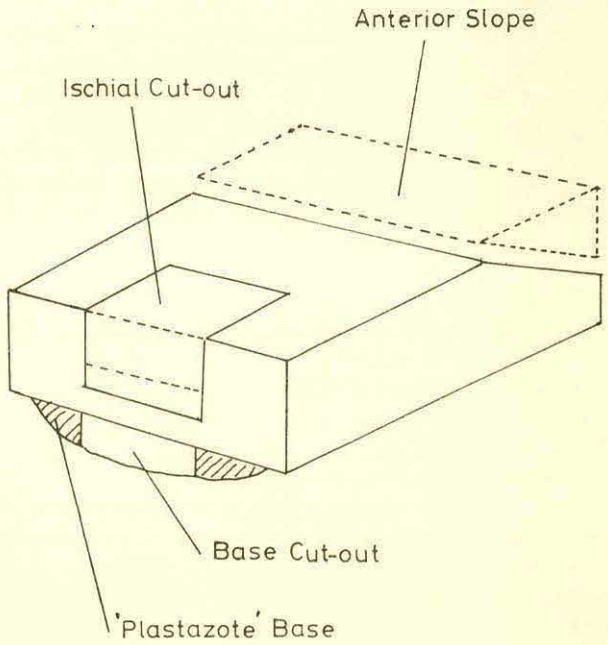
Assuming a regular pressure relief regime (the adoption of which is strongly dependent upon patient motivation and the quality of post-injury rehabilitation) then the key variable to control is 'pressure' (14) in the tissues beneath the ischial tuberosities. The objectives established by the authors are to reduce and maintain the pressure beneath the ischial tuberosities, whenever possible to less than 40 mmHg (5.3 kPa) using the simplest combination of cushion design features available. Patients are instructed to sustain 'push-up' regimes, skin inspections etc. and to increase their frequency where pressures greater than 40 mm Hg (5.3 kPa) have to be tolerated.

Patients are referred to the clinic by their G.P. or clinical specialist. Records are kept of the patient's general condition, the type and condition of his wheelchair and wheelchair cushion, and of his skin condition. Wheelchair faults such as worn back supports, faulty brakes etc. are recorded and action initiated. Measurements are made of the pressure beneath the ischial tuberosities, sacrum, coccyx, and trochanters in order to determine whether the patient's existing cushion is adequate. If the sub-ischial or sacral pressures exceed 40 mm Hg (5.3 kPa), 60 mm Hg (8.0 kPa) at the trochanters or 30 mm Hg (4.0 kPa) over the coccyx, then the patient's cushion is replaced.

An assessment cushion (figure 2) has been designed to enable a series of design options to be tried before making a specific cushion for the patient.

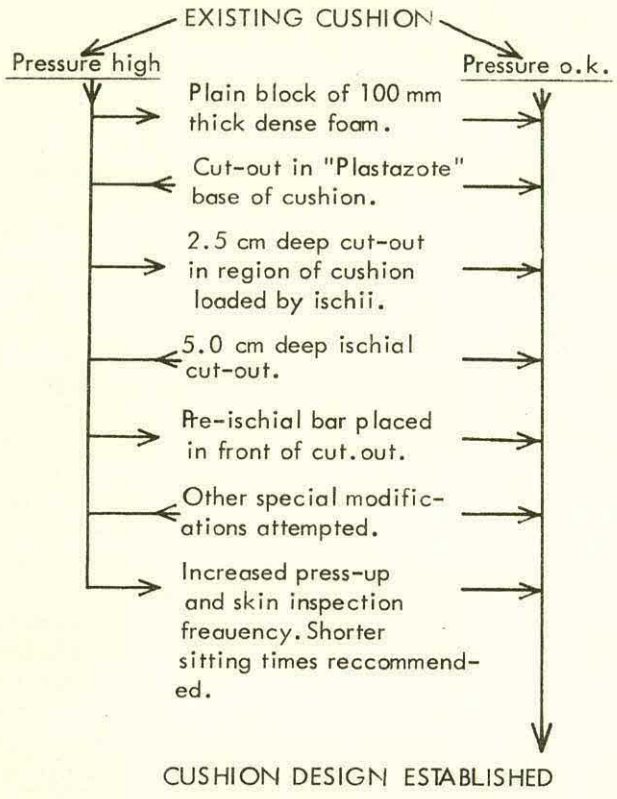
Figure 2

Design options available on assessment cushion.



The assessment cushion consists of a block of polyurethane foam of appropriate density with removable sections which when appropriately selected give a range of cut-out depths and the option of an anterior slope.

The following sequence is followed in trying to establish an effective design for the patient.



If a cut-out is indicated then the inter-ischial distance is measured as formerly described in order to determine its dimensions.

Once acceptable conditions have been established, the dimensions and design of the cushion are transferred to an order form which is then sent to a local contractor who then manufactures the cushion to specification.

Patients are recalled at three-monthly intervals to enable cushion wear and pressures to be monitored. If pressure levels deteriorate the cushion is redesigned or replaced as appropriate.

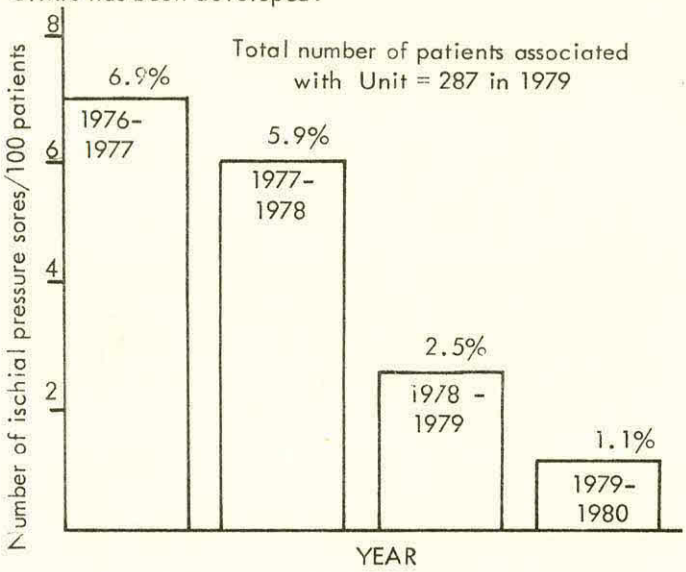
RESULTS

Since establishing their Pressure Clinic in Glasgow the authors have monitored patient admissions to the Philipshill Spinal Injuries Unit. A measure of the effectiveness of the procedures developed in the Pressure Clinic has been obtained from these statistics and provides a stimulus for its continuation.

Since the Pressure Clinic was established in 1976 the patients admitted to the Spinal Unit with pressure sores fell from 10.0% to 3.7% of the total number of patients associated with the Unit. The proportion of ischial pressure sores compared with other sores for which patients were admitted fell from 68% to 30%. The number of patient days per year per patient associated with the Spinal Unit, required for the treatment of ischial pressure sores fell from 4.8 in 1976 to 2.3 in 1979. Figure 3 indicates the reduction in incidence of ischial pressure sores during the time that the Pressure Clinic has been in existence.

Figure 3

Reduction in incidence of ischial pressure sores in spinal injury patients during the period in which the Pressure Clinic has been developed.



The reduction in ischial pressure sore incidence appears to correspond directly with the introduction of the Pressure Clinic. The benefits are substantial and appear to be sustained. No other changes in patient care techniques or policy are known which could account for these results and the incidence of pressure sores on sites unaffected by sitting has remained relatively stable.

Development of Pressure Sore Prevention Unit

As a direct consequence of the promising results obtained from the Pressure Clinic, a unit has now been established at Philipshill Hospital which provides the services of the Pressure Clinic as a routine facility rather than a research programme and in addition forms a focus for education in pressure sore prevention techniques, information, and advice to both nurses and patients. The Pressure Sore Prevention Unit is staffed by two nursing sisters, both with extensive community and hospital nursing experience and caters for in excess of 1000 outpatients and community visits per year.

CONCLUSION

The evidence supporting the principles of local pressure relief by modification of simple foam cushions is most promising. For spinal injury patients undertaking a regular pressure relief regime a 'safe' threshold of 40 mm Hg (5.3 kPa) appears to be effective. The results indicate substantial reductions in pressure sore incidence and hospital bed occupancy and the authors would suggest that the Pressure Clinic concept is highly cost-effective.

Further developments will require to accommodate patients with other disabilities using other support surfaces. The authors have already begun to consider these problems, and have as a first step extended the Pressure Clinic to include broader aspects of pressure sore prevention by providing a centre specialising in instruction, information and advice to patients, nurses, and patients' relatives requiring assistance.

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EFFECTS OF CUSHION MODIFICATION ON PRESSURE DISTRIBUTION

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Wheelchair cushions made of polymer foam are effective in lowering sitting pressures in disabled persons (1). However, no one cushion will work well for all persons and modification of standard cushions may be necessary. "Wedging" is one method of cushion modification and consists of slicing out a wedge from the underside of a block of foam. It is less sensitive to lateral shifting and positioning than cut-outs done on the top surface of cushions. This study investigates the effects of depth, location and distance between wedges on magnitude and location of pressure. Results suggest that 1" deep wedges positioned posteriorly frequently reduce peak pressures and total load. Wedges 2" deep were equally effective for light and heavy subjects. Anterior placement of wedges was effective in shifting weight away from high pressure regions in heavier subjects. Wedged cushions were the most effective surfaces for 2/3 of the subjects tested.

INTRODUCTION:

Wheelchair cushions are used by patients with physical disability to reduce pressure and better distribute their weight. This is especially important for those individuals with absent or diminished sensation in the areas most vulnerable to pressure-induced tissue trauma (2-4). There are numerous cushions manufactured with the handicapped population in mind. However, it has been shown (1) that no single cushion is effective in helping to prevent breakdown for all patient groups and that individual evaluation of commercially available devices is necessary to determine the best cushion for each person. There are those times when no standard cushion appears to relieve pressure and it is then that modification of the cushion is necessary in order to provide such relief. Various types of "cut-outs" from the top surface of a cushion have been used for many years as a means of relieving pressure from ischial tuberosities. The major objection to this type of modification necessitates having the patient in the exact same position each time if he is to really attain pressure relief. This is often very difficult especially if the patient is a high-level quadriplegic. The "wedging" process is an alternative to traditional cut-outs. The philosophy of this type of modification is that by cutting the underside of the cushion, there is much less sensitivity to lateral positioning and shifting, thus ensuring that the patient is achieving desired pressure relief.

"Wedging" consists of slicing a wedge from the underside of a block of foam, usually of the polymer type (Figure 1a and b). The objectives of the present study are as follows: 1) to determine the effects of varying the wedge parameters of depth, location and distance between wedges on maximum pressure and pressure distribution; 2) to ascertain the interdependence of wedge parameters and the combined effects on pressure distribution and reduction of peak pressure; 3) to quantitate and compare the effects of wedging and pressure distribution; 4) to correlate body build based on height, weight, age and sex and wedge parameters with pressure distribution.

METHODOLOGY

In this study, cushions of T-41 Temper Foam, 4"x16"x18" were used. Each wedge was made on the underside of the foam by making one vertical and one 45^{o} angle cut into the foam using a standard kitchen electric knife. The wedges are made on the underside of the cushion so that the surface next to the skin remains smooth thus eliminating geometric irregularities at the cushion/tissue interface. A description of the test surfaces can be found in Table 1.

The sitting pressures of each of ten normal subjects were monitored on test surfaces using the Pressure Evaluation Pad developed at the Texas Rehabilitation Engineering Center (1). Each subjects' body build was classified as obese, average, or thin based on height, weight, sex and age (5). Obese is defined as being 10% or more over average weight and thin is defined as being 10% or more under average weight.

RESULTS

Preliminary results are given in terms of hypotheses and findings as follows:

1. Hypothesis: More wedges of lesser depth should result in greater reduction of peak pressure (P_p) and effective total pressure load (P_E) due to increased cushion surface area in contact with tissue.

   Findings: The use of smaller wedges (five 1" wedges) reduced both P_p and P_E with greater frequency than the use of larger wedges (three 2" wedges).

2. Hypothesis: Cushions modified with 2" wedges may be more effective than 1" wedges in lower-

ing P_P and P_E for heavy subjects due to the greater deformation produced by heavy loads in comparison to light loads.

Findings: In instances where 2" wedges reduced P_P, all of the subjects were light. Of those instances where 2" wedges reduced P_E, no differences in frequency between heavy and light subjects were observed.

3. Hypothesis: Anterior positioning of wedges should shift the centroid ($\bar{x}$) forward more effectively for heavy subjects than for light subjects.

   Findings: Where anterior positioning of wedges caused an anterior shift of $\bar{x}$, all subjects were heavy (body weight greater than 150 lbs). No anterior shifts of $\bar{x}$ were observed for any of the light subjects (body weight of 100-110 lbs).

4. Hypothesis: Posterior positioning of wedges should shift $\bar{x}$ forward more effectively for light subjects than for heavy subjects.

   Findings: Where posterior positioning of wedges caused an anterior shift of $\bar{x}$, there was no significant difference between the number of instances observed for light subjects and the number observed for heavy jects.

CONCLUSIONS

It appears that wedging can be an effective means of using one cushion type i.e. foams, to fit large numbers of people in the patient population. This technique can make it possible to reduce the cushion inventory at a hospital or rehabilitation center and provide a custom fit for a person relatively inexpensively. It is a technique that can be accomplished by most members of the rehabilitation team.

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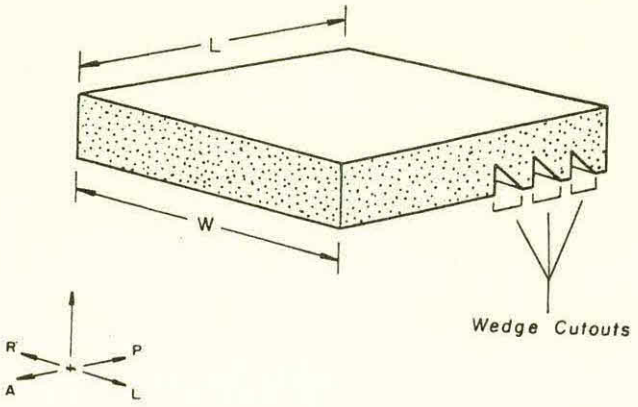
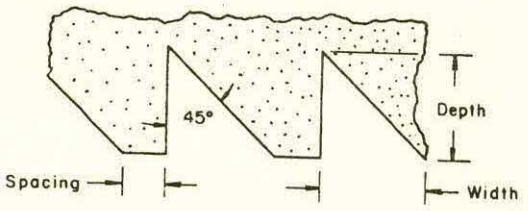


(a.) Schematic view of wedged wheelchair pressure relief cushion
(Coordinate axes are: A = anterior; P = posterior; R = right; L = left)



(b.) Cross-sectional view of cushion showing wedge dimensions

Figure 1

Table 1 - Description of Test Surfaces

Test Surface**	Wedge Parameters*			
	Number	Depth	Spacing	Position
1	Plywood Surface			
2	Unmodified Temperfoam			
3	5	1"	3/4 "	Anterior
4	5	1"	3/4 "	Posterior
5	3	2"	3/4 "	Anterior
6	3	2"	3/4 "	Posterior

*Wedges are removed over 1/2 of the cushion surface only

**Test surfaces (3) through (6) are identical to (2) except for the modification by wedging

PRESSURE DISTRIBUTION UNDER THE SEATED DISABLED WORKER PERFORMING MAXIMAL REACH

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The Institute for Rehabilitation and Research
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Severely disabled persons attempting to maintain gainful employment are confronted with two major problems: 1) Work time lost due to tissue pressure and renal problems and 2) Decreased work output due to lack of specialized work station design.

The major objective of this study is to investigate the relationship between pressure distribution and arm positioning. To measure pressure relief and coordinate the body movements that affect it, A Range of Motion Sensor (ROMS) has been used in conjunction with the Pressure Evaluation Pad System (PEP) (1), both instruments developed at The Institute for Rehabilitation and Research (TIRR), Houston. In this manner, general directions of reach and reach distances which induce pressure relief could be determined.

To date, 16 subjects have been tested and the following inferences have been drawn: 1) Contra-body arm movements in the direction 30^{o}-90^{o} left of center and 120^{o}-150^{o} right of center locate ischial pressure relief zones and 2) Arm reaches approximately 1.5-1.75 times mean normal reach are required to affect pressure relief.

STATEMENT OF THE PROBLEM

Pressure induced tissue problems are a major cause of time lost from the job for the severely disabled. In the industrial setting the person with diminished or absent sensation and limited lower body movements may be limited to positions which induce excessive tissue pressures leading to skin breakdown, especially under the ischial tuberosities.

Weight shifting, which normally requires conscious attention and is necessary for pressure relief and thus prevention of tissue breakdown, is more difficult for the working individual with sensorimotor impairment, since the focus of attention is on work-related activity. Upon observation of the traditional work space design, a question arises as to whether a working environment can be defined which inherently will induce sitting pressure relief for such an individual. By altering the limits of the traditional workspace, the disabled worker may be required to assume positions of pressure relief as a part of his job.

Barnes describes a maximal work area as that space encompassed by two arcs, one drawn by each arm extended and pivoting at the shoulder joint (2). The person with sensorimotor impairment is unlikely to achieve much pressure relief while working within this space. However, reaching some distance beyond the described arc should result in truncal and pelvic displacement, thereby providing ischial pressure relief.

OBJECTIVES

This project's broad objectives are to study how the work environment can be designed or modified to: 1) Permit disabled worker activity, as part of the work-task performance, to contribute to pressure relief and thus the long term productivity and 2) Consider future alternative work-station layout equipment designs to assist in the improvement of short-term productivity to competitive levels.

The major purpose of this study is to investigate the relationship between pressure fluctuation and arm positioning under the seated client reaching to various points within and beyond the conventional work zone. This will allow identification of positions which may be incorporated into the work routine for pressure relief.

In addition to tissue pressure data, relevant anthropometric data are gathered relating to the physical layout of the workplace that accomodates the severely disabled worker.

METHODOLOGY

The major objective of this project is being accomplished by the application of the Pressure Evaluation Pad System (PEP) (1) and the Stereometric Range of Motion Sensor (ROMS), developed at The Institute for Rehabilitation and Research, in Houston, Texas.

The subjects (N=75) for this study are individuals with sensorimotor impairments including diagnosis such as spinal cord injury, amputation, cerebral palsy, stroke and orthopedic problems. The investigations are being conducted within the Vocational Industrial Center (VIC), a vocational rehabilitation program of TIRR.

The collection of data defining the workspace and subsequent movement beyond this area relative to pressure relief measurement is accomplished with ROMS and PEP. The ROMS unit operates as a surveyor's theodolite which records spherical coordinates of the location of a point in space (two angles and a distance). Optical encoders convert the mechanical movement

of the sensor into measurable pulses passed to a quantizer which converts this output to scaled digital spherical coordinates representing location in space of the free end of the sensor arm (subject's hand). The output in its digitized form is passed to a computer which provides both numeric and graphic representations of the workspace and the locations representing tissue pressure relief zones relative to the defined workspace.

The PEP System monitors pressure relief under the ischial tuberosities of the seated subject. The system consists of a pad containing a flexible printed circuit, upon which the subject sits, and an electronic display of the pressure distribution under the buttocks of the seated client. The precise location of the ischia (and sacrum if necessary) is determined by palpation of the subject.

The ROMS is first used to identify four standard reference points for each arm (left and right table edge, sternum and acromial process), which are recorded. Subjects, holding a T-handle of the ROMS are asked to describe maximal arcs with each arm, while points along the arcs are recorded on magnetic tape. Subjects are then asked to reach as far as possible along five grooved radii on the table, contralaterally in respect to each arm. The points of pressure relief at 40 mmHg under the ischial tuberosities are monitored through the PEP system and recorded on tape. The procedure is repeated three times for each arm.

Body dimensions including arm length, trunk length, and sitting height are measured manually with a tape measure. The wheelchair cushion resistance is measured with a spring balance durometer.

The data are analyzed by the computer to determine if there are feasible points on the table surface that force a subject to achieve pressure relief.

RESULTS

Up to this date sixteen (16) subjects have been tested. The data are represented in the graphic plots of each subject's set of points that represent their normal extended arm reach (Barne's maximum) and a set of points that represent the reach required to achieve relief of pressure at or below 40 mmHg for the right and left ischial tuberosities (Figures 1 and 2).

Little statistical validity can be associated with a sample of this size, however useful trends are apparent. In order to generate more reasonable inferences, a minimum of 75 subjects will be tested.

CONCLUSIONS AND SIGNIFICANCE

As depicted in Figures 1 and 2, certain prominent zones of pressure relief are apparent. It appears that contra-body arm movements in the direction 30^{o}-90^{o} left of center for the right arm and 120^{o}-150^{o} right of center for the left arm locate pressure relief zones. In conjunction with these zones, it appears that reaches approximately 1.5-1.75 times the mean normal reach for each subject is required. Further research, that better defines these zones and reach distances, is needed to reach statistically valid inferences.

The information gained from the research evaluation of anthropometric data, arm reach studies and the pressure relief studies will add to the basic knowledge of better integrating the workplace for the disabled worker. An understanding of the relationship of pressure distribution and arm reach will provide a new concept of the "normal workspace" for the wheelchair bound disabled worker and contribute to the development of systems which provide pressure relief, thus reduce the incidence of pressure ulcers. Ultimately, when completed, these findings may alter the conventional industrial practices relating to workplace layout and may increase long and short term productivity of the disabled worker to competitive levels.

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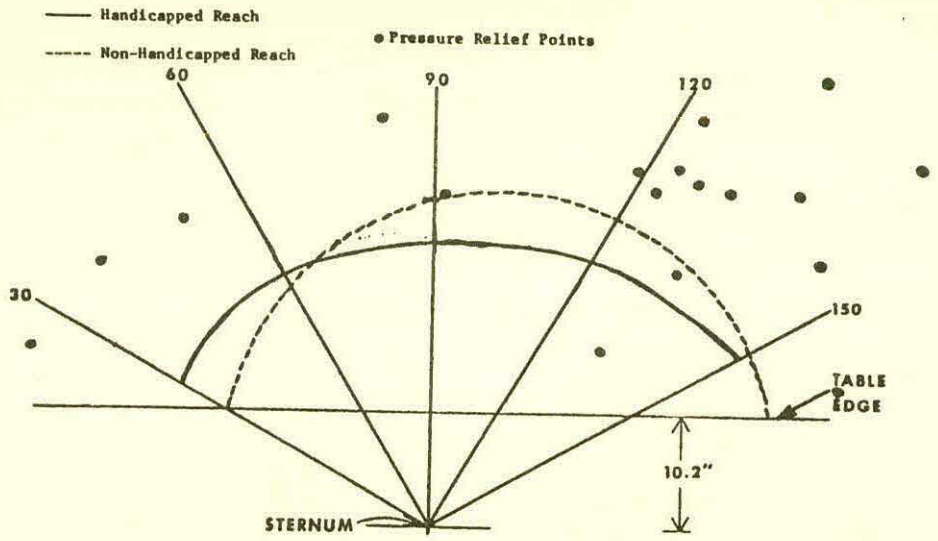
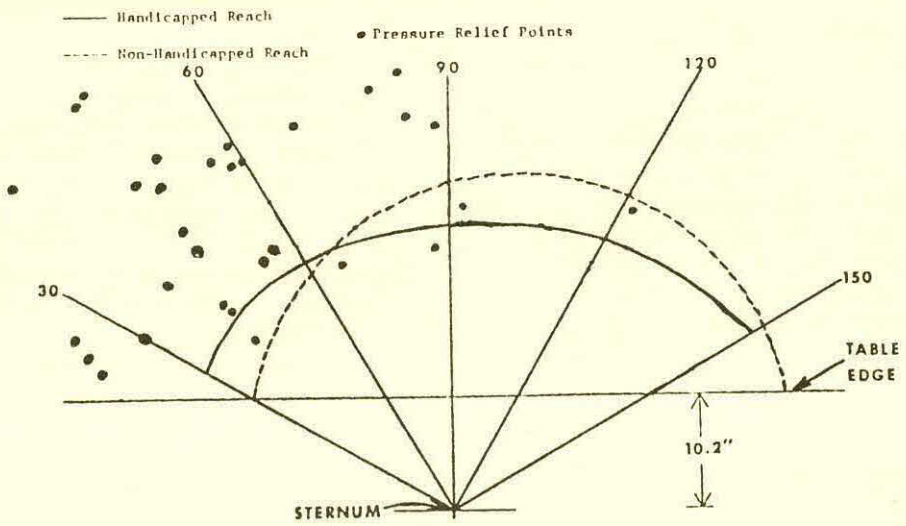


Right Arm Reach - Left Ischium Pressure Relief Points

Figure 1



Right Arm Reach - Right Ischium Pressure Relief Points

Figure 2

THE PRESSURE DISTRIBUTING PROPERTIES OF HOSPITAL MATTRESSES AND THEIR COVERS

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ABSTRACT

A requirement of mattress es for patients predisposed to pressure sores is to minimise the compression of body tissues over bony prominences supporting the patient's body weight. In the immediate future the authors foresee the continued use of flexible foam hospital mattresses for many patients and consequently have investigated the variability of body/mattress interface pressures produced by groups of spinal injury and able-bodied volunteers during sleep and on a selection of mattresses and mattress covers currently being issued to patients in UK hospitals.

A low profile, variable capacitance pressure sensor was developed for this study. Laboratory-based indentation studies were also performed in order to characterise the mechanical properties of the mattresses used. Data collected in this study confirm that the able-bodied volunteers moved frequently during sleep, producing pressure relief on average every 40 minutes. Significant differences in interface pressure were observed and corresponded to differences in the mechanical properties of the mattress and covering materials used.

INTRODUCTION

Pressure sores impose an additional burden upon the already weak and handicapped patient, invariably lengthening the duration of his stay in hospital following injury, surgery or treatment for a primary condition. Many patients trying to cope with limited physical capabilities in the home or sheltered community are frequently immobilised and separated from their family and friends in order to receive treatment for pressure sores. A recent survey of pressure sore prevalence in the Greater Glasgow Health Board Area showed that 8.8% of over 10,000 patients in hospital or the community had at least one pressure sore (1). More than 70% of these patients were elderly (>65 years) and many were incontinent, weak, and confused. As the average age of the population increases and techniques for treating primary clinical conditions are further developed, it has been anticipated that the number of patients vulnerable to pressure sores will increase substantially (2).

The development of methods for preventing pressure sores is desirable in the interests of the patient and in order to conserve scarce and expensive clinical resources.

The physical and mechanical properties of the support surfaces upon which the patient sits or sleeps are thought to influence the conditions which lead to necrosis of the skin and subcutaneous tissues. A number of active support surfaces (e.g. ripple beds, tilting beds) are available in addition to the many beds which provide hydrostatic support (e.g. Low Air Loss, Water and M.U.D beds, and fluidised particle beds). Their cost, complexity, size and weight tend to limit their use to patients who are particularly at risk, or who are being treated for serious multiple pressure sores or burns. The authors while recognising many of the advantages of such support surfaces consider that in the immediate future a large group of patients vulnerable to pressure sores will continue to use conventional foamed polyether or latex mattresses both in the home and in hospital.

The following study was established in an attempt to demonstrate the influence of the properties of mattresses and their coverings on body/support interface conditions thought to be significant factors in the development of pressure sores.

AETIOLOGICAL FACTORS

Fernie (3) attributes the cause of superficial pressure sores to maceration of the tissues associated with sweating or urine spillage which are frequently irritants and serve as a source of infection. Deeper sores follow prolonged ischaemia due to compression of the tissues overlying bony prominences supporting the body weight. Additional factors thought to exacerbate the development of pressure sores include elevated skin temperature, the presence of moisture, and the patient's general nutritional and clinical condition.

The occlusion of blood vessels and lymphatics in tissues may be initiated by tissue deformations resulting from local force distributions, loosely described as 'pressures'. The relationship between the applied forces and resulting deformations within the tissues are likely to be complex (4). The prediction of physiological behaviour in the tissues from data collected at the boundary between the support surface and the body presents in general considerable difficulties. Such data does, however, enable a comparison of the mechanical and physical properties of the patient/support interaction which can be used to identify differences to aid in support selection. Due to the difficulties in equating such measurements with physiological effect it is the

extent of the improvement obtained from making such selections that is difficult to quantify.

In order to compare the characteristics of standard hospital mattresses and covers the authors have monitored body/support interface pressures and temperatures, and body movement.

MOVEMENT DURING SLEEP

The damaging effects of the forces applied to tissue when supporting body weight are thought to depend upon both the magnitude and duration for which they act (5, 6). The total force directed through the bony prominences is determined by the patient's body weight, but its distribution and related pressures depend upon the patient's build, posture, and the properties of the support system.

The normal individual ensures tissue viability by making movements which alternate the body sites required to support the weight of the body. The vulnerable patient is usually too weak to make such movements, or is not stimulated to do so due to sensory loss. These patients are either required to execute a conscious regime of movements, or may be deliberately turned by nursing staff at regular intervals. These intervals vary considerably, but for the bed-bound patient, pressure relief (turning) frequencies ranging between 1 and 3 hours are generally recognised as adequate.

In an attempt to compare the frequency of pressure relief for a group of healthy volunteers with the turning frequencies provided for patients at risk, the authors have developed a facility equipped to measure pressure and temperature variations over the trochanters and sacrum, and movement during sleep (7, 8).

A group of 13 able-bodied volunteers were monitored on a standard 5" (127mm) thick polyether foam mattress of density 32-34 kg m^{-3} (Dunlopillo Ltd., Harrogate) covered with a proofed nylon cover and hospital sheets (1 full length sheet, a plastic drawsheet, and a linen drawsheet) supported on a King's Fund bed. Matrices of thin flexible pressure sensors (7, 9), four attached to each trochanter and two to the sacrum, were employed in the study. Local skin temperature measurements were obtained using small thermistors (ITT, Type M52).

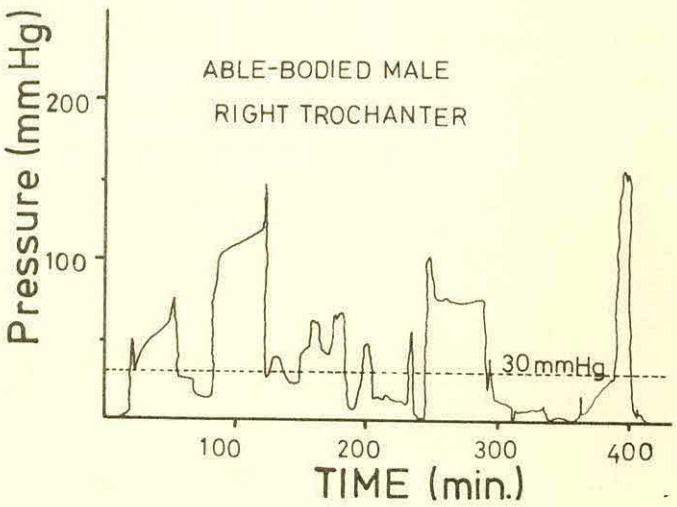
The temperature and pressure sensors were 'read' automatically every 10 minutes using a data logger (Mycalex, Series 5) and output onto punch tape. Scan intervals of 1 and 5 minutes were also employed for a sub-group and indicated only a very modest loss of information when compared with data produced from monitoring sessions with 10 minute scan intervals. The longer scan interval thereby limited computing time to a practical level and obviated the need to change punch tapes during the monitoring session.

A 35mm camera with motorised film drive loaded with film sensitive to infra-red radiation was placed above the bed which was illuminated using a flash gun covered with an infra-red filter (10). Photographs were taken every 10 minutes in synchronisation with the temperature and pressure sensor readings.

The results from these overnight monitoring sessions indicated intermittent elevations in interface pressure with periods of varying duration (figure 1).

Figure 1

Variation in interface pressure at body/mattress interface during sleep. Scan interval = 1 min. Mattress = polyether foam + proofed nylon cover.



Fluctuations in the pressure between 0 and 30 mm Hg (4.0 kPa) were observed throughout the night, even when photographs indicated that the site was not bearing body weight. These small fluctuations were consistent with variations in blanket pressure and minor transducer artefacts.

The range and distribution of pressure readings obtained are shown in figure 2 and give an average pressure of 76 mm Hg (10.1 kPa) for the transducers placed over the trochanters and 75 mm Hg (10.0 kPa) over the sacrum.

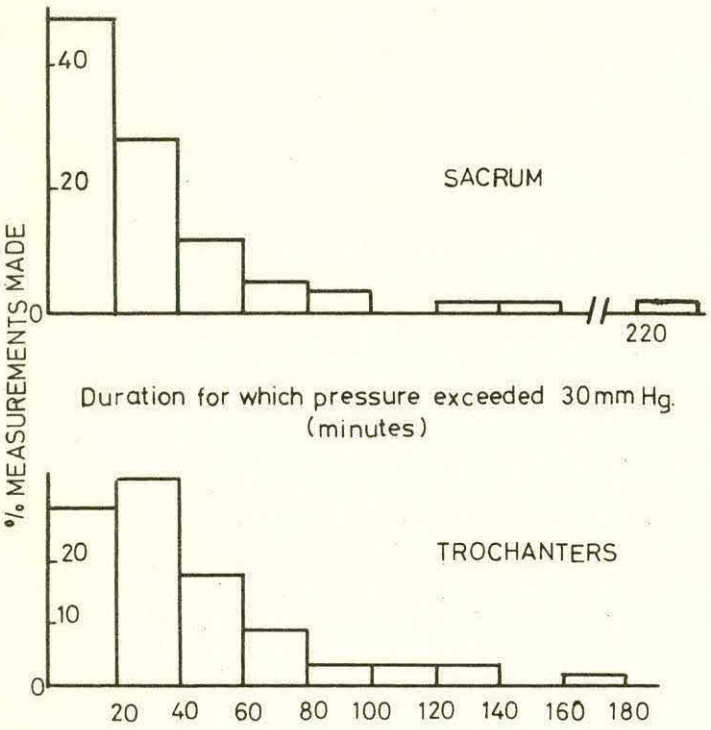
The duration of periods for which the pressure exceeded 30 mm (4.0 kPa) also varied widely within the ranges 0-240 minutes for the sacrum and 0-180 minutes for the trochanters. The average durations for which these sites were loaded were 41 minutes and 34 minutes for the trochanters and sacrum respectively. This difference did not prove to be statistically significant using Student's t-test.

The duration for which the able-bodied volunteers subjected their tissues to pressures greater than 30 mm Hg (4.0 kPa) was considerably shorter than that tolerated by patients being turned every two hours. Over 90% of the loading periods generated by the volunteers were shorter than 2 hours in duration.

Elevated skin temperatures at the body support interface were found in all tests for both able-bodied and patient volunteers. The maximum elevations were of

Figure 2

Distribution of loading period durations for 13 able-bodied volunteers during sleep.



the order of 4°C but both the magnitude and duration of the raised temperatures were very variable.

The major difference between the pressure and temperature histories was the time dependence of the temperature elevation. The pressure at the interface rapidly reached an equilibrium value, and in the absence of movement, remained constant. The temperature rose slowly, and an equilibrium value was reached only after a prolonged period. The increase in temperature depended on the body temperature and thermal flux, both of which are under physiological control and vary during the night. These factors prevent the display of temperature elevations in a manner analogous to the pressure results.

PRESSURE DIFFERENCES ACCORDING TO SUPPORT SURFACE

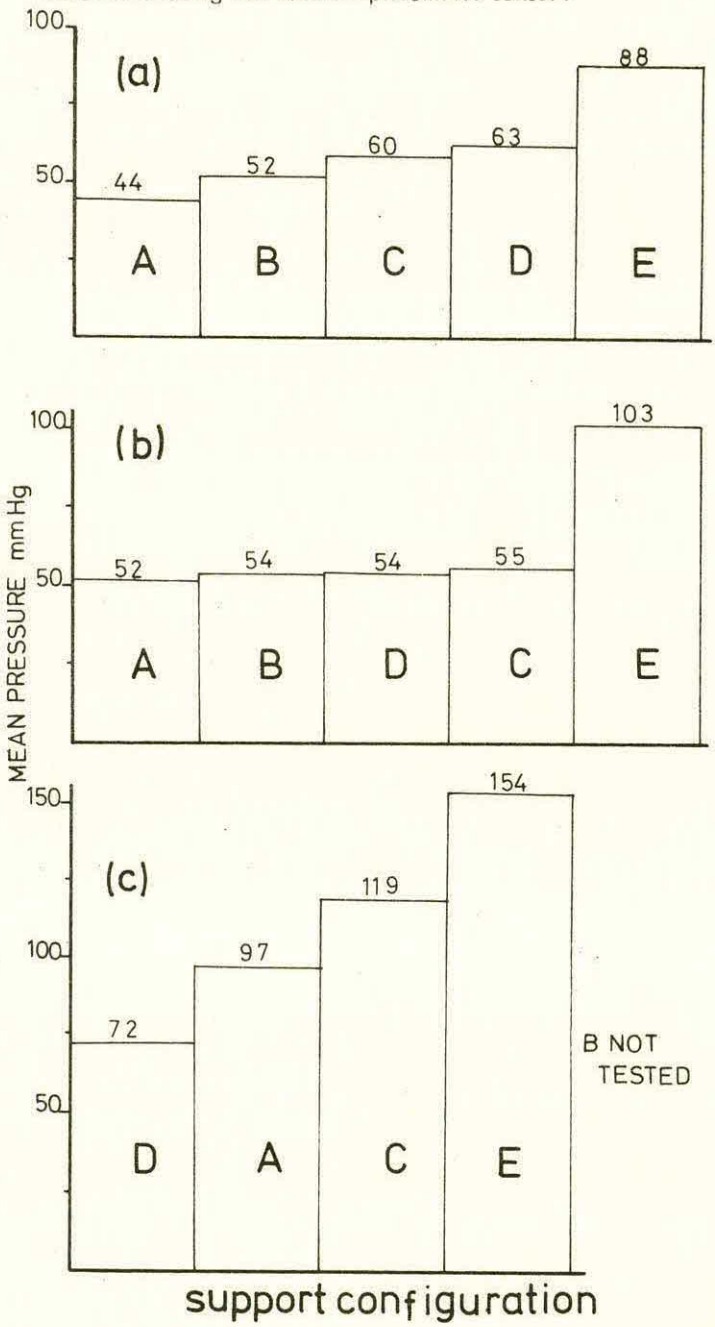
Mechanical indentation tests (11, 12) have demonstrated the influence of non-stretch mattress covers on the compressive stiffness of the support surface. In order to determine whether corresponding differences in body/support interface pressures occur, a series of tests was undertaken with a group of five able-bodied and nine spinal injury volunteers lying on a selection of mattress/cover combinations.

In addition to the thin, flexible, capacitive pressure sensor (7) used in the overnight monitoring sessions a series of parallel measurements was made with an electro-pneumatic (Talley Surgical, Borehamwood, U.K.) sensor 10 cm in diameter developed by Rogers (6).

Measurements of interface pressure were obtained with the volunteer's body assuming a specific posture. The legs were extended in line with the trunk and the intertrochanteric axis was vertical. The pressures measured with the body in this position using the capacitive pressure sensor are shown in figure 3(a) which contains the pooled results for the able-bodied volunteers on the support surfaces studied. Comparable measurements using the electropneumatic device are shown in figure 3(b).

Figure 3

Pressures measured on different mattress/cover configurations with (a) able-bodied volunteers using variable capacitance pressure sensor (b) able-bodied volunteers using the electro-pneumatic sensor (c) spinal injury volunteers using the electro-pneumatic sensor.



A DS184 Standard NHS 5" medium density (32-34 kg/m^3) flexible polyether foam with extensible knitted cover.

B As in 'A', fitted with drawsheet

C As in 'A' but fitted with inextensible proofed nylon cover, no drawsheet fitted

D 'Polyflotation' mattress, 6" thick with grooved surface.

E Interior sprung mattress with waterproof cover.

The grand mean of 63 mm Hg (8.4 kPa) obtained for all support surfaces using the pneumatic device does not differ significantly from that of the capacitive sensor which gave a grand mean of 63.5 mm Hg (8.4 kPa). The pneumatic sensor appears, however, to give poorer discrimination of the effects of altering the support system characteristics.

The grand mean of the results for the standard polyether mattress covered with a proofed nylon inextensible cover was compared, using Student's t-test, with the other mattress/cover configurations. The spring interior mattress produced a grand mean interface pressure which was significantly higher than the standard mattress ($p < 0.001$). Pressures measured with the capacitive sensor were significantly lower for a polyether mattress covered with an extensible knitted fabric when compared with the proofed nylon covered mattress.

The 'Polyflotation' mattress (Talley, Borehamwood, U.K.) differed from the other foam mattresses in two respects. The mattress was 2.5 cm thicker than the standard foam mattress, and its surface scored to a depth of about 5 cm in an orthogonal grid pattern. The scoring is thought to locally soften the foam and reduce the generation of shear stresses in the tissues. The mean pressure produced by the able-bodied volunteers on the thin mattress was slightly higher than that of the standard mattress.

The effects of introducing linen and plastic drawsheets to the support surface were investigated for the mattress fitted with a knitted cover. A substantial increase in mean pressure was observed with the able-bodied volunteers which was significant at the $p = 0.08$ level.

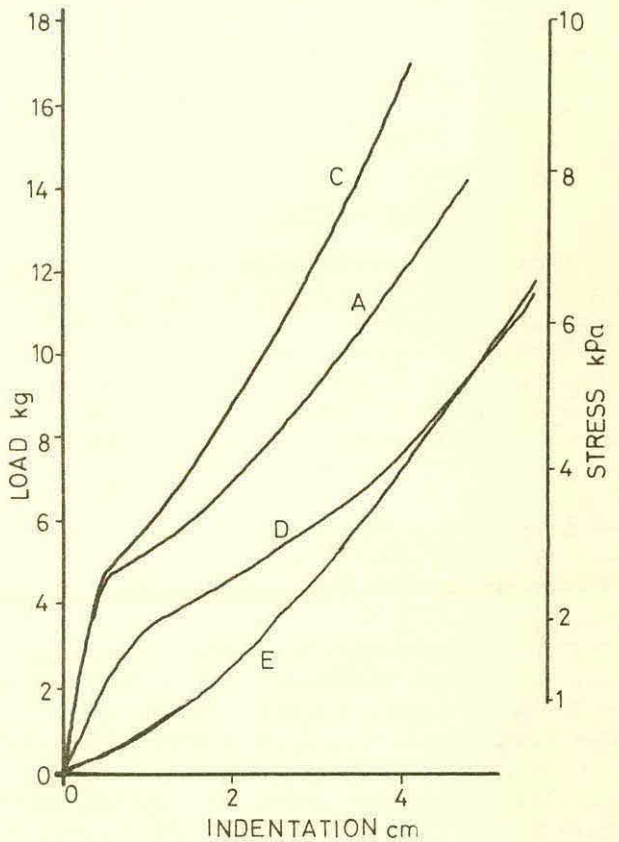
Figure 3(c) represents the distribution of pressures obtained with the group of spinal injury volunteers on similar mattresses. Pressure measurements were obtained with the electro-pneumatic device. The overall grand mean for interface pressures generated by the spinal injury volunteers was 109 mm Hg (14.5 kPa) and differed significantly from that of the able-bodied volunteers. The ranking of the mattresses according to interface pressure is the same for both groups of volunteers, apart from the 'Polyflotation' mattress which produced the lowest mean interface pressure for the spinal injury volunteers but ranked less favourably for the able-bodied volunteers.

MATTRESS INDENTATION TESTS

The load indentation properties of the mattress/cover combinations investigated with recumbent volunteers were determined using controlled rate indentation tests (figure 4). The deformations to the mattress which was supported on a metal bed frame topped with a rigid wooden board, were applied with a mechanical test machine (Instron, model TTDM) modified to enable investigation of the complete mattress (11).

Figure 4

Load indentation properties of hospital mattresses indicating yield point for flexible foam mattresses A, C, D, and differences in properties attributable to mattress covering. Key given in figure 3. Indentation rate = 5 cm/min., indentor diameter 15 cm. (After Small, 11)



Flat-ended rigid indentors were used to indent the mattresses and ranged in diameter from 5 cm - 40 cm.

The flexible foam mattresses produced a characteristically non-linear load-indentation response. The initial response was relatively stiff with a modulus of approximately 6 kPa. At a critical nominal stress of about 2 kPa, yield occurred, and the post yield behaviour was less stiff. When the indentation was continued, the mattresses' stiffness increased as the foam was compressed to a solid mass. The clinically useful range is in the second phase.

The spring interior mattress exhibited low initial stiffness but on progressive indentation the stiffness increased.

The magnitude of the nominal pressures at different indentations reflected the non-linear foam characteristics but was found to be highly dependent upon the diameter

of the indentor used. The 'Polyflotation' mattress consistently showed lower loads than the other foams for equal indentations. This was particularly marked for small indentors, and may also be attributed in part to its greater thickness (15.2 cm).

The mattress covers appeared to have little or no effect on either the pre-yield response or the magnitude of the yield stress. The effect in the second stage of the load-indentation response depended on the mattress cover material, the indentor diameter and indentation. In all cases the modifying effect of the covers was smallest with large diameter indentors.

The effect of the mattress covers depended on the mechanical properties of the cover material and their fit on the mattress. The biaxially stretchable knitted cover produced only minimal alteration at all indentations. The proofed nylon cover produced a large increase in the measured load at specific indentations. The effect was particularly marked for large indentations and small indentors.

DISCUSSION AND SUMMARY

Measurements on able-bodied volunteers at the interface between bony prominences and the mattress indicated that pressures in excess of 30 mm Hg (4.0 kPa) were endured on average for periods of 40 minutes. Average pressures of 75 mm Hg (10.0 kPa) sufficient to disrupt tissue nutrition, were produced but pressures normally associated with severe ischaemia occurred only occasionally ($< 20\%$ of measurements exceeded 100 mm Hg, 13.3 kPa).

Frequent movements encourage ventilation of the body/support interface, reducing elevated skin temperatures and the maceration of tissues due to the presence of moisture.

The conditions developed by patients on these support surfaces are more severe. Less frequent movements lead to extended periods of ischaemia which is often severe since higher pressures are normally generated by patients due to tissue wasting over bony prominences.

If two-hourly turning periods cannot be reduced for practical reasons, it would seem desirable that the properties of hospital mattresses should be such that they prevent moisture build-up, distribute interface stresses effectively, and have thermal properties that maintain skin temperature at a moderate level.

Investigation of the pressure distributing properties of flexible foam mattresses covered by various materials indicated that pressures could be significantly reduced by the use of biaxially stretchable mattress covers. An interior-sprung mattress in widespread use in the U.K. produced substantially greater interface pressures than the standard foam mattresses. Drawsheets were found to increase pressure significantly and such data suggests that the tension produced in sheets when the bed is made is likely to negate any benefits obtained from improved mattress and covering materials.

ACKNOWLEDGMENTS

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REHABILITATION THROUGH NONVOLITIONAL BIOFEEDBACK AND POSSIBLE NEURAL MECHANISMS

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ABSTRACT

A technique to control one's physiological functions is to make one aware of one's on-going physiological parameters such as selected waveforms of EEG, heart rate- GSR, skin temperature etc. The subject achieves control of these parameters through volitional (though undefined) thought processes. This report presents a novel non-volitional EEG biofeedback scheme to control the psychophysiological parameters of a subject. A hypometabolic state is induced during alpha feedback. Possible neural mechanisms are discussed in the light of preceptual ability and altered arousal level of the subject. Clinical trials indicate the usefulness of this non-volitional biofeedback procedure as a rehabilitative technique in controlling certain psychosomatic disorders.

INTRODUCTION

Information or volitional feedback is made available to a subject by providing monitoring facilities for the physiological variable and presenting it in a form that can be understood by the subject. For example, a heart rate monitor attached to a subject will provide information on a meter, the subject's own heart rate. By generating appropriate thought processes, the person can learn to change his heart rate, skin temperature, blood pressure, muscle tension or even certain brain rhythms. A large body of literature has grown in the recent past to establish several therapeutic procedures in psychosomatic medicine through biofeedback (1, 2, 3, 4).

In the autogenic feedback system (2), the subjects are trained (over a number of sessions) to control three physiological variable, namely reduction of muscular tension in fore-arm, increase in percent of alpha rhythm in the occipital EEG waveform and increase in temperature of the fingers. The subjects learned to reduce muscle tension and were able to increase alpha content of EEG thus achieving a state of relaxation. In another system, the alpha activity triggers a blue lamp. Here too the subjects are instructed to proceed to a 'feeling state' which would increases the intensity of the lamp. In both these systems, as well as in all other systems of biofeedback developed to date, the subject's interaction with the instrumentation and his volitional effort to change the parameter is required. We present a system hereunder termed the non-volitional biofeedback wherein the subject's will and cooperation are not necessary to effect a change in his physiological processes. The system has inherent advantages with patients who cannot coordinate a volitional change. This report presents systematic studies with normal subjects and possible mechanism and learning processes in the Central Nervous System (CNS) responsible for bringing about a psychosomatic response.

MATERIALS AND METHOD

Figure 1 shows the schematic of the non-volitional visual EEG biofeedback system developed here. The unit consists of an electrode system which is used for monitoring occipital and temporal EEG of the subject undergoing feedback. EEG amplifiers and a polygraph for contineous recording forms the basic unit. Any one of the scalp lead output is used for deriving the feedback signal. Four bandpass filters span the beta, alpha, theta and delta (14-18 Hz, 8-13 Hz, 4-7 Hz and 1-3 Hz respectively) bands. The output of one of these filters is power amplified and drives a filament lamp which forms the visual feedback signal to the subject. This report deals only with alpha feedback wherein the subject's own scalp alpha drives

the lamp.

To measure quantitatively the changes in alpha content of the EEG, an alpha counter is designed and used. This gives the abundance of alpha present above a predetermined threshold. The alpha waves are fullwave rectified, compared to a reference voltage and gated for counting. The gate is allowed to pass clock pulses of one microsecond width and the number of clock pulses during the presence of alpha is displayed. Thus, a measure of alpha occurance is obtained from the alpha counter.

All experiments reported herein are conducted with alpha band only which is nearly a band limited Gaussian random noise. An initial assessment of efficacy of cortex derived alpha as a feedback signal in comparison with a noise source having similar characteristics is done as follows. A random noise generator having the characteristics of band limited Gaussian noise was designed and was made to drive the lamp of the feedback monitor. The abundance of alpha is counted during this simulated EEG feedback and compared with alpha feedback as described below.

EEG is monitored in a quiet, dark room. Scalp recordings from T_3-A_1 and its alpha derivative is used for biofeedback. Initially, EEG from occipital, perietal and frontal channels are also compared before and during feedback. However, since the alpha increase in these channels are not significant, only T_3-A_1 recordings are carried out for eight subjects. The following procedure is adopted for normal volunteers. The subjects are made to relax with closed eyes and their basal alpha is counted every half a minute over a five minute period. The subject is then given simulated alpha wave derived from random EEG generator and again the subject's alpha is counted as above. After a break of ten minutes, the subject is given the scalp derived alpha feedback and the alpha is counted over the same time interval. Table I gives the summary of the counts obtained before and during alpha feedback.

RESULTS

A student 't' test performed on the alpha counts of normal subjects shows that the alpha counts increased significantly during alpha feedback and the changes are not significant during random simulation. During alpha feedback, the changes in total EEG pattern are significant. Refer to table I. Firstly, an increase in alpha amplitude is observed. Secondly, alpha activity become alpha rhythm with stable, long time waveforms from the temporal area. Prolonged alpha feedback to epileptics have resulted in decreased intensity and number of seizures (5). The above feedback procedure is carried out without any active cooperation from the patient; hence this is found to be useful as a therapy for the young and for the uneducated/ uncooperative.

An important procedure such as this nonvolitional alpha feedback system should be understood in the light of available physiological processes of the CNS. We discuss hereunder the mechanisms and modalities of alpha feedback and its role in influencing the midbrain and brain stem structures.

STATUS OF CNS DURING BIOFEEDBACK

During nonvolitional alpha feedback to a subject, as discussed earlier, alpha activity is increased with generation and maintenance of alpha rhythm. Further, some autonomic activity depression also takes place. We now present a neurophysiological model of status of CNS during biofeedback on the basis of available studies.

As early as 1933, reverberating circuits have been suggested between thalamus and cortex to account for the synchronization of alpha-like rhythms (6). A schematic of the reverberating corticothalamic circuit is shown in figure 2. Stimulation of thalamic nuclei at alphs frequencies produce characteristic alpha-like recruiting responses which is modulated by a low frequency waveform. Synchrony of alpha activity is produced by a thalamic circuit which includes a pathway for recurrent collateral inhibition. This pathway is shown in the figure. When neuron 2 discharges, an inhibitory neuron 3 is activated via a collateral branch. After a period of inhibition of neurons in this branch, they become hyperexcitable and discharge rhythmically. Thus, increase in alpha amplitude in the cortex is explainable through reverberation of this rhythm in the corticothalamic pathways as proposed in the above model.

Facultative Pacemaker Theory proposed by Anderson and Eccles (7) formulates

that all major thalamic nuclei have an ability to control an appropriate part of the cortex. In otherwords, a rhythmic activity in thalamus will generate and sustain similar activity at a specific cortical region. Further, Basar (8) has shown through carefully implanted electrode studies at various subcortical structures that not only cortical and thalamic nuclei but various areas of the brain including reticular formation and limbic system have also networks with similar design and alpha selectivity. These common selectivities exist both in the auditory and visual pathways. Indeed, alpha resonance seems to be a fundamental property of various subcortical nuclei as well as that of the cortex itself. An input in the alpha frequency range will thus, produce an increased alpha over the cortical area. Thus, both these characteristics of the model, namely cortical increase of alpha and alpha resonance are supported by the nonvolitional biofeedback experiments reported here.

A last point pertains to psychophysiological response of a subject to nonvolitional biofeedback as a therapy. Studies have been conducted on epileptics and patients with psychophysiological deficit. Figure 3 shows an EEG of a generalized epileptic before feedback and during a biofeedback session. A dramatic change in the scalp monitored EEG is observed. Indeed such changes are observed in about sixty percent of the subjects during NBF. Initial studies indicate that the number and intensity of epileptic attacks reduce after a period of NBF therapy (5). However, long term rehabilitation of these patients after a short period of therapy is yet to be assessed. Results do indicate possible facilitation of certain neural connections resulting in improved EEG waveforms and in decreased deficit. Follow up studies are required to establigh these results unequivocally.

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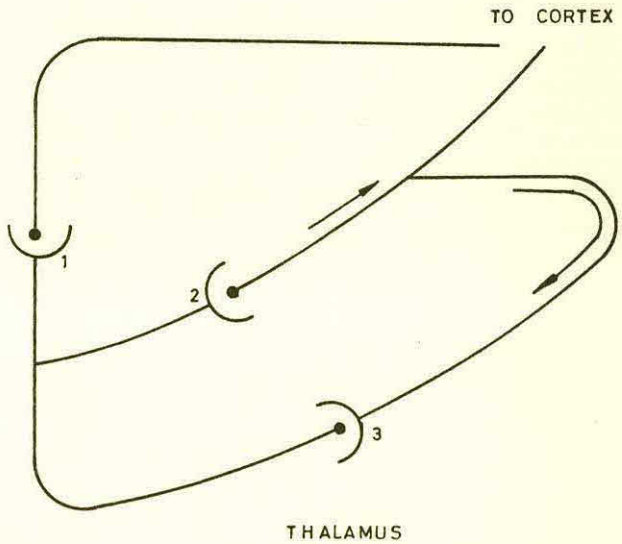

FIG.2

EEG ALPHA VISUAL FEEDBACK : LEAD PLACEMENT T_3 A_1

Time	Sub I		Sub II		Sub III		Sub IV		Sub V		Sub VI		Sub VII		Sub VIII	
	B	D	B	D	B	D	B	D	B	D	B	D	B	D	B	D
1	3	5	14	22	6	6	7	15	26	23	14	19	7	11	25	34
2	6	16	33	45	10	15	16	34	62	65	30	41	16	24	52	68
3	8	25	54	74	15	23	31	50	98	109	51	65	22	41	79	100
4	17	31	72	100	17	27	46	64	134	144	66	90	30	58	110	137
5	25	43	90	127	20	33	62	82	164	184	87	110	36	76	142	170

B = Before ; D = During : Changes statistically significant (P less than 0.05)

TABLE I

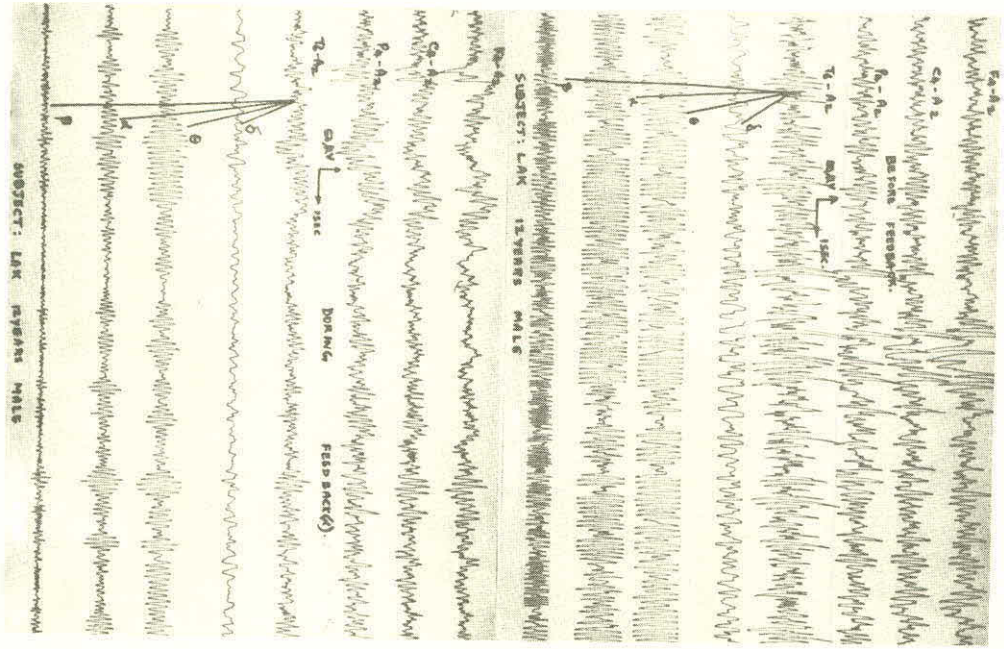

FIG. 3

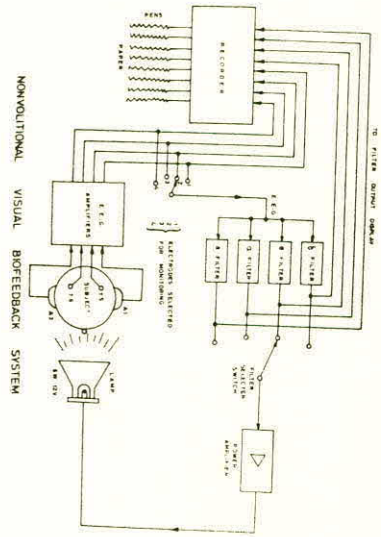

FIG. 1

ON THE INTEGRATION OF BIOFEEDBACK INTO A CHILDREN'S WORLD

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ABSTRACT

There is considerable potential for the use of biofeedback techniques in a paediatric rehabilitation population. However, to attain maximal effectiveness in the use of biofeedback with children it is necessary to make alterations to the usual paradigm that is used with adults. This paper will discuss one approach to achieving the integration of biofeedback to a children's world through the translation of purely informational auditory or visual feedback into feedback which has rewarding qualities. Case studies will be used to illustrate the effectiveness of this approach.

INTRODUCTION

In a recent Task Force Report on Physical Medicine and Rehabilitation to the Biofeedback Society of America, Fernando and Basmajian(1) recommended the routine inclusion of biofeedback techniques in the treatment of patients with: hemiplegia, lower motor neuron lesions, cerebral palsy, spasmodic torticollis, temperomandibular joint pain and low back pain due to muscular spasm. Although the vast majority of published research in this area has focused on an adult population, there is a large paediatric rehabilitation population that might potentially benefit from the use of biofeedback techniques. Finley, Lubar, Miller, Musicant, and Wansley(2) recently surveyed the journal Biofeedback and Self-Regulation and found that only 14.3% of its articles dealt with a paediatric population. It seems likely that a similar situation also exists in relation to the use of biofeedback in paediatric rehabilitation settings. Finley et al(2) hypothesize that the relative dearth of studies employing biofeedback with children is related to a frequent lack of comprehension and/or internal motivation in this population group.

In spite of these problems, it is speculated that children may, in fact, have a greater potential than adults for learning self-regulatory responses through biofeedback due to the malleability of the developing nervous system and due to a lack of attitudinal constraints regarding the possibility of self-control(3). A recent study by Loughry-Machado and Suter(4) provides support for this hypothesis. These authors compared 38 children (ranging in age from six to ten years old) and one of their parents, in ability to voluntarily regulate hand skin temperature using auditory feedback. Results obtained showed that the children were significantly better than their parents in controlling hand skin temperature in a short (20 minute) biofeedback session.

APPROACHES TO BIOFEEDBACK WITH CHILDREN

Finley, Niman, Standley, and Wansley(5) have developed an approach which they term "electrophysiologic behavior modification" as one possible means of overcoming the lack of interest and motivation often demonstrated by children in a biofeedback setting. Their approach, derived from a behavioural paradigm, utilizes concrete rewards (such as candy treats or toys) which are delivered to the subject instantaneously by a Universal Feeder which is connected to the biofeedback apparatus, upon the occurrence of a specified physiological response. For example, after maintaining EMG levels below a set criterion for a specified time period, children with cerebral palsy who are learning relaxation skills are immediately rewarded with a prize they have previously chosen.

An alternative approach to enhance the cooperation and interest level of children within a biofeedback setting has been developed by the Rehabilitation Engineering Department at the Ontario Crippled Children's Centre (OCCC). This approach, which combines aspects of motor learning theory with principles of behavioural psychology, consists of providing a feedback signal that is itself rewarding to the child. Forms of reinforcing feedback that have been developed and applied within head position and EMG biofeedback training programmes include: turning on music from a transistor radio, activating a television set, running an electric train, and operating a slot-car racing set. The feedback signal, or reward, is presented in a binary (on/off) fashion contingent upon the desired head position or level of muscular activity(see Figure 1).

A similar technique has been reported by Brown and Basmajian(6), in which a "Bioconverter" is used to increase the positive valence (ie. reinforcement value) of feedback information during EMG biofeedback applications to upper extremity rehabilitation in an adult population.

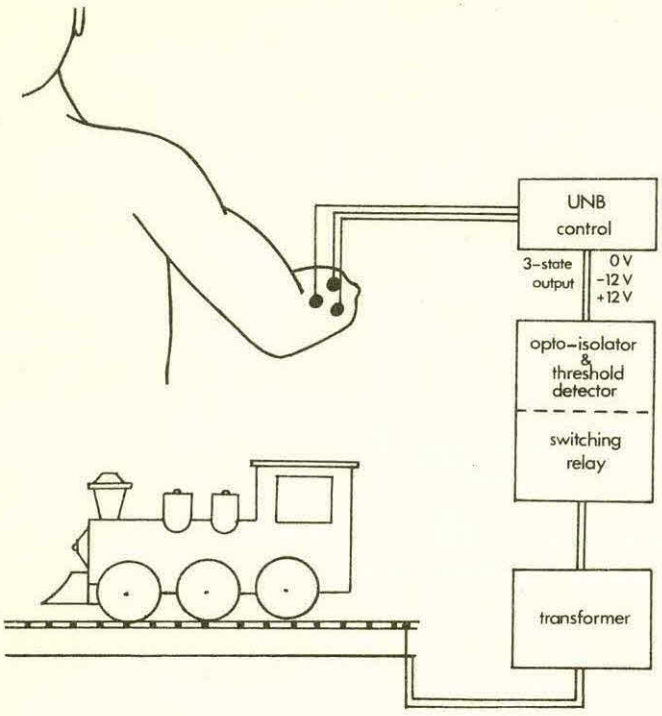

Figure 1: Block diagram for EMG training utilizing rewarding feedback. In this particular application, the 3-state output from the UNB myoelectric control governs the operation of an electric train. An output of +12V causes the train to move forward, an output of -12V causes the train to reverse direction, and an output of 0V stops the train.

Advantages of Providing Rewarding Feedback

Several advantages can be seen in this approach to biofeedback with children:

(i) A central tenet of learning theory is that if a behaviour is immediately followed by a reward, the future probability of the occurrence of that behaviour is increased. Thus, a response such as upright head position or decreased muscle activity is more likely to occur in the future if it has been immediately reinforced in the past.

(ii) Motivation and frequent practice are extremely important aspects of motor rehabilitation. The OCCC approach to biofeedback training ensures increased levels of motivation by making the training sessions fun for the children. This stimulates them to practice specific responses for greater lengths of time. This is especially important if one considers the proposition made by Kottke, Halpern, Easton, Ozel and Burrill(7) that millions of repetitions are necessary before engrams of coordinated action are established in the brain.

(iii) Children like to have concrete goals to work towards. Turning on a television set is more concrete than slowing down the frequency of a series of clicks, and thus is more meaningful from a child's viewpoint. It can be seen that there is an overlap between the biofeedback approach being discussed and the area of environmental control. Specifically, rewarding environmental events are placed under the "control" of the subject when he or she makes a desired response. This has advantages in overcoming what might be labelled a state of "learned helplessness"(8) in severely handicapped children. These children, because of repeated failures in past attempts to accomplish specific goals, have literally given up, since "nothing they do makes any difference". This syndrome has often been noted to change coincident with the introduction of biofeedback techniques in which the child has some degree of personal control over a desirable external event. In addition to promotion increased feelings and cognitions of personal control, the fact that the child is actively controlling a desired external event is also important in facilitating the future development of concept formation, especially cause-effect relationships.

(iv) Inherent in a quantitatively based biofeedback approach is the possibility of competing with oneself, that is, trying to improve upon one's previous performance. Also, through the slot car racing set adaptation, a means has been provided whereby severely handicapped children can actively engage in competitive activity with each other, perhaps for the first time in their lives. This aspect of competition increases the children's enjoyment, interest, and motivation towards the task at hand, and may overflow into other areas such as schoolwork with positive benefits.

(v) Another way of making biofeedback more meaningful for the paediatric rehabilitation client has been accomplished through the translation of responses learned through biofeedback training into functionally-useful activity. Within the OCCC programme, electric wheelchairs have been specially adapted so that the usual hand-operated joystick has been replaced with a head control, thus allowing tilting head movements (learned through head position biofeedback) to control mobility. Other possibilities for the translation of learned control to functionality exist in the areas of communication and environmental control.

The following case studies illustrate the application of the biofeedback paradigm employed at the OCCC with a paediatric rehabilitation population.

Case Study 1

Single case study methodology using an A-B-C-B reversal design(9) was employed to assess the differential effectiveness of purely informational (auditory) feedback and feedback consisting of a more rewarding form of information in establishing improved head control in T.B., a five year old severely handicapped girl with cerebral palsy. A Head Position Trainer (HPT) and Time Event Counter (TEC)(10) were used to monitor

appropriate head position, to provide relevant feedback information, and to obtain quantitative data regarding the percentage of time during a 15 minute session that the head was held appropriately upright. As shown in Figure 2, the provision of feedback information resulted in a consistent improvement in head control relative to the initial no feedback baselines. It can also be seen that T.B.'s performance was maximized during sessions in which feedback consisted of turning on the television set when her head was in the correct position as compared with sessions in which she received auditory "buzz" feedback when her head was in the incorrect position.

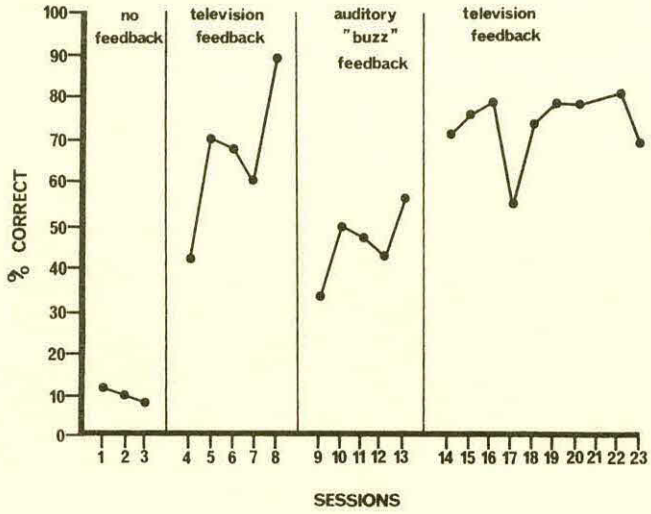

Figure 2: Acquisition of head control under conditions of no feedback, "buzz" feedback and motivational (TV) feedback for subject T.B., a 6 year old girl severely involved with cerebral palsy. Points on the curve represent the percent of time that the head was within a 25 degree range from midline during 15 minute sessions of TV watching.

This case study suggests that the type of feedback utilized within a biofeedback session may be of importance in eliciting maximum cooperation and performance with a paediatric population, although more data with a larger subject group are necessary before any strong conclusions can be drawn regarding this hypothesis.

Case Study 2

C.W. was a two and a half year old boy with a congenitally absent left forearm who attended the OCCC for fitting of a myoelectric prosthesis. In an effort to increase his attention span during biofeedback training prior to receiving his prosthesis, C.W. received practice using specially modified biofeedback equipment which provided feedback in a manner which was enjoyable for him. The University of New Brunswick three-state myoelectric control system (11) employed in his prosthesis was interfaced with an electric train set so that a light contraction of the flexor muscles of his forearm stump correlated with the train moving in one direction and a strong contraction in movement in the opposite direction (See Figure 1). This provided a strong motivation for C.W. to practice fine levels of muscular control for continuous periods of up to one half hour, where previously he had become bored in a matter of minutes when engaged in operating lights or display meters, and in opening or closing a prosthetic hand which was resting on a table surface.

CONCLUSIONS

Further research is necessary in order to fully assess the clinical benefits that biofeedback techniques might bring to the paediatric rehabilitation setting. Work to date has shown that paediatric patients are able to increase their degree of muscular self-control with biofeedback training. There are indications that through the use of feedback which has rewarding qualities, levels of motivation are enhanced in children, thus promoting more frequent and longer periods of practice. Future research will be directed towards the controlled comparison of rewarding feedback with other forms of feedback information in a paediatric population.

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INNOVATIVE GAIT TRAINING WITH KNEE-JOINT POSITION BIOFEEDBACK

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ABSTRACT

Biofeedback training is gaining importance as a treatment modality in physical medicine. This paper presents two case studies in which the concept of knee-joint position training was utilized. The Joint Position Trainer (JPT), an electrogoniometric feedback device, was used to train hemiplegic patients exhibiting excessive knee extension at the stance phase of gait, to maintain a knee-joint angle between 0° (neutral) and 30° (flexion) during weight-bearing. The accurate and immediate feedback of knee-joint position provided by the JPT seems responsible for the positive results obtained in these two case studies.

INTRODUCTION

The restoration of full physical capabilities is a primary goal in the rehabilitation of patients with neuromuscular-skeletal disorders. Unfortunately, full physical restoration is often an impossibility, and therefore the realistic primary goal becomes the attainment of appropriate functional motor control.

Gait treatment regimens such as exercises and neuromuscular facilitation techniques utilize a feedback process involving clinical observation, corrective verbal instructions from the therapist and a patient response. These therapeutic techniques are highly refined and in most cases produce improved performance. However, even in the hands of a skilled professional there are two shortcomings with these usual methods: (a) the subjectivity of assessments, and (b) the time delay between the action and pertinent corrective instructions. Motor learning theorists (1,2) have demonstrated that the timing of instructions following a movement, the clarity of those instructions, and immediate knowledge of performance are crucial if optimal learning is to occur. Using these learning theory criteria, researchers and clinicians have, during the past decade, investigated methods of augmenting existing treatment techniques.

This paper presents two case studies in which an electrogoniometric biofeedback device, the Joint Position Trainer (JPT)(3), was employed to augment the training of correct knee-joint position during weight-bearing (the stance phase of gait). The JPT is designed to facilitate the training of knee-joint position through positional feedback rather than feedback of muscle activity, and it is suitable for use in extensive ongoing gait training programmes which can be conducted in settings other than the clinic. The choice of positional feedback related to knee-joint action may yield advantages over an electromyographic (EMG) biofeedback approach in gait training because:

a) it provides a distinct signal related to the correct phasing of a number of muscle groups, thus reducing the complexity of information that the subject must attend to;

b) it is more directly related to actual function. That is, people are more cognitively aware of joint position than they are of the process of coordinated muscle activity; and

c) it is quick and simple to apply. The identification of individual muscles, electrode preparation and skin preparation are not required, thus increasing time efficiency.

The patients in the case studies presented in this paper exhibited genu recurvatum (full or hyperextension of the knee during the stance phase of gait). Gait training was aimed at achieving a target knee-joint position of between 0° (neutral) and 30° (flexion) during stance phase.

EQUIPMENT

The JPT consists of an electrogoniometer attached to a leg cuff and a processing module which provides continuous or discrete auditory feedback information related to knee-joint position. In the studies reported here feedback was provided only during stance. As shown in Figure 1, adjustable 'windows' result from the setting of two boundaries (limits) so that a tone is provided when the joint motion exceeds either the flexion or extension limit. To obtain objective data, the JPT is connected to a time-event counter (TEC)(4) which records total number of steps, number of excursions beyond set thresholds (errors) and total time.

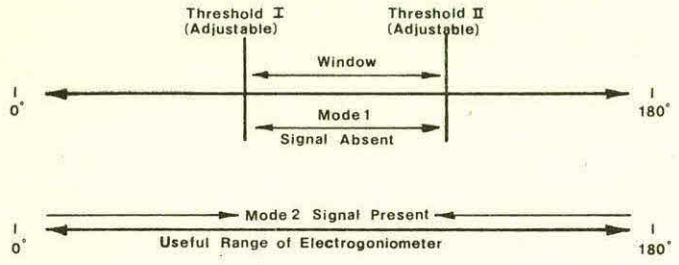

Figure 1: Graphic representation of threshold settings for the Joint Position Trainer.

CASE STUDY I(5)

S.C., a 15-year-old female exhibiting a right hemiparesis resulting from a subarachnoid hemorrhage nine months earlier, was referred for EMG assessment of gait and for right lower limb muscle retraining.

Physical evaluation revealed an independent gait but the right knee exhibited a genu recurvatum of approximately 5°. Sensory abilities (proprioception and touch) were present but assessed to be slightly decreased.

The EMG assessment revealed that the right hamstrings and gastrocnemius muscles were out of phase. In an attempt to correct this gait dysfunction, EMG biofeedback training of these muscles was attempted for seven sessions. Although changes were demonstrated on individual EMG tasks, no overall functional gait improvement occurred. The patient was unable to phase successive patterns of muscle activity accurately on a conscious level. As a result of this information processing problem, it was decided to attempt knee-joint position training with the JPT, which provided a less complicated feedback signal.

Method

An A1-B-A2 single case study design with an 18-month follow-up was used to assess the effectiveness of the knee-joint position feedback training. Independent EMG gait analysis was conducted during baselines (A) and the 18-month follow-up. The JPT training period (B) constituted a home programme of two months duration and was comprised of: a daily 25 step no feedback (NFB) baseline, a 1/2 hour steady walk with biofeedback (FB) and a 1/2 hour period with FB during general activities. Auditory feedback was received if knee extension exceeded the neutral position during stance.

Results

S.C. exhibited a high level of compliance with the programme and required minimal supervision. Gait analysis directly after JPT training (A2) revealed four positive gains and one negative aspect in knee-control performance. The improvements were: increased push-off on the affected side; reduction of some inadvertent EMG activity during swing; more normal heel-off; and a reduction in tendency to hyperextend the knee during mid-stance.

The negative aspect of training with the JPT was manifested as a deceleration of the right thigh and shank at the end of swing, that is, there was decreased 'reach' due to increased hamstring activity.

The 18-month follow-up analysis revealed a regression in muscle control to previous (pre-JPT treatment) levels.

CASE STUDY II(6)

Mrs. D., a 53 year old woman, had suffered a cardio-vascular accident (CVA) resulting in right hemiplegia six months prior to gait training with joint position feedback. She had been receiving physiotherapy treatments for her right upper and lower extremities and gait training prior to this study.

The right leg showed full passive range-of-motion (ROM) with good voluntary movement at the hip and knee but no active movement at the ankle or foot. Proprioceptive, touch and pressure sensory awareness were essentially normal. Moderate spasticity was evident in the right gastrocnemius. Mrs. D. walked with a right short-leg brace and a quad-cane (four-legged cane).

Method

The experimental design employed was an A1-B1-A2-BC-A3-B2-A4 format(7). Each treatment period (B1,BC,B2) was one week in duration and consisted of 10 half-hour sessions. Baselines (A) consisted of five 25-step walks with no feedback. Physiotherapy phases (B1 and B2) involved weight-shifting exercises (forwards/backwards and right/left) and active gait training using a below-knee brace and quad-cane. In treatment week BC the same regimen was followed but with the addition of auditory feedback supplied by the JPT if the knee-joint angle exceeded the preset limits of 0° (neutral) and 30° (flexion) during stance.

Results

Figure 2 graphically depicts the progress made by the patient. Each point on the graph represents the error index (number of genu recurvatum errors/total number of steps) for a 25-step no feedback walk, conducted at the beginning of each session in order to estimate retention of the previous day's training. Phase B1 shows a slight tendency towards improvement but the subsequent baseline (A2) reflects a rapid decay of this improvement. Dramatic improvement in knee control occurred during feedback stage BC and continued to be manifested in B2 although some loss of control occurred. The patient's cadence remained stable throughout the study and progress was made from a quad-cane to a standard cane.

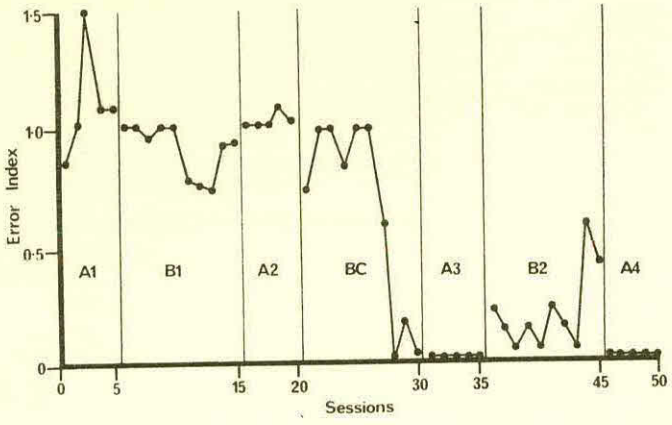

Figure 2: Patient's progress as reflected by changes in error-index during 25-step no-feedback walks.

DISCUSSION AND CONCLUSION

The accurate and immediate feedback of knee-joint position provided by the JPT seems responsible for the positive results demonstrated in these two case studies. The task of concentrating on a functional joint position may be less confusing than attempting to consciously phase successive muscular activity.

The JPT, through the provision of auditory joint position feedback, fulfills several learning criteria designated necessary for optimum learning: immediacy of feedback, simplicity of information and knowledge of performance.

Two challenging issues have arisen as a result of the two case studies discussed. First, marked decay of positive results evident in S.C's 18-month follow-up highlights the issue of the potential usefulness of regular "booster" biofeedback sessions to maintain a desired response. Second, results raise a query as to how much training is required to produce overlearning, a suspected requirement for good response maintenance.

The results of these two case studies emphasize the potential usefulness of joint position biofeedback as a part of the total physiotherapy approach to gait training in hemiplegic patients. This supports the conclusion of Fernando and Basmajian(8) that biofeedback is a useful adjunct to physical rehabilitation.

ACKNOWLEDGEMENTS

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The cooperation of Dr. D.A. Winter, Department of Kinesiology, University of Waterloo, in conducting the gait analysis for Case Study I is greatly appreciated.

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GLOBAL QUANTIFICATION OF NEUROLOGICAL RECOVERY BY A PREVIEW TRACKING TASK

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One of the primary techniques developed for a Quantitative Assessment-Training Centre is a preview tracking task. Generated and monitored by a graphics display computer, it accurately quantifies performance of the upper-limb sensory-motor system. Neurological patients, particularly those having sustained a head injury or stroke, can be assessed at regular intervals allowing recovery graphs to be drawn. Controlled studies using this type of technique should be able to help substantiate the effectiveness of various neurological rehabilitation procedures, such as medicine, physiotherapy, occupational therapy, and the tracking task itself. Clinical application of the tracking task from both an assessment and therapeutic point of view is illustrated with some preliminary patient results.

INTRODUCTION

One of the recurring basic questions in the area of neurological rehabilitation is: To what extent do various medical and rehabilitation procedures influence the recovery process following acute brain damage? That is, when a patient regains neurological function following a stroke or head injury, to what extent do medicine, physiotherapy, occupational therapy, and any other rehabilitation procedure influence the rate or eventual endpoint of neurological recovery. There appears to be at least three major reasons for the continuing relative ignorance regarding this question:

(i) Prevailing lack of understanding of the underlying neurological mechanisms of recovery;
(ii) Distinct lack of the use of quantitative procedures in the area of neurological rehabilitation;
(iii) Very definite practical and methodological difficulties in carrying out longterm controlled clinical trials on brain damaged patients, such as in the standardization of patient treatment in all respects other than the technique under scrutiny.

In an attempt to help alleviate the lack of use of quantitative methods in neurology, a Quantitative Assessment-Training Centre has been set up to act as a focus for development, application, and evaluation of various quantitative techniques (1). In practice, of several techniques developed and applied clinically, a computerized preview tracking task maintains central importance. As a complex visual-motor task it permits accurate global measurement of integrity of the upper-limb sensory-motor system. When applied serially to stroke or head injury patients, long-term performance data is gained from which neurological recovery information can subsequently be derived.

METHOD

System Description

The preview tracking task is generated and analysed by a PDP11 computer with VT11 dynamic graphics display unit. In the standard procedure, the input tracking signal consists of either a 0.14 Hz sine-wave or a 0.21 Hz bandwidth random-wave which moves down the screen with a preview time of 8.0 s. An arrow near bottom of display moves horizontally in response to the patient output transducer which in this case is a steering wheel with variable friction and position sensitivity. During each run the horizontal tracking error is sampled at 28 Hz and on run completion an error graph is generated and displayed. Because the sine-wave input produces a task of constant complexity, it gives direct feedback of run performance. Erratic variations due to physical difficulties or intermittent concentration are immediately obvious as are slower changing tendencies such as learning and fatigue. Although a large number and type of error analyses are possible, the only error measurement used to any extent clinically is the 'mean absolute error' over a run.

Standard Test Procedure

Each session of the standard tracking task program comprises five 120 s runs separated by approximate two minute rest intervals. The standard run protocol adopted comprises one run for each hand on the sine-wave, followed by one run for each individual hand and for both hands on the random-wave. This pattern is then varied so that any dominance effects due to arbitrary run

assignment are eliminated.

A straight assessment referral for what might be essentially static sensory-motor dysfunction consists of two sessions one week apart. If however a patient with recent brain damage is referred, he will be initiated on a longer term series of sessions and on what is generally a three per week basis.

TYPICAL LONG-TERM RESULTS

A wide variety of brain damaged patients have been assessed serially with the tracking task with recoveries being anywhere from substantial, to erratic, or to static. To highlight various aspects of these recoveries, three different graphs have been developed for the presentation of patient data. Illustration of the graphs and their interpretation is best shown by considering the results from a series of eight normal subjects (4 males, 4 females, ages 20 to 57 years, all right-handed, all drivers) and a typical head injury patient:

> W.M. was an 18 year old moderate head injury who was unconscious for two hours as a result of a motor vehicle accident. His main deficits were a moderately dense left-hemiparesis with poor concentration, motivation, and cooperation.

Average Tracking Error (Fig.1.)

The average tracking error is the average of the sine and random mean run errors for either the dominant or non-dominant hand.

Normal results - Surprisingly, there was no significant difference found in tracking performance between the dominant and non-dominant hands. Consequently the normal range does not need to be separated into these two components. A wide variance was found for first session results but this was reduced considerably for the subsequent 8 sessions of the overall 9 session/7 day interval normal trial series.

Patient T.W. results - Grossly abnormal left arm performance was evident (intially left arm error was actually above the 'no-input' error score of 160)although the right arm showed only minimal involvement. This latter point does in fact confirm that there were no major perceptual, cognitive, or motor planning deficits which would have otherwise showed up bilaterally. The left arm then showed rapid and substantial improvement and after 136 days is only slightly subnormal and no worse than the right arm.

Total Performance Increment (Absolute) (Fig.2.)

The total PIA is the total sum over 5 runs of record increments, or improvements, in performance occurring during a particular session.

Normal results - Although the expected initial task learning/familiarisation improvements are very evident, the hoped for existance of a quickly reached and well defined performance

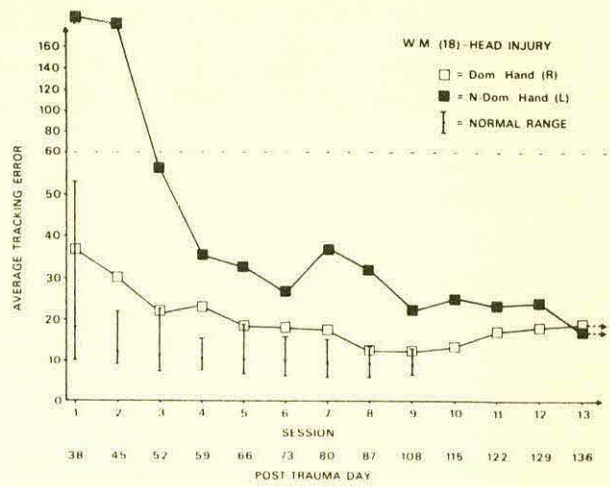

Fig.1. Average Tracking Error.

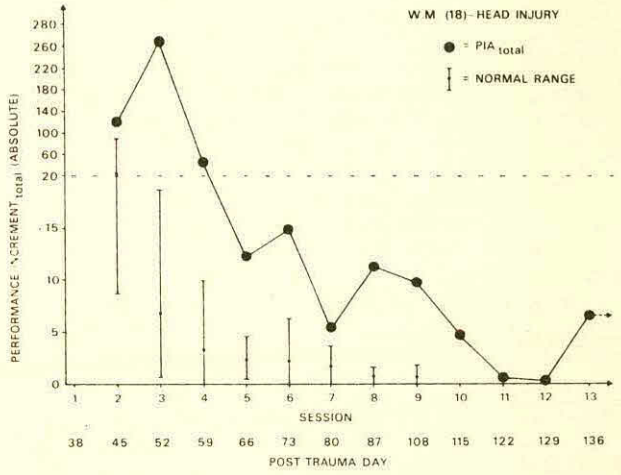

Fig.2. Total Performance Increment (Absolute).

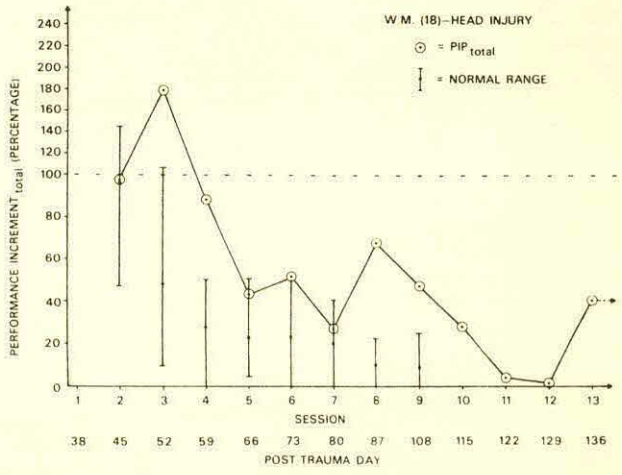

Fig.3. Total Performance Increment (Percentage).

plateau is not. Performance improvements continue, even if to a lesser extent, over the 9 sessions. This does of course complicate the desired differentiation between the recovery of neurological function and of normal learning processes.

Patient W.M. results - Performance improvements significantly above the normal range are seen for nearly all sessions. These indicate that return of temporarily lost neurological function occurred in addition to normal expected learning improvement.

Total Performance Increment (Percentage) (Fig.3.)

One possible deficiency of the PIA measurement is that it can fail to reliably detect static but severely impaired neurological conditions. For example a record improvement from 100 to 98 appears equivalent to 10 to 8 on the PIA graph yet obviously the latter is of much greater significance. By taking the percentage rather than the absolute performance increments, this improvement distortion is eliminated and the PIP's for above become 2 and 20 percent respectively.

Normal results - The result of taking the percentage increments is to expand the tail of the normal range compared with the PIA graph.

Patient W.M. results - Correspondingly, the margin between patient and normal improvements is decreased in the PIP graph. Some PIP results (particularly sessions 5,6 and 7) are now within the normal range whereas they were moderately above PIA normal range, however the other results substantiate even more conclusively recovery of neurological function.

DISCUSSION

Global Assessment

Performance on a tracking task is dependent on a large number of component functions: sensory (visual, somatosensory, hearing), perceptual (visuospatial), motor planning, motor (strength, range of movement, coordination), higher psychological (cognition, concentration, motivation, memory), as well as their integration as required in a continuous complex visual-motor task. Consequently the tracking task provides global information on the integrity of the upper-limb sensory-motor system considered in its widest sense. If improvement occurs in any one of the component functions it will be detected and measured. Conversely it cannot indicate in which of the component functions improvement occurred. This can only be done by the application of a battery of much simpler tests each of which theoretically depend on only one neurological component. However it is considered that with the prudent addition of different error measures, the tracking task itself can be used to provide information on at least some of the individual component functions. Fourier, cross-correlation, direction of error, and other analyses are potential avenues in this direction.

Clinical Application of Tracking Tasks.

Considering the extensive use of tracking tasks in motor skills psychology (2), and their accessability to modern control engineering analysis and modelling, their application to the neurologically disabled subject is surprisingly limited.

Cassell et al (3) and Flowers (4), both used pursuit tracking tasks in limited trials to measure abnormal movements of Parkinsonian patients. Potvin et al (5,6) have developed an extensive tracking task battery which is itself a component of a comprehensive clinical quantitative neurological examination. They use both pursuit and compensatory modes with various tracking input signals and patient output devices. Most of their clinical effort has been directed towards Parkinsonian and multiple sclerosis patients. Lynn et al (7) shows a different emphasis from the above progressive neurological diseases by applying a pursuit tracking task to stroke patients. They have constructed preliminary recovery-type curves for potential use in the validation of rehabilitation management.

Advantages of the Preview Tracking Task

Although the preview tracking task does not appear to have been used clinically elsewhere it does appear to have distinct advantages over the non-preview type. Because of the 'preview' of the future desired course it is more closely allied to most of the sensory-motor activities met in everyday life such as walking, driving, sport, and just the simple manipulation of objects at home and work. Therefore any assessment data from the preview tracking task should have a closer correlation and validity to these tasks.

In addition to its accurate global assessment role the preview tracking task has proved subjectively to have valuable therapeutic attributes. The relearning of sensory-motor control patterns and the increasing of concentration ability are the most obvious of these. These training processes are optimally facilitated by a combination of (i) practice at a complex visual-motor task having a direct correspondence to many ADL tasks, and (ii) maximization of motivation through immediate and accurate performance feedback. These features considerably enhance the preview tasks usefulness in routine rehabilitation.

The inherent therapeutic qualities were in fact very aptly demonstrated by the previously detailed head injury patient. On his first tracking session 38 days after admission, T.W.'s left-arm performance was severely abnormal. He had previously shown extreme non-cooperation towards therapists and refused to put any effort into using his left-arm. After four sessions on the tracking task, his general attitude, motivation, as well as awareness and use of left arm improved considerably. This leaves little doubt that positive feedback of improvement alone can help considerably in promoting the self-rehabilitation process.

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A STUDY OF DAMPING TO IMPROVE MANUAL DEXTERITY OF THE HANDICAPPED

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ABSTRACT

Athetoid or spastic persons may experience considerable difficulty in operating switches, joysticks, keyboards, and other forms of control. The application of controlled amounts of damping to the affected arm enhances the subject's success in performing specified activities. This paper examines three methods of providing damping: frictional, hydraulic, and electrical. The equipment used is described and preliminary results discussed.

INTRODUCTION

Various disease states, such as stroke and cerebral palsy, may impair an individual's ability to control arm and hand function, even when adequate strength exists. Our investigations have been primarily concerned with cerebral palsy, where it is observed that spasticity or athetosis can often preclude effective arm-hand function. Functions lost may include writing, typing, wheelchair control via a joystick, feeding, operation of switches (TV, etc.), and many taken-for-granted manual movements.

In order to replace some of these missing functions, considerable ingenuity is expended on the design of "interfaces" which make the most of the candidates residual functions. These devices may exploit head or foot control instead of hand control; they may take the form of oversize keyboards to reduce the accuracy required; they may be switches of special configuration. Many other variations have been developed to match the patient's abilities.

We had occasion to construct a damped joystick for an assessment technique we had under study. This unit had its motion restrained by air cylinders embodying controlled adjustable leakage. Although this had the usual "bouncy"characteristics of air damping, it nevertheless dramatically improved the function of certain cerebral palsy children. This led to the premise that application of damping to the hand or arm, instead of to the device, might have similar benefits. This approach would then obviate the need to build special controls. The economic benefits are obvious.

As this concept was under study some supporting evidence came to our attention[1,2,3,4,5]. Haworth el al, and Cowan et al, both found that support alone, without damping, had beneficial results. Rosen et al found, as we did, that application of damping to a joystick greatly improved a tracking task. They, however, used hydraulic damping, thus obviating the bounce problem. Ring also used hydraulic damping in a feeding aid, with good results.

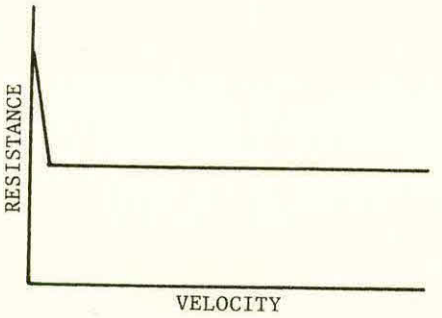
This paper presents our preliminary laboratory results using 3 methods of damping, with somewhat differing properties. It includes test data on a newly-developed rotary hydraulic damper.

DAMPING IN BRIEF

Damping which can be applied to an arm rest or other device for the handicapped can be accomplished in several ways, including gases, magnetic fields, eddy currents, and hysteresis. However, we have examined three techniques: friction, fluid and electromagnetic.

Friction

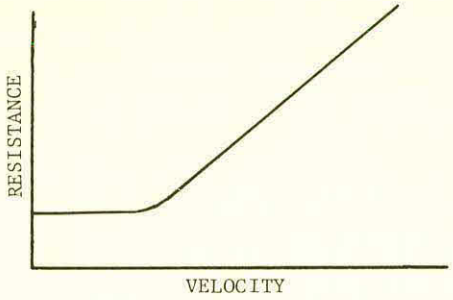
This effect is familiar to anyone who has moved heavy furniture. It has characteristics similar to the sketch:



There is a high static friction at first which drops rapidly to the kinetic or "running" value. The kinetic friction then remains relatively constant as speed increases.

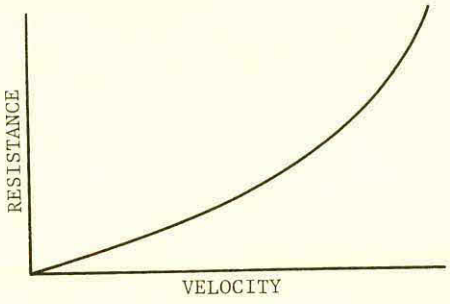
Hydraulic

This type of damping is provided by fluids in shear, or by fluids passing through an orifice. The characteristic of a commercial rotary damper resembles this:



There is an initial resistance to motion followed by an approximately linear increase in resistance with rotational velocity.

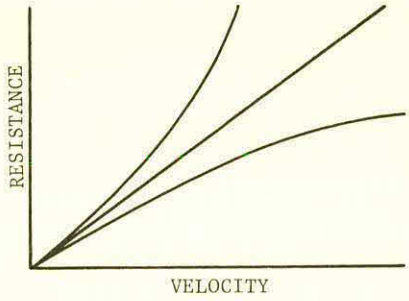
Intuitively, we believe that the characteristic should be more like this:



There is little or no resistance to starting. (At the moment, it is not clear whether the curve should start at the origin, or a little higher). The resistance thereafter increases more and more rapidly with speed.

Electrical

We have constructed an experimental setup in which an electric motor provides the opposing torque. Speed of rotation is detected and used to control the power applied to the motor. While too complex for routine use, this system is valuable for determining the best damping characteristics. Since the mechanism is controlled electronically, the use of non-linear amplifiers makes it possible to achieve curves of various shapes, as shown in the following sketch. By adjusting the gain, the slope of the curve can be changed as well as its shape. This is the only system with no discernible resistance to starting (unless friction is deliberately added).



PRELIMINARY RESULTS

Friction

A commercial forearm support was purchased, which incorporates dry friction elements which are adjustable. This device is known by various names, such as "ball-bearing feeder". It was modified for children (Fig. 1). Its characteristics are shown in Fig. 2. The two curves

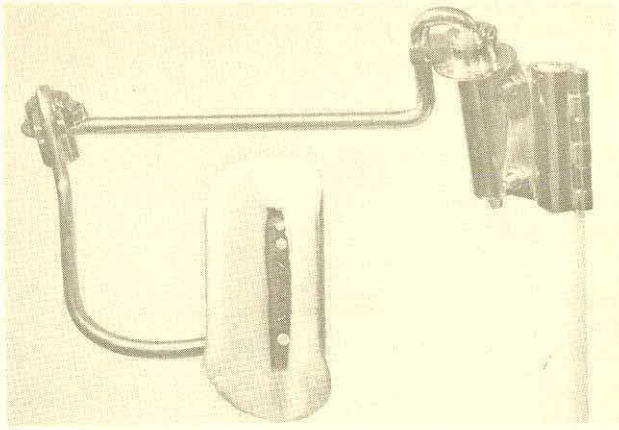


Fig. 1 A forearm support with friction damping.

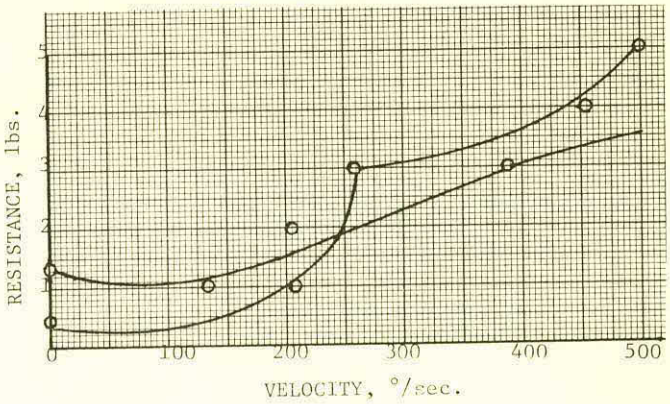


Fig. 2 Some characteristics of a commercial friction damper.

represent two different settings. These curves were taken in some haste on a quickly-assembled rig, and the somewhat odd shapes may be attributed to this. However, the basic trend is for only a small change in resistance as velocity increases.

Preliminary studies have been carried out, to obtain an idea of the value of damping applied to the user's forearm rather than to the controlled device. A cerebral palsy candidate was selected, who had no ability with pencil or crayon. It was observed that support of the forearm alone enabled her to grasp a crayon effectively - previously impossible. The addition of damping permitted some success with a basic drawing task (tracing a circle). The improvement was dramatic.

A study is currently underway in collaboration with the Ottawa Crippled Children's Treatment Centre. The test apparatus has five very large push buttons connected to a timer. On a cue, the subject brings his hand from a rest position to the target area, thus stopping the timer. The initial part of the study is examining the effect of audio location on the candidate's performance. A second phase studies the effect of the friction-damped support on speed and accuracy in this test.

Hydraulic

We have found only one manufacturer of small rotary hydraulic dampers. These, unfortunately, have only a limited arc of rotation (about 40°). This is inadequate in some cases. An example of such a case is the operation of a very wide expanded keyboard. Characteristic curves for a unit are shown in Fig. 3. The starting friction is less than expected, but the general linearity of the characteristic is shown. The three curves represent three different settings of the adjustable device. Note that the scale of the abscissa is changed. This device is very much "stiffer" than the other units discussed here.

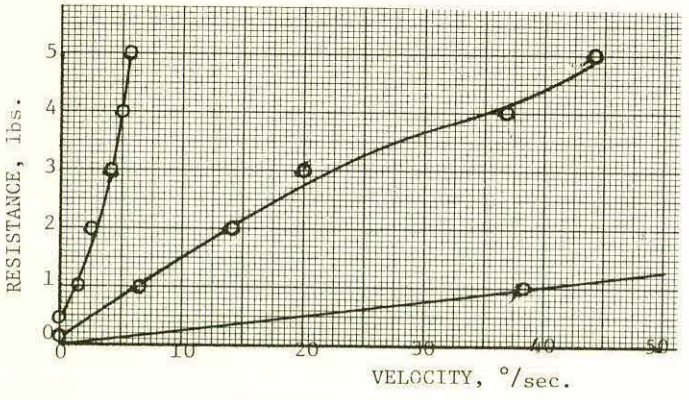


Fig. 3 Some curves for a commercial hydraulic damper.

We have developed a new damper which has a large arc of rotation (about 270°), and a "steeper" characteristic. Test results are shown in Fig. 4. The device is now being fitted to a commercial arm support for clinical trials.

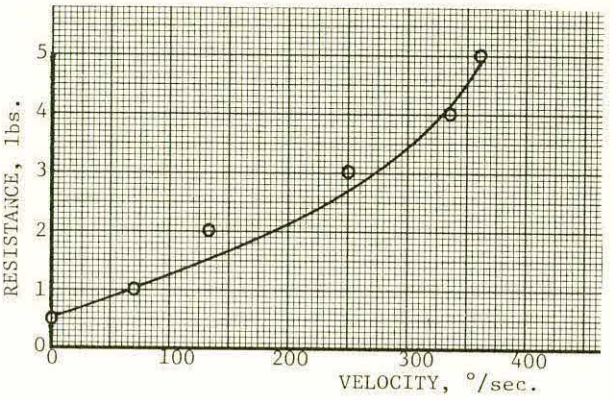


Fig. 4 A test result on a prototype hydraulic damper.

Electrical

As mentioned, this technique is not suitable for routine use, but is valuable for determining the best characteristics for a damping system. The circuit diagram is shown in Fig. 5, and the characteristics in Fig. 6. Modifications to the curve shape have not been made at the time of writing. Modifications in curve slope are easily made by an external potentiometer. A complete support system incorporating this is now under construction.

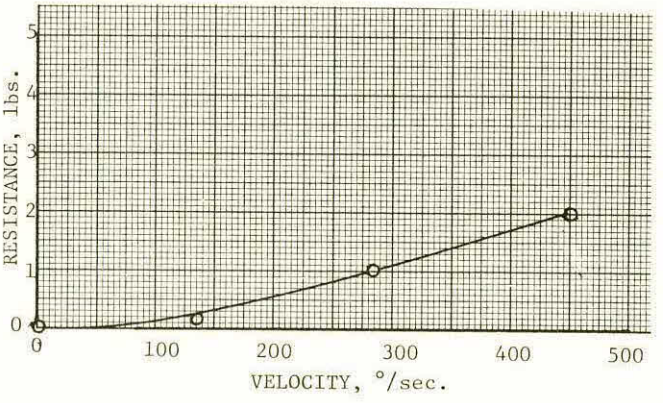


Fig. 6 An example of the electrical system's characteristics.

CONCLUSION

Preliminary results have indicated that damping of the forearm is as effective as damping of the controlled device. This has the obvious benefit that many patients will require only one special aid - the damped support. More severely

involved patients will effectively use conventional interfaces without the complication of damping on each interface.

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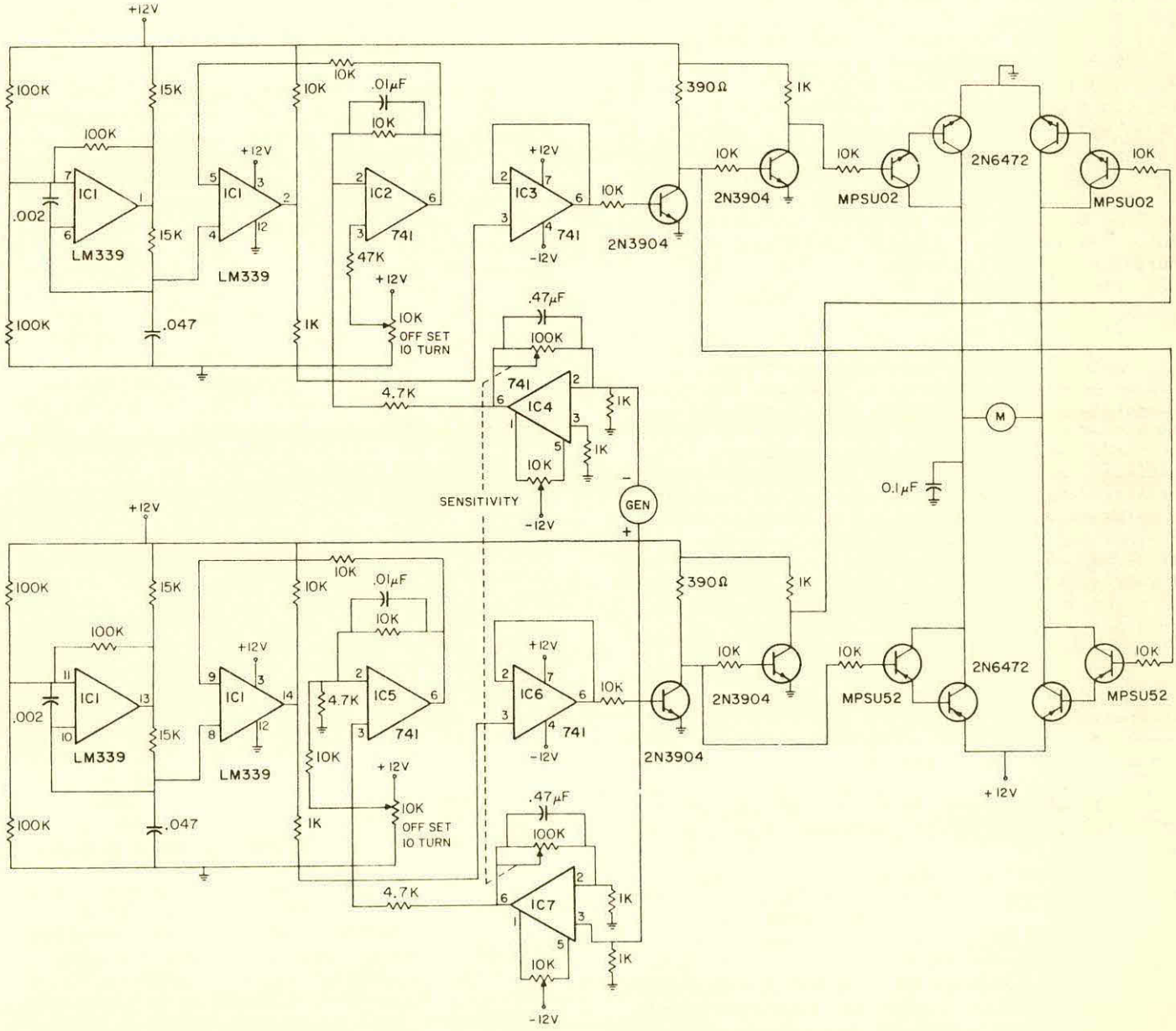

Fig. 5 The circuit of the electrical restraint system.

ATTENUATION OF ABNORMAL INTENTION TREMOR FOLLOWING VISCOUS EXERCISE: WORK IN PROGRESS

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It is hypothesized that abnormal intention tremor may be selectively attenuated by application of viscous loads. Data taken in experiments being conducted to test this concept have revealed an unexpected carryover of reduction of wrist tremor following removal of the load. The possibility that movement against a damper serves as a therapeutic exercise for tremor-disabled individuals is now being verified experimentally. The apparatus employed is entirely mechanical and is easily fabricated from commercially available components. It is based on a rotary damper with a handle, providing viscous opposition to hand extension-flexion. It is designed to facilitate application of a controlled exercise regime and graphical measurement of tremor without technical assistance. Data is presently being collected from a subject with hereditary essential tremor.

INTRODUCTION

Individuals who have sustained cerebellar injury, basal ganglion lesions, Multiple Sclerosis, Friedreich's Ataxia and familial neurological disorders are often disabled by abnormal "intention tremor". This large amplitude involuntary limb oscillation is characterized by its occurrence primarily during voluntary movement or postural maintenance. The degradation of function inherent in this feature is readily apparent and the loss of useful limb performance may be complete if the tremor amplitude is large enough to obscure intended movement. Conventional physical therapeutic, surgical and pharmacological treatment is often of little or unpredictable value (1,2).

In response to this problem, work has been underway for some time at the Harvard Medical School-M.I.T. Rehabilitation Engineering Center intended to test the hypothesized effectiveness of applying mechanical loads to attenuate tremor selectively with respect to purposeful movement(3). While the long range clinical goal of the investigation is the development of multi-degree-of-freedom orthoses and special purpose control interfaces, present objective experiments measure the effect of viscous damping on tremor of wrist extension-flexion. This type of load is characterized by opposing movement with a force proportional to the speed of the movement and is commonly observed in door dampers and automobile shock absorbers. Results from several subjects with diverse etiologies have been hopeful, often showing dramatic decreases in tremor at load settings which have little effect on the performance of voluntary tracking movements(4). An early clinical spinoff of this work is the ongoing development of a damped joystick (presented at last year's Interagency Conference) for improved input to environmental control, mobility and communication devices by tremor-disabled people(5).

An unexpected result observed in tremor measurements obtained from 3 subjects formed the basis of the work to be reported here: Experimental trials were organized in sets of seven. An initial tracking trial was performed with no damping applied, followed by trials with five different load settings in random order. The set was completed with a final trial with no damping. It was noted that a statistically significant reduction of tremor was observed in the final undamped run relative to the level measured initially. Apparently, the effect of repeated applications of viscous loads during voluntary movement was a decrease of tremor which persisted after the load was removed. This result is intriguing, not only because it suggests a possible therapeutic exercise mode, but also because one might reasonably have predicted the opposite effect. Given the well-documented ability of neuro-muscular control systems to adapt to and compensate for environmental changes, it might be expected that an applied load would have gradually decreasing effectiveness followed by a worse-than-usual "rebound" of tremor upon its withdrawal.

Several questions need to be answered. Is this result repeatable? If so, how long does the effect persist following exercise? Does the nature of the exercise - movement against springs or weights as opposed to viscous loading - really have a bearing on the reduction in post-exercise tremor? Is there a cumulative improvement from regular performance of the exercise? What is the physiological mechanism of this reduction? The experiment reported here is intended simply to verify the repeatability and measure the approximate magnitude, persistence and longer term change of the effect. The testing apparatus is designed for simplicity of fabrication and ease of use so that daily sessions may be self-administered in a home or clinical setting. While data-taking and analysis are far from complete, work-in-progress is presented below in part to encourage rehabilitation engineers and clinical practitioners to undertake similar or complementary studies.

EXPERIMENTAL TECHNIQUE

Our apparatus consists of two major components, the viscous exerciser and the self-testing easel, shown in Figures 1 and 2 respectively. The exerciser is based on an Efdyn S-CRD-30,000 rotary damper. This passive load produces a torque in opposition to rotation of its shaft approximately proportional to angular speed. An adjustment knob mechanically alters the space between the two internal surfaces varying the effective damping constant by a ratio of about 1:10. With the adjustment set for maximum damping, the unit produces a damping constant between 2.5 and 5 lbf/(rad/sec) in the range of velocities of interest. As seen in Figure 1, the damper is mounted with its shaft horizontal, coupled to a simple tubular L-shaped handle which can be adjusted so that a natural grasp of the handle aligns the axis of rotation of the wrist with the damper's axis. Also attached to the shaft are two lever-like components which limit the travel of the handle in extension and flexion and actuate a small mechanical counter. The latter, a three digit unit available from several manufacturers, increases its count by one when the handle nears the bottom of its travel and thereby indicates the total number of extension-flexion cycles in a trial. The counter is easily reset to zero at the start of each exercise session.

This exercise apparatus also includes a comfortable forearm restraint padded with "Temperfoam" (Beckton-Dickinson), a material which, unlike conventional foam rubber which is spring-like in its behavior, behaves viscoelastically. In other words, it changes its shape slowly when forces are abruptly applied to or removed from it. The suitability of this material for restraining a limb in which tremor is present can be demonstrated on both experimental and theoretical grounds.

The self-testing apparatus shown in Figure 2 consists of a simple block of Temperfoam intended to restrain the forearm and raise it off the table surface, and an easel which presents many copies of the pattern to be traced. Since the present experiment is intended to verify the post-loading tremor reduction apparent in earlier sinusoidal tracking experiments, the target patterns are sinusoids requiring approximately 40° of extension and flexion. To trace a pattern, the subject holds a nylon-tipped pen with a plastic cross piece as shown in Figure 3. While this is certainly an unaccustomed manner in which to draw or write, it requires the same extension-flexion movement of the hand with the forearm pronated as the exercise unit, thereby testing movement accuracy in the same muscles which have been exercised. An assistant moves the easel laterally at a roughly constant speed, creating, in effect, a vertically moving target in front of the subject's fixed forearm. The total parts and materials cost of the apparatus is approximately $160, of which the damper accounts for $120.

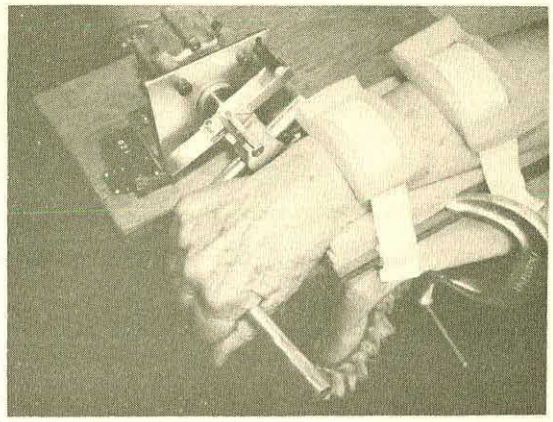

Figure 1- Viscous exerciser. Note rotary damper, counter, travel-limiting stops and arm restraint.

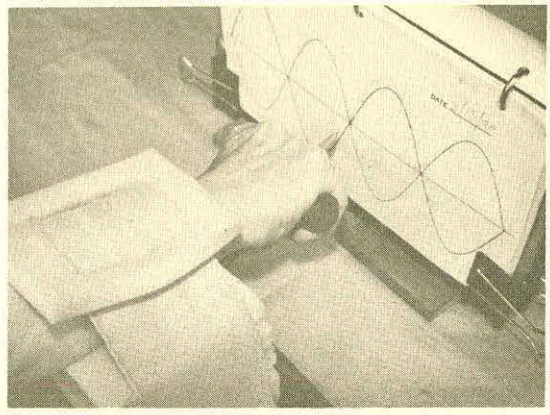

Figure 2- Self-testing apparatus showing sinusoidal tracing patterns and arm restraint.

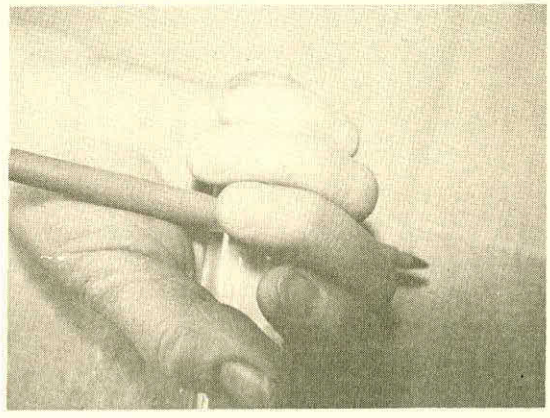

Figure 3- Pen holder. This permits pattern tracing by wrist extension-flexion movements.

The present protocol is as follows: at a roughly fixed time of day, the subject begins a session by making two tracings each of the solid and dashed sinusoid (seen in Figures 2 and 3). This is followed by 60 cycles at the rate of 40 cycles/min on the viscous exerciser. This rhythm is guided by an electronic metronome and was chosen to be within the range of tracking frequencies employed in the loading experiments described above. The pattern tracing task is then repeated immediately and one-half hour later. Pattern sheets are dated and numbered by the assistant (in this case the subject's husband) for identification during subsequent data analysis. The total commitment of time each day is approximately five minutes during a period of over thirty minutes. The protocol is to be repeated daily for four weeks.

DATA ANALYSIS

Clearly, a laboratory-restricted apparatus could be based more conventionally on an oscilloscope display of target and response, goniometric recording of wrist angle, magnetic tape-recording of the error signal, and on-line data analysis. Due to the self-administered character of the present protocol, graphical analysis of pattern tracings is required to extract the subject's tracking error as a measure of tremor. (This is similar in concept to the standard motor accuracy tests administered by physical and occupational therapists.) Reduction of data from graphical to numerical form will be simplified by the use of a "digitizing tablet" and a digital computing facility. The tablet, a specialized computer-input device, allows entry of the subjects' collected sinusoid tracings into the computer by re-tracing them with a specialized stylus whose position in two dimensions on the tablet surface is automatically detected. It should be noted that a manual scoring technique employing a clear plastic overlay with numbered error ranges may be employed in the absence of a computer-based data reduction facility.

At present, data is being collected from subject F.S., a 76 year-old woman who presents a familial action tremor at the wrist but is otherwise healthy. Her major functional problem due to the tremor is difficulty producing legible handwriting. If results of this experiment verify our earlier observations of tremor reduction following viscous exercise, a technique may be developed for her to permit reduction of tremor in writing-related muscles prior to undertaking everday writing tasks. Collection of similar and complementary data on tremor-reducing exercise and its mechanism is also planned. Simple physical therapeutic approaches to management of tremor which are supported by our data will be published in journals directed particularly at clinicians.

ACKNOWLEDGMENTS

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TOPICAL ANESTHESIA IN THE TREATMENT OF SPASTICITY

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ABSTRACT

Effect of skin receptors discharge on the α and γ-motoneuron excitability was measured in normal subjects. Topical anesthesia was applied on the different skin areas and dermatomes, and H-reflex and Achilles Tendon reflex (ATR) were measured over a period of time. Most of the anesthetized skin areas and dermatomes induced a significant facilitation on the H-reflex and a significant inhibition on the ATR. It is postulated that there is an ongoing inhibition from the skin sensors on the α-motoneurons, which when blocked may result in an increase in the α-motoneuron excitability.

Application of the topical anesthesia in one spastic hemiplegic patient resulted in substantial reduction in the muscle tone, and improvement in the gait pattern towards normality. Neurophysiological studies indicate that this could be due to α-γ-motoneuron interaction. More detailed clinical studies currently in progress show encouraging results.

INTRODUCTION

Previous work by Hagbarth (1960) and Eldred and Hagbarth (1954) has indicated that noxious skin stimulation has a potential effect on either α or γ motoneuron discharge. More recently Sabbahi and Sedgwick (1976) have shown that natural stimuli also have a demonstrable effect on the H-reflex response of normals. Based on these reported results we have conducted an investigation on the effect of topical anesthesia on the motoneuron excitability in normal subjects, the findings made in this study led to a clinical investigation on the reduction of spasticity in patients with cerebral vascular accidents and M.S..

MATERIAL AND METHODS

Study on Normals

Fifty-eight normal subjects whose ages range from 19 to 53 years were tested in this study. The subjects were asked to lie prone on a padded table and the ankle joint was secured in a specially designed apparatus which enabled the foot to rest freely at the end of the table. In three subjects the calf muscle was slightly stretched by placing the foot in mid-dorsiflexion. This was done to set a higher threshold level for the motoneuron pool.

The H-reflex and the Achilles Tendon reflex (ATR) were obtained. The H-reflex was recorded by two electrodes spaced 2 cm apart and located over the soleus muscle. The posterior tibial nerve was stimulated with 1 msec. pulses at a frequency of 0.2 Hz. The ATR was elicited by an electrically activated solenoid plunger which delivered a constant force impact to the tendon every 10 seconds. The response was recorded via the same electrodes.

Control measurements of both reflexes were obtained at the beginning of each experiment. Then, topical anesthesia (20% Benzocaine) was sprayed for 20 seconds to various skin areas and dermatomes of the lower limb. The responses of the reflexes were recorded at intervals. In three subjects the complete lower limb was sprayed with the anesthesia in order to observe any summation effect. The peak-to-peak amplitude of the H- and AT-reflexes were measured and averaged over 10 consecutive stimulations. The mean of the reflexes amplitudes obtained after the application of the topical anesthesia was compared to that of the control values (before the topical anesthesia) with a t-test.

Study on Patient

A 34 year old male patient afflicted with left pure motor hemiparesis volunteered for the study. A gait study consisting of knee and ankle joint displacements was performed using a Selspot System of Selcom AB. Gait measurements were taken before and after the application of the topical anesthesia to the skin of the affected leg and thigh. Furthermore, active free range of movement in the hip, knee and ankle joints was tested before and 10 minutes after the application of the anesthesia.

RESULTS

The results showed that topical anesthesia applied to the anterior tibial skin and S_1 dermatome produced either mild inhibition or no significant changes in the H-reflex while the application of the anesthesia to other skin areas and dermatomes produced significant facilitation. The value of the grand mean of H-reflex obtained

30 minutes after the application of the anesthesia ranged from 140 to 230% of the control value. In some subjects the H-reflex amplitude was six times that of the control value. In those cases where the recording was continued for 40 or 50 minutes after the application of the anesthesia, the amplitude of the H-reflex continued to increase. When the anesthesia was applied on the whole lower limb no summation effect was noticed and the degree of reflex facilitation was within the usual results obtained on single area or dermatome.

The ATR showed a statistically significant reduction in amplitude in most subjects. This was noticed in both the free resting foot and in the slightly stretched foot. This implies that the effect of skin hypothesia is not governed by the threshold level of the motoneuron pool. However, it is important to note that the degree of reflex inhibition in the ATR was not as great as the H-reflex facilitation. This implies an interaction in the α and γ motoneuron excitabilities with predominant reduction in the γ-drive.

In the stroke patient application of the topical anesthesia resulted in substantial reduction in the muscle tone, an increase in the active range of movement in the ankle dorsiflexion and eversion and hip flexion movement from 10° to 25°. Also, there was improvement in the normal movement pattern and functional activity of the limb. In the gait study, the ankle joint showed a substantial shift toward dorsiflexion (Fig. 1) and eversion while the knee joint showed full extension with free flexion movement, 16 minutes post-anesthesia (Fig. 2). The patient reported that this improvement in movement pattern continued for three to four hours post-anesthesia. The H-reflex showed significant facilitation post-anesthesia although it was not as large as those of normal subjects. The ATR showed mild inhibition post-anesthesia.

TREATMENT

Application of the topical anesthesia in conjunction with routine physical therapy for 13 sessions resulted in substantial progression toward normal walking pattern of the hemiplegic patient. This was not observed when the same physical therapy was applied solely without anesthesia in prior treatment. It was noticed that when topical anesthesia was applied 30 minutes prior to the physical therapy session, the patient was more able to relax his muscles, to overcome more resistance in exercises, and gain more control of his lower limb during volitional activity.

DISCUSSION

The H-reflex may be considered to be a reliable measure for the excitability of the α motoneurons, while the ATR may be considered to be a measure of both α and γ motoneurons excitabilities (Matthews 1070). Therefore, it appears that the decreased sensitivity of the skin receptors (induced by topical anesthesia) causes an increase in the α motoneuron excitability. Other investigators have also noted modifications in the motor output with varying excitability of the skin sensors. Hagbarth and Finer (1963) demonstrated a significant effect of skin stimuli on the EMG signal and flexor reflex in man. A direct evidence was recorded in cats by Hagbarth (1952) and showed that some skin areas are excitatory while others are inhibitory to the extensor motoneurons. Also, Sabbahi and Sedgwick (1976) reported an inhibition in the H-reflex in man with skin stimulation. Our data points towards the conjecture that there is an on-going inhibitory effect from the skin on the motoneuron discharge, a reduction of which may result in an increase in the activity of the motoneuron pool. The mechanism by which the topical anesthesia induces the facilitation of the H-reflex is still unknown. Segmental and suprasegmental mechanisms could be involved. However, the fact that L_5 S_1 dermatomes, the roots which supply the soleus muscle, and the anterior tibial skin showed either H-reflex inhibition or no significant changes implies a segmental effect of the skin on the soleus motoneuron pool. The inhibition observed in the ATR may indicate a reduction in the discharge of the γ-motoneurons after skin hyposthesia. Direct evidence for the effect of skin on γ-motoneuron discharge was reported by Eldred and Hagbarth (1954). Again, the mechanism of inhibition could be caused by segmental and suprasegmental pathways.

Study on Patient

The previous discussion could be used to explain the results noticed in the stroke patient after anesthesia. Over-activity of the γ-motoneurons has been reported to be a primary cause of muscle hypertonia in spasticity (Bishop, 1977). A reduction in the γ-motoneuron discharge induced by skin anesthesia could be the main cause of the reduced muscle hypertonia. The α-motoneurons appear to compensate for the reduction in the excitability of the γ-motoneurons by increasing their discharge as shown by the increased H-reflex amplitude.

The initial results indicate that the use of topical anesthesia may provide a potent new rehabilitation technique for temporarily and quickly reducing muscle spasticity which would enable the patient to perform exercise programs which he could not otherwise perform.

FOLLOW-UP

The topical anesthesia technique was tested on three additional patients afflicted with spasticity in the lower and upper limbs due to head injury, stroke and M.S.. In both limbs an increase in the active range of joint movements was noted. Also, there was an improvement in the functional movement pattern of the limbs. All three patients stated that after the application of the topical anesthesia the muscles and limbs felt looser. Treatment and continued investigation of these patients are still in progress.

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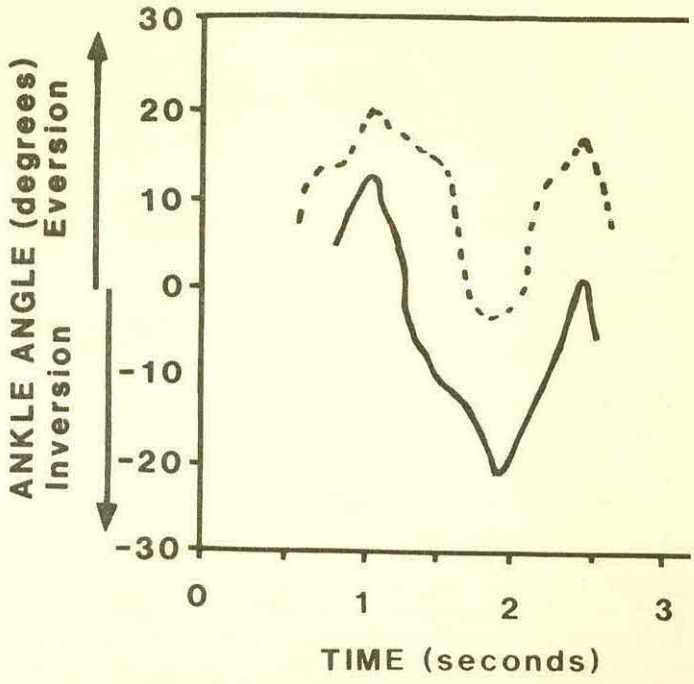

Fig. 1. Angular displacement of the ankle joint before (solid) and 16 minutes after (dash) application of topical anesthesia to the skin of the leg. Note the movement shift towards normal eversion.

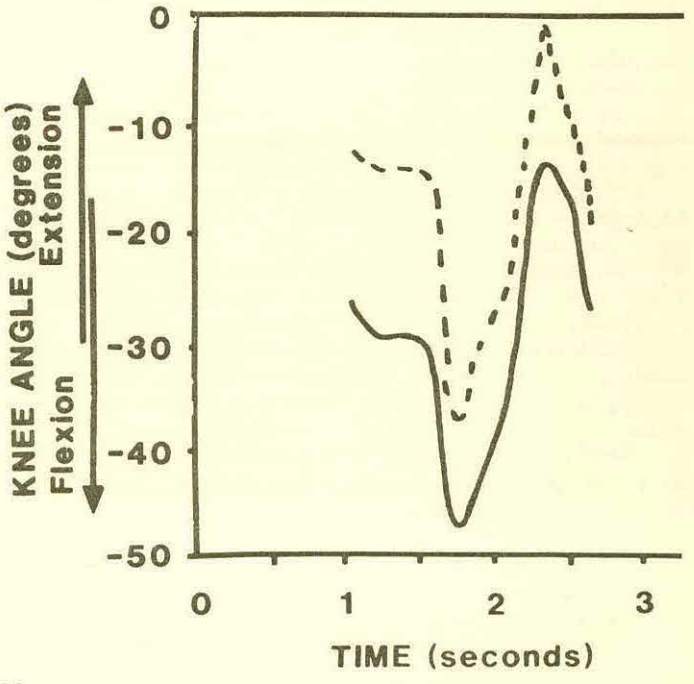

Fig. 2. Angular displacement of the knee joint before (solid) and after (dash) application of topical anesthesia. Measurements were taken after the anesthesia had been applied to the skin surface of the thigh and leg (13 minutes for the thigh and 45 minutes for the leg). Note the movement shift towards normal extension.

THE OCULOMOTOR CONTROL MECHANISM OF CEREBRAL PALSIED CHILDREN

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ABSTRACT

Oculomotor disorders are frequently observed in cerebral palsied children and in many children with "minimal brain damage". These disorders are often associated with an abnormal perceptual - cognitive function which retards the development of the learning process. It has been suggested that when head-vestibular-ocular coordination fails, it can result in a reading disability; when eye-head-hand coordination fails, it can result in writing disability; when both coordinations are disturbed, reading and writing disabilities can occur. Clinically, impeded head control is also observed in many cerebral palsied children. The lack of isolated head movements of cerebral palsied children is essentially related to their abnormal postural mechanism and sensory motor development.

INTRODUCTION

An effective remedial program for visual perceptual skills cannot be instituted without a complete and precise identification of visual system disorders.[1,2,3,4] Existing visual-perceptual batteries require the synthesis of more than one aspect of sensory motor function; that is, visual, auditory, perceptual motor, and cognitive functions. The children with reading and writing problems, and unsatisfactory performance of activities of daily living often begin their remedial program with an assumption that their visual system is normal. This may not be true at all. Thus a need exists to separate visual and oculomotor dysfunction from perceptual dysfunction. In other words, one needs a separate examination of:

1) Optometric function (visual acuity and visual organs).
2) Oculomotor function (can the child move the eyes efficiently to obtain precise visual information?)
3) Visual perceptual function (can the child adequately interpret the visual information?)
4) Visual motor function (can the child translate the visual information into a proper motion?)
5) Cognition (does the child have adequate attention span and memory to retain the visual information to be processed?)

In May 1978, it was decided to embark on a pilot study of the oculomotor function in children, specifically the physiological and clinical analysis of the oculomotor control mechanism of cerebral palsied children, with the following objectives:

1) To determine the oculomotor performance of cerebral palsied children by quantitative analysis.
2) To examine the eye-head coordination pattern.
3) To compare their performance with that of normal children and to determine oculomotor mechanism disorders.

EQUIPMENT AND METHODS

The testing apparatus is similar to that of Funk and Anderson,[5] in order to allow comparison. The horizontal eye-in-head angle is recorded by standard electrooculography (E.O.G.) which consists of affixing 3 small silver-silver-chloride electrodes at the outer canthers of each eye and one ground electrode on the forehead between the eyebrows. The E.O.G. is then amplified and filtered by a medical grade amplifier with a pass-band of DC to 100 Hz.

The horizontal head rotation angle is measured by a low torque potentiometer mounted on a helmet which maintains the transducer in line with the axis of the head rotation. This device produces a DC voltage proportional to the horizontal angle of the head from centre gaze. The resolution is 1 degree. The helmet is packed with sponges in the inner lining to prevent slippage.

A dual channel chart recorder records both of these signals simultaneously at a speed of 125 mm/sec. which permits a greater accuracy in the interpretation of the results. A third pen on the edge of the paper records the "on" time of the targets.

The 3" x 4" rectangular targets are mounted on a semicircular board with a radius of 3 feet with the child at center. The semicircular board maintains a constant focal distance. The targets are mounted at fixed positions of 20, 40 and 60 degrees each side of the center target. The target sequences, which are the same for all the children, are controlled by an Intel microprocessor which also controls the on time of each target and produces negligible time between target switching. Presently a data acquisition system is being implemented in which the data will be stored on magnetic tape in a form suitable for entry into

a mini-computer for analysis. Therefore, comparison of the results of the individual cases will be instantaneously available along with any other pertinent data.

Ideally, the target population should be all cerebral palsied children with a mentality within the normal limits (I.Q. above 80). However, practical considerations limit the scope of our study to children that meet the following criteria:

1) Ages between 6 years and 13 years old.
2) Verbal I.Q. within normal limits (above 80).
3) Ability to maintain an upright sitting position without any physical support for more than 15 minutes.
4) A fair head control to maintain stabilization of the head in alignment with the trunk and to move it independently to scan the 120 degrees span of the visual board.

Presently, no standardization of the head control of these children is available. Thus the evaluation of the head control will be based on a subjective judgement. However, the examination will be administered by one investigator for consistency.

This, of course, may not be a random sample of our target population, since there may be specific characteristics of CP children of the Ottawa-Carleton region that differ from other CP children. However, in the absence of knowledge regarding such disparities we are assuming they represent our target population fairly well.

It is intended to include a sample of 100 CP children in the study. This is primarily based on the projected number of children meeting the criteria that would be available during the study. It seems that at least this number is needed to obtain a reasonably accurate estimate of the regression line. An estimate of the B-errors in the statistical analysis has not been done when comparing these results with those of normal children from the literature. It is not known definitely whether the data available in the literature for normal children (which are mainly U.S. children) are reproducible in Canadian children. Therefore 40 mentally normal children (I.Q. above 80) of the same age group and with no known physical disabilities will also be studied. The results from this group will then be compared with those of the CP children. If, after the present study is completed, it becomes evident that the sensitivity of the statistical tests can be improved by increasing the sample size, we may extend the study.

To date 16 CP children have been tested: seven girls and nine boys. The chronological ages of the children were 6.0 to 13.2 years. In terms of diagnosis: 9 spastic diparesis, 4 spastic quadriparesis, 3 left hemiparesis, and 1 right hemiparesis. Four children in the diparesis and quadriparesis group had a mild strabismus of either right eye or left eye and two children were near-sighted and wearing glasses.

All children had a verbal I.Q. within normal limits. All had a fair head control and were able to sit upright in a wheelchair and use their eyes and head to look at the illuminated targets.

The data collected provides the following results:

1) Mean saccade duration vs amplitude with head stationary.
2) Mean saccade duration vs amplitude with free movement of head.
3) Mean reaction time with and without the head movement.
4) Corrective eye movement (onset, duration and amplitude).

TASKS

The child is seated in a chair that is positioned at the center of radius of the semi-circular target board. The target height is adjustable to compensate for various heights of children. The room lights are then dimmed and noise is kept to a minimum. The child is then given time to adapt to the surroundings during which time the tasks are explained.

Three tasks are performed by the child. The first is a short sequence used to calibrate the equipment. The second task is the eye-head coordination test. The child is asked to move his head and eyes in a normal manner and to maintain the fixation on the target until it disappears (3 seconds). Thirty-one sequences requiring eye movement from 20 to 100 degrees are included in this randomly chosen sequence. Every third target is the center, in order to check on drift of the electrodes and the amplifier.

For the third task, the child's head is held stationary. The span of the illuminated targets is limited to ± 40 degrees since the eyes do not normally travel a full 60 degrees without head movement. Thirteen targets are presented, again in a random sequence.

RESULTS

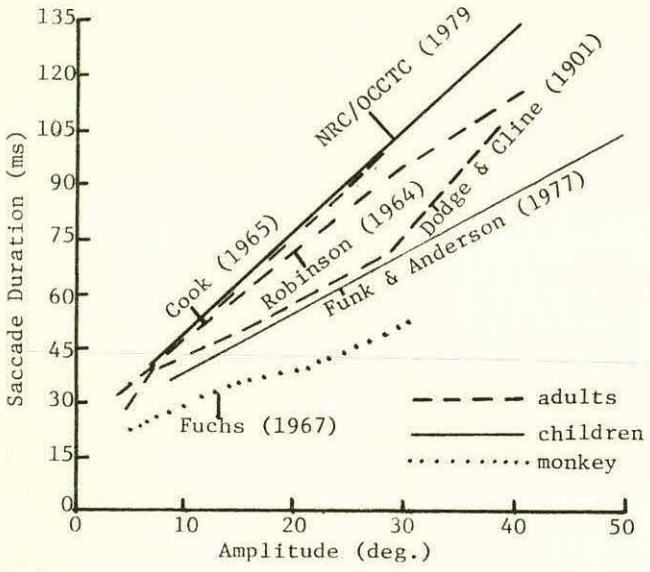
From the results of the (stationary) head task, plots of the mean duration vs amplitude are constructed and the linear regression line is calculated. This linear regression is compared with the results of the existing studies and will also be compared to that of the normal children used as a control group.

The results of the free head, the linear regression of mean duration and amplitude is calculated. In addition the corrective movement (i.e. onset, duration, amplitude) is also calculated and the results are compared with that of the normal group.

Mean reaction time without head movement is calculated and this will be used to determine the discrepancy between the established norm and the records of the CP children.

Analysis of Data

Plots of mean saccade duration vs amplitude of eye movement with the head stationary were constructed and the linear regression line was estimated. There was an apparent linear relationship. The regression line for all the children is shown in the graph.



Comparison of mean saccade amplitude-duration relationship for cerebral palsied children to those reported previously for normal children and adult humans.

DISCUSSION

When the data of the individual children are looked at, the regression lines are very much different. The slope varies between 0.51 and 3.78 and the intercepts between -33.1 and 78.8.

This suggests that there is a lot of variation between children, and therefore to obtain a reasonable estimate of regression line for all children, a very large sample will be required. Otherwise, such a large variation should be explained by other factors. For example, a number of records had to be discarded because the eyes of some children drifted off the target before the illumination of the next target. These drifts of the eye "contaminated" the magnitude of eye movement in relation to the target position. This may be a result of the child's inability to fixate the eye on the target. However, it also appeared that behaviour factors, such as distractibility, impulsiveness or lack of attentiveness were interfering with the measurement of the raw eye movement.

The characteristics of the children's eye movements were inspected in detail:

1) Reaction Time

There was a considerable variation within each target amplitude. As a result, no significant relationship between visual target and reaction time were detected. A significantly long reaction time was noted in seven children. It may suggest an inadequate central processing. A very fast initiation time was also noted in three children.

2) The shape of eye movement in acquiring the target in relation to the duration of the saccades

With the head held fixed, 3 children showed inconsistent and abrupt decelerations. With the head free to move, two children showed long gradual decelerations with a series of short durations and small amplitudes. Four children exhibited overshooting so that the target fell within an intact area of retinal projection. In two children with strabismus, compensatory eye movement (with the head free to move) generated randomly a slow scanning movement off the target and then a return to the fixation point. This appeared to be a random search for additional visual information.

3) Number of saccades required to acquire a target

With the head immobilized, three children made more than one corrective saccade to acquire a target with a span of 20 degrees and 40 degrees, indicating poor accuracy. For the compensatory eye movement, as a result of vestibulo-ocular feedback, 6 children required 4 to 5 corrective movements. The other children made only 1 to 2 corrective eye movements to acquire the foveal fixation. The onset of the corrective movements showed a wide range within the same subject.

4) Binocularity of eye movement

There was no appreciable right and left difference in the children, including the hemiparesis group.

5) Eye-head Coordination Pattern

All children's records showed large variations in the relationship between the eye and head within the same amplitude. Consequently no statistical conclusion has so far been drawn in terms of determining specific abnormalities in the eye-head coordination patterns. However, it was noted that four children presented a hypometric record of head movement to acquire a target, and three children showed a hypermetric record. Both cases appeared to suggest some difficulty in oculomotor control in relation to the vestibular-neck afferent feedback system.[6] When the head was free to move, the records appeared to be also disturbed by the child's poor postural background adjustment (subtle indication of poor equilibrium reaction). It is a common feature of the CP child to compensate for his inadequate integration of neck righting and equilibrium reactions by fixating at the neck and the trunk. Due to a lack of extensor tone or asymmetrical distribution of muscle tone, a

number of children tended to flex the head and trunk to the side, as they were preoccupied by the task and the cognitive control was inhibited.[7] Therefore, an additional limitation is that the head helmet was not able to detect asymmetrical head position in the record.

The pilot study has suggested a need for further investigation. First of all, the wide range of records indicates that a very large sample is necessary in order to arrive at a reasonable statistical measure. Also, sufficient normal subjects are required to make a reliable comparison of the records. Secondly, reduction of behavioural interference of the data must be considered. Thirdly, random eye and head movements were noted that appear to be indicative of some abnormalities in OM control mechanism of these children. Thus, it requires a careful review of the children's pattern of eye movement recordings by a specialist.

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MULTICHANNEL ELECTRICAL STIMULATION WITH THE REGARD TO THERAPEUTIC OR ORTHOTIC USE

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ABSTRACT

Therapeutic effects due to multichannel electrical stimulation of paretic lower extremities during gait have been observed considering its use after clinical therapy. Step length and velocity have been compared in stimulated group of 8 and control group of 4 from intended 10+10 patients. Measurements have been performed at free gait without stimulation before, at the end, and 8 to 11 months after the therapy. The choice of parameters and results of still continuing study have not given statistically valid answer. Measurements of ground reaction forces combined with goniograms and EMG recordings are more explicit. Further use of simpler orthotic stimulators is indicated in several cases.

INTRODUCTION

Since the invention of multichannel electrical stimulation of lower extremities as a rehabilitation method for paretic patients (1), it has been proved that application of surface stimulation to all the main muscle groups can normalize paretic gait significantly (2). During the simulation, typical gait anomalies estimated by kinesiological analysis are lessened, step length and step time are improved, while symmetry of hemiplegic gait is corrected as well. In the early phase of rehabilitation, multichannel surface stimulation helps establishing antigravity support and initial gait patterns. With the extended multichannel therapy, the recovery rate appears to be faster, correction level higher, and fatiguing of patients lesser than with other therapies.

In studies of post–stimulatory improvement observed since the begining of functional electrical stimulation, results of single channel peroneal stimulation have been discussed on short term (3,4) and long term basis where significant increase of muscle force (5), improvement of motor coordination, gait pattern and neurological condition (6) have been noticed.

With the intensive multichannel therapy and a complex information input through the afferent pathways to higher nervous centres (7), considerably larger therapeutic effects are expected after the rehabilitation. Establishing that, only simple orthotic devices or even no stimulation would be required after the therapy. In the opposite case, the development of multichannel stimulators for permanent use is strongly indicated. The purpose of our work is to investigate quantitatively and qualitatively the therapeutic effects of multichannel stimulation as it is now to initiate new directions of research.

A control study between stimulated and non–stimulated group of patients is continuing measuring main gait parameters and performing clinical tests. Measurements of additional parameters have been included to explicit primary investigation.

METHODOLOGY

For the control study, patients have been divided into a stimulated and a control group. Disabilities, age, and initial gait pattern have been considered to select both groups. At the selection of two similar groups, difficulties concerning localisation and extent of the insult as well as restrictions from ethical reasons have been encountered; namely, all the patients had to be given the maximal therapy. The control group has been so selected from patients who refused the stimulation or with whom it was contraindicated. Similarity has been achieved in: status after cerebrovascular insult, adequate motivation, period from the insult to the begining of therapy (about 5 month), accompanying pathologies (hypertension, myocardiopathy, etc.), period of therapy. Each group should consist of 10 patients. Investigation on 7 hemiplegic and 1 paraparetic patient due to lesion (barotrauma) on the thoracic level (Th 7) in the stimulated group and 4 hemiplegics in the control group have been completely concluded so far. There have been 5 left and 6 right side hemiplegic cases.

The patients in the control group have been involved in regular therapy program excluding any kind of electrical stimulation. The patients in the stimulated group have been treated by multichannel stimulation and partly also included in the regular program, so both groups have been treated daily for the same amount of time. Patients in the stimulated group have been walking with multichannel surface stimulator 5 times a week, each session lasting from 10 minutes at the begining to more than 1 hour at the conclusion of the therapy in some cases. For hemiplegic patients, adequate 6 channel stimulation has been applied to pretibial muscles and plantar flexors of ankle, knee extensors, knee flexors, hip extensors, and hip abductors during swing and stance phases according to the observed disabilities and established methodology (2). For the paraparetic patient with the main motoric deficit on the right side, new stimulation sequences have been determined and 8 channel stimulation applied to left and right hip flexors, right hip abductors, knee flexors, knee extensors, ankle plantar flexors (m. soleus), left and right pretibial muscles. The therapy of all the patients lasted from approximately 2 to 3 months.

The rehabilitation has been accompanied with measurements and tests throughout the therapy, starting before its begining. For both groups, all the measurements and tests have been accomplished without stimulation and at gree gait speed. Only so, the changes can be compared in both groups and patients can be observed from cybernetic aspects as dynamic systems enabled by the therapy (7,8). Control measurements and tests have been carried out 8 to 11 months after the conclusion of the therapy as well.

For each patient, test of motor functions and clinical kinesiological gait analysis rating anomalies with marks from 1 to 3 several parts of double support, swing phase, stance phase, and symmetry in all leg joints (9) have been fulfilled. Average step length and average gait velocity have been measured for basic comparison, both parameters being important biomechanically for

the final intent of locomotion. Ground reaction forces and their spatial distribution on the foot during gait have been measured together with average loading on crutch where patient and measuring system permitted (10). Goniometric measurements of joint angles have been added in the course of investigation as well as surface EMG recordings over the 6 main muscle groups during gait to get an insight into the neuromuscular system and to objectify measurements of external biomechanical parameters.

INSTRUMENTATION

Two clinical and research oriented 6–channel digital stimulators (11) have been used for the stimulation therapy. Channels of both stimulators can be programed by arrays of switches whose position graphically represent the stimulation sequences. Separate channels are triggered by heel–switch in the shoe of the impaired leg. Optionally, two switches under both heels can be applied with the later developed stimulator triggering 3 channels each. This has proved useful with the paraparetic patient where besides, electrodes on m. gluteus max. and those on m. glutei med. + min. have been connected together on one channel and a small orthotic stimulator attached under the patients knee (12) has been applied to n. peroneus comm. All the stimulators adapt themselves fully to the walking rate of the patient (8).

Ground reaction forces and their distribution on the foot during stance phase have been measured on line with a system of shoes with 8 strain gage transducers each, amplifiers, and programs implemented on miniprocessor. The system is connected together by cables (13). Similar system with the force transducer in crutch has been used for on line measurements of crutch loading. A goniometric system (14) combined with the force–shoes above instead of regular foot–switches and refined with lately developed software has processed joint angles in the same time. EMG recordings have been registered by silver cutaneous electrodes and a standard polyelectromyograph.

RESULTS

Average step length $\bar{L}$ and average gait velocity $\bar{V}$ as functions of time in months are given for 8 patients in the stimulated group (Fig. 1 and 3) and 4 patients in the control group (Fig. 2 and 4). Each point on the diagrams contains 3 courses of 14 to 30 steps each where step length and step time have been measured. The intended number 10 + 10 for both groups have not been achieved yet due to the large extent of the therapy and accompanying measure-

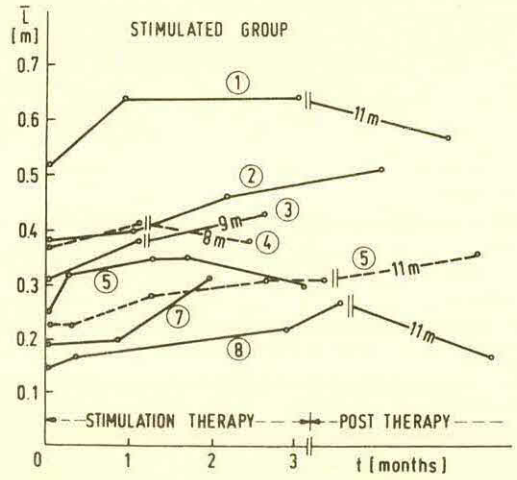


Fig. 1

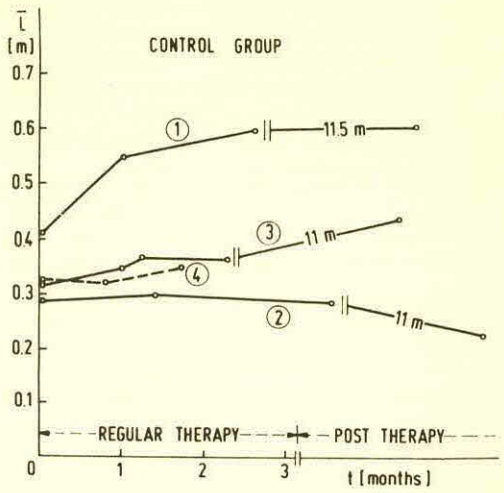
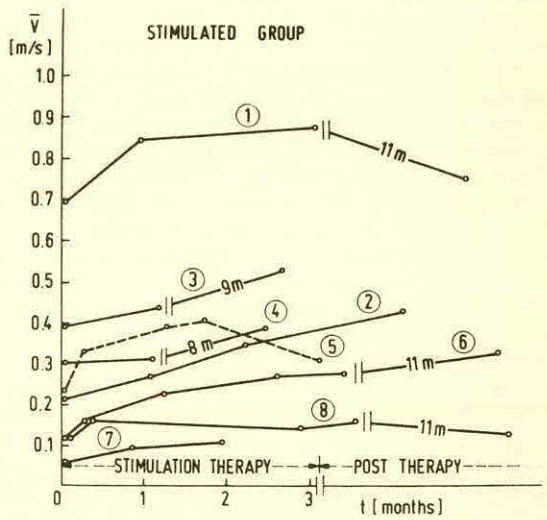


Fig. 2



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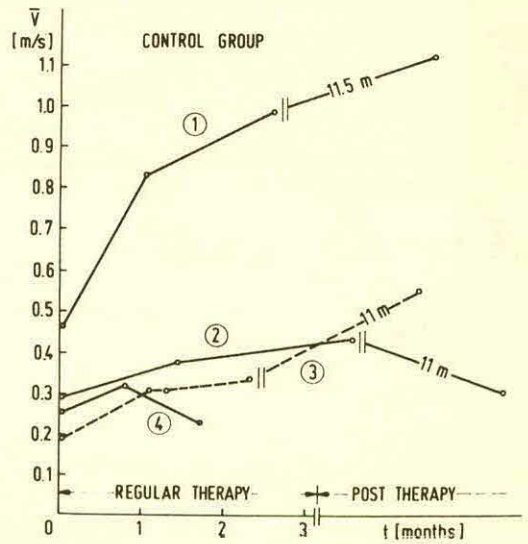

Fig. 4

ments. Time axes are interrupted after concluded therapy. Continuations of curves to the points of control measurements 8 to 11 months later steepened by shorter axes can not be compared as the effects of therapy. They are more picture of psycho–somatic statuses of patients, their home environments. motivations, etc.

The majority of curves approaches an expected exponential improvement of both parameters with the shape:

$\bar{L} = \bar{L}_1 + (\bar{L}_0 - \bar{L}_1)(1 - e^{-kt})$ and $\bar{V} = \bar{V}_1 + (\bar{V}_0 - \bar{V}_1)(1 - e^{-kt})$,

where index 1 represents initial value before treatment, index 0 limiting value close to that before the lesion (7,8), and k covers the rate of therapy, psycho–physical condition of the patient, his motivation, etc. Average curves for both groups have not been computed and even when the intended 10 + 10 patients would be accomplished, they could not be interpreted only statistically due to large differences among patients hidden mostly in the constant k in the exponent.

Tests of motor functions, clinical gait analysis, and other data about patients show that ethical moment has been presented at the selection of both groups. The stimulated group has included patients with age under 60 and extensive lesions attended by disturbances of sight (hemianopsy), partly aphasy, sensoric lacks, and contractures. The majority of 5 could not move alone and with them gait has been initiated by the stimulation. Initial measurements of $\bar{L}$ and $\bar{V}$ have been achieved with the support of accompanying person(s). The control group has consisted of patients with typical capsular lesions. Their age has been over 60 and they have been able to move partly by themselves before the therapy. Analysis shows that better motoric improvements have been achieved in the stimulated group, whose patients have been all able to walk independently before the conclusions of treatment. Control examinations after several months have showed that stimulated patients recovered further in part observing motoric functions and gait anomalies. While at the same time, active correction of gait has remained equal or become worse in the control group, where some applications of assistive devices have been required. These control observations can also not be regarded as the effects of therapy only.

Results of ground reaction forces measurements and their distribution on feet together with crutch loadings on one patient from the stimulated group are given before treatment (Fig. 5) and after 2.5 months of therapy (Fig. 6). In the upper force versus time diagrams, higher reaction forces under left and right foot (full lines) and lesser crutch loading (dotted line) after the therapy can be observed for both feet. Support on left leg of left hemiplegic patient has been improved (for 50N) as well as step times and symmetry of gait. Lower diagrams show the shift of reaction force center to the left during stance phases and lack of left heel landing

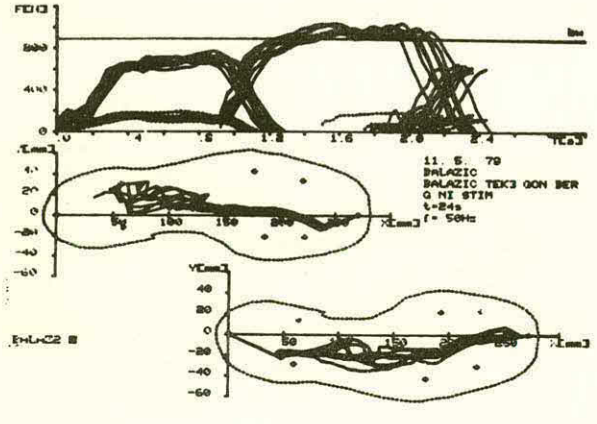

Fig. 5

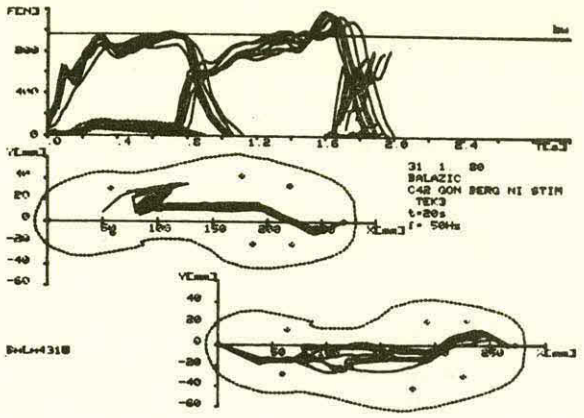

Fig. 6

in the initial stance which has not been improved during treatment. Dispersion can be observed for those parameters as well.

EMG recordings before the therapy (Fig. 7) and after 3.5 months (Fig. 8) are presented for paraparetic patient with motoric lack on right side predominantly. Activity detected during free gait over right m. tibialis ant., m. gastrocnemius cap. med., m. quadriceps, m. gluteus max. – all stimulated during therapy – and non–stimulated m. errector trunci, has been strengthened. Cyclical activity of individual muscle groups, adequate reciprocal innervation, and better muscular coordination can be observed after stimulation therapy. In this case, stimulation has been applied 2 years after lesion and improvements can be regarded as pure therapeutic effects.

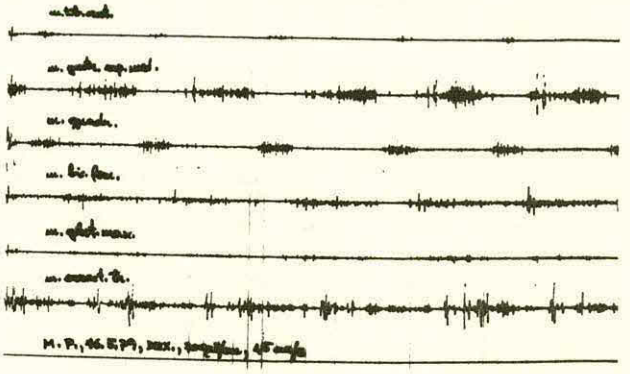

Fig. 7

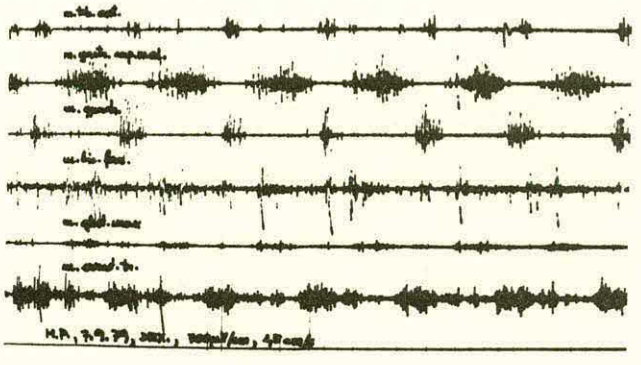

Fig. 8

CONCLUSION

Experience with immediate effects of multichannel stimulation (2) partly arises ethical obstacles at the selection of similar groups for the control study of therapeutic effects. Their presence implied in the secondary biomechanical parameters $\bar{L}$ and $\bar{V}$ seems shaded even further. Introduction of endurance measurements, which would explicate the state after therapy, is questionable from medical aspect. The social status of patient, the impact of his living environment, his motivation, and challenge for further improvement influence further to the achieved recovery level. Statistically significant evaluation of the therapeutic effects is hardly to be given so far. Observation of closer biomechanical parameters together with the EMG activity are more promissing. The latter can show the changes before their appearance in biomechanics of gait.

The multichannel stimulation can reestablish organisation of motor activity after the therapy. Localisation of its influences as well as their extent remain still to be given. Larger effects can be expected at proximal muscle groups. Application of simple orthotic stimulators after the therapy is indicated in several cases. Development of two– and more–channel orthotic devices should also esteem correlation of antagonistic muscle groups on both legs during gait to achieve more by less.

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FUNCTIONAL PERONEAL NERVE STIMULATION AND ITS IMMEDIATE EFFECTS ON MEASURES OF GAIT PERFORMANCE

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ABSTRACT

Long term use of functional electrical stimulation (FES) during human locomotion has been reported to improve measures of gait including velocity, cadence, and stride length. This study seeks to determine whether changes in measures of gait also occur after immediate application of FES. Twelve patients with upper motoneuron lesions were studied. After briefly determining the clinical response to peroneal nerve stimulation, the investigators had each patient walk with and without FES under command conditions of "natural" and "fast as possible" walking speeds. Results of a two factor factorial analysis utilizing the parameters of velocity, cadence, step and stride lengths did not reveal any significant changes attributable to FES at either command speed. Since no short term changes after immediate FES application were found in this study, the authors infer that previous reports of gait performance changes after long term FES usage may have resulted from long term facilitation of gait patterns in the central nervous system.

INTRODUCTION

Functional electrical stimulation (FES) of the peroneal nerve has been advocated in the literature as an orthotic treatment for patients with upper motoneuron syndromes with clearance difficulties[1]. Direct motor stimulation of the peroneal nerve causes dorsiflexion at the ankle during swing phase and improves or eliminates dragging of the foot on the floor and provides the patient with a heel strike at initial contact. The literature also suggests, however, that peroneal FES results in the improvement of certain performance measures of gait both qualitatively and quantitatively. A 1973 study by the Committee on Prosthetic Research and Development[2] reports that the Warsaw School of Medicine found an improvement in hemiplegic gait "both in elegance and speed" when peroneal nerve stimulation was utilized. Alfieri, Grillo, and Merletti[3] reported an increase in natural cadence and in gait as a whole after only one to two months of usage of functional electrical stimulation to the peroneal nerve. Waters, McNeal and Perry[4] measured gait velocity, stride length and cadence at both comfortable and fast walking speeds in thirteen patients and found an increase in these gait parameters after six months of stimulation. The studies mentioned above seem to support the idea that chronic usage of FES results in an improved gait performance beyond the clinically obvious direct motor response of the peroneal nerve which caused dorsiflexion of the foot. However, the literature does not specify whether the mere application of FES is sufficient to cause an immediate change in gait performance measures or whether long term usage of FES as an orthotic device is necessary in order to produce a training effect which brings about improvement in these performance measures. The objective of the current study on peroneal FES is to determine whether changes in measures of gait performance occur immediately after the application of FES or whether long term usage is necessary to develop such changes.

METHOD

Twelve patients with upper motoneuron lesions who had difficulty controlling swing phase of gait were studied. Eight patients had had a cerebrovascular accident, two patients had multiple sclerosis, one patient had cerebral palsy and one patient had spinal cord disease of unknown etiology. All patients were neurologically stable for at least six months or more. The main clinical reason for patient selection in this study was inadequate clearance with drag of the foot along the floor during some part of swing phase. (General criteria for selection includes the presence of upper motoneuron pathology with intact peroneal nerve function, ability to ambulate but with impaired swing phase clearance, passive ankle range of motion to approximately neutral, ability to perceive, respond, and adequately tolerate stimulation, and absence of a cardiac pacemaker.) When a suitable patient was identified clinically, the REC 2 Functional Electrical Surface Stimulator (see appendix) was applied to locate the best site of peroneal nerve stimulation producing balanced dorsiflexion. Patients were ambulated a short distance to evaluate and adjust the response during gait. After checking the integrity of the response as well as patient tolerance to stimulation, patients were brought to a walkway to study the immediate effects of peroneal nerve stimulation applied during swing phase. Each patient was instructed to walk a distance of thirty feet in sets of two command speeds: (1) "walk naturally,"

and (2) "walk as fast as possible." Peroneal nerve stimulation was applied to half the group during the first set of commands and was not applied during a second set of commands. The other half underwent the reverse procedure (i.e. no stimulation during the first set of two command speeds followed by stimulation during the second set). This procedure was adopted to eliminate effects of test order. Measures of maximum velocity, cadence, step length, and stride length were made with and without the application of functional electrical stimulation at both sets of command speeds. Cord connected to a tachometer was attached to the patient's belt by means of a safety pin in order to measure velocity. The analog velocity signal was recorded on a calibrated strip chart recorder and twenty to twenty-five measures of the maximal velocity in each gait cycle were averaged over the distance walked. Dyed moleskin was applied to the heel and sole of the foot of each shoe in order to stain a sheet of rolled brown paper on which measurements of step length and stride length were made. An average of at least five to seven measurements in each run was used. Two precision stop watches were used to determine the time of walking thirty feet and deriving cadence. Patients were studied under the above protocol immediately after the initial application of functional peroneal nerve stimulation and were to be studied after six months of daily FES usage. At the time of this writing, six month follow-up has not as yet been completed but these results are expected to be available at the time of the conference. Data on the immediate effects of FES were handled by means of analysis of variance consisting of a two factor factorial design.

RESULTS

Analysis of the data reveals that for the group as a whole, patients were able to distinguish the two sets of commands, namely, "walk naturally" versus "walk as fast as possible" (Table I). With regard to the parameter of maximal velocity, no significant difference was observed with FES compared to the runs without FES. Table II illustrates the findings for cadence. Patients were able to walk at different average cadences ($p < .1$) when given the different commands of "walk naturally" versus "walk as fast as possible." There was no difference, however, between the conditions with FES and without FES as far as cadence is concerned. Table III illustrates that the right step length varied as a function of the differing command speeds. However, left step length as well as left and right stride length were not influenced by functional electrical stimulation nor were they significantly different at different command speeds. When the stimulation unit was turned on, all patients were noted clinically to have visibly improved dorsiflexion of the ankle and clearance during swing phase as well as heel strike at initial contact of the stance phase. All patients who had pathological inversion of the foot without FES were converted to a neutral or even partially everted foot position when FES was turned on. Thus, from a clinical point of view, the orthotic goal of gaining ankle-foot control during swing phase was achieved.

DISCUSSION

Even though stimulation of the peroneal nerve produces primarily a direct motor response of the dorsiflexor musculature, many patients and many investigators have reported that gait performance is improved beyond the effects immediately attributable to the direct motor response[5,6]. In our experience, many patients subjectively claim it is "easier to swing the leg" and that their endurance improves with consistent daily usage. Improved gait performance measures have been reported after chronic usage of peroneal nerve stimulation during gait after six months[4]. In this study, however, no change in maximal velocity, step length or stride length were noted after immediate application of FES was made. (It should be noted that a good clinical response was obtained in every case, and even though some patients were unaccustomed and apprehensive about the initial application of the unit, the majority of the patients had satisfactory tolerance.) Our negative findings in the acute situation, therefore, suggest that if there is a change in gait performance measures following long term FES usage, the change may result from a long term training effect. It is hard to ascribe such a change in gait performance measures solely to direct motor nerve stimulation. Direct peroneal nerve stimulation provides the patient with immediate dorsiflexion at the first application of the device. Since no acute change has been observed in gait performance measures, we suspect that chronic usage may alter the central nervous system's response in such a way as to facilitate the gait performance measures that have been described in the literature as improved. It is not likely that changes in focal muscle strength which can be produced by direct chronic neuromuscular stimulation will result in changes in gait velocity, step length, cadence, and stride length. Rather, it would appear to be more reasonable to assume that facilitation of central nervous system activity may be taking place as a result of long term usage. From a rehabilitation point of view, further exploration of this idea is warranted.

SUMMARY

Functional electrical stimulation to the peroneal nerve was applied in a group of twelve patients with upper motoneuron lesions. Measures of gait performance including maximal velocity, step length, stride length, and cadence were obtained after the immediate application of the stimulation unit. No changes in these gait performance measures were noted at either of two command speeds. Negative acute findings in the face of previous literature reports of improvement following long term usage of FES suggest that FES may be inducing a long term facilitation of gait performance in the central nervous system.

TABLE I: MAXIMUM VELOCITY

Source	SS	df	ms	F	p
Total	4.54	43	--	--	--
Command Speed (walk naturally or walk as fast as possible)	.67	1	.67	6.98	<.025
With FES vs. without FES ($\bar{c}$ FES vs. $\bar{s}$ FES)	.003	1	.003	.03	NS*
Command Speed vs. $\bar{c}$ FES vs. $\bar{s}$ FES	.017	1	.017	.18	NS*
Error	3.85	40	.096	--	--

TABLE II: CADENCE

Source	SS	df	ms	F	p
Total	17602	47	--	--	--
Command Speed	683	1	683	1.80	<.1
$\bar{c}$ FES vs. $\bar{s}$ FES	39	1	39	.10	NS*
Command Speed vs. $\bar{c}$ FES vs. $\bar{s}$ FES	136	1	136	136	NS*
Error	16744	44	380	--	--

TABLE III: RIGHT STEP LENGTH

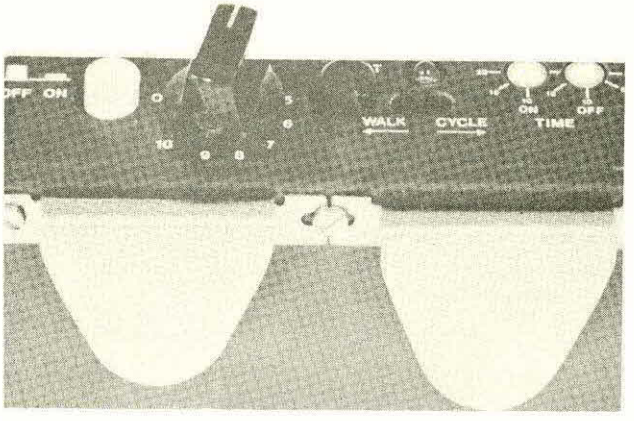
Source	SS	df	ms	F	p
Total	9008	47	--	--	--
Command Speed	391	1	391	2.01	<.1
$\bar{c}$ FES vs. $\bar{s}$ FES	.04	1	.04	.002	NS*
Command Speed vs. $\bar{c}$ FES vs. $\bar{s}$ FES	58.96	1	58.96	.303	NS*
Error	8558	44	194.5	--	--

NS* = not significant

SS=Sum of Squares
df=Degrees of Freedom
ms=Mean Squares
F = F-ratio
p = Probability

APPENDIX

The peroneal nerve stimulation unit used in this study is the REC 2 Functional Electrical Stimulator. The unit is designed to apply a train of stimulating pulses to the surface of the skin along the course of a motor nerve in order to produce muscle contraction. In a walking mode, stimulation of the peroneal nerve is controlled by an insole footswitch located under the heel. When the patient lifts his heel during terminal stance, the train of stimulating pulses to the peroneal nerve is activated. The train is terminated by weightbearing on the heel during the next ground contact. The unit operates by means of a 9 v. transistor battery and with a load of 560 ohms, has a maximum output voltage of 40 volts along with a pulse duration of about 80 microseconds. Pulse frequency can be varied between 30 and 200 pps. Controls available to the patient on the top of the unit are an on-off switch, a ten position intensity selector switch, a walk-cycle operation selector (cycle mode enables the patient to exercise a group of muscles in a therapeutic but ambulatory mode), a battery test button and indicator light, and adjustments for cycles on and off time intervals. The unit clips onto the patient's belt and has a long cable connected to the insole footswitch located in the shoe. It also has separate cables to two conductive rubber electrodes which are held in place on the skin over the peroneal nerve by velcro straps.



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AN INVESTIGATION OF THE CARRY-OVER OR THERAPEUTIC EFFECTS OF FES IN THE CORRECTION OF DROP FOOT IN THE CEREBRAL PALSY CHILD

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ABSTRACT

The "carry-over" or therapeutic effects of FES of the peroneal nerve given in syncrony with the patient's gait is being studied in the correction of drop foot in cerebral palsy children. For six subjects who have been studied thus far, the dorsiflexion produced while the FES was in use, was lost immediately when the stimulator was turned off, and no carry-over could be demonstrated.

INTRODUCTION

Since its inception in 1961 by Liberson et al[1] experimenters who have applied electrical stimulation of the peroneal nerve in synchrony with the gait cycle for the correction of drop foot, have repeatedly observed that for many of their patients, the dorsiflexion produced during the period of the FES continues for some time after the stimulator has been turned off (for examples and for additional references see Waters et al. 1975[2]). This "carry-over" effect is reported to persist for several hours or even throughout the following day, after which it wanes and must be repeatedly reinstated by periodic application of the FES. We have sought to induce this carry-over effect in Cerebral Palsy children having drop foot and to document its properties using quantitative gait analysis techniques[3].

METHOD

Three types of data are recorded simultaneously as follows: While the subject walked, we measured foot-to-floor contact patterns, electromyographic activity from surface electrodes over the ankle flexors and extensors, and rotation of the foot in the sagittal plane at the ankle joint. Foot-to-floor contact patterns are obtained via a tri-electrode foot-switch system consisting of three separate strips of aluminum tape affixed to the medial, lateral and heel portions of the patients shoes. Finally, video taping of each subject's gait assists in the interpretation of these abstracted parameters of walking function.

Experimental Paradigm

Each subject's gait performance is evaluated on three separate occassions spaced one week apart before any FES is applied, in order to assess baseline performance. This is important when working with CP subjects since the degree of spasticity observable from day to day (and hence the patient's walking ability) is somewhat dynamic.

Two additional gait analysis sessions are then given over the next two weeks. During each of these sessions, the patient uses the FES for 20 minutes. Data are taken immediately preceding, during and immediately following the 20 minute period of use of the FES. Following the second of these sessions the patient is given the stimulator to take home and is instructed to use it for 40 to 60 minutes daily. The patient returns for identical gait analysis sessions after two weeks of "home use" of the FES and then at roughly monthly intervals thereafter. This paradigm allows both "short" and "long" term effects of the FES to be assessed.

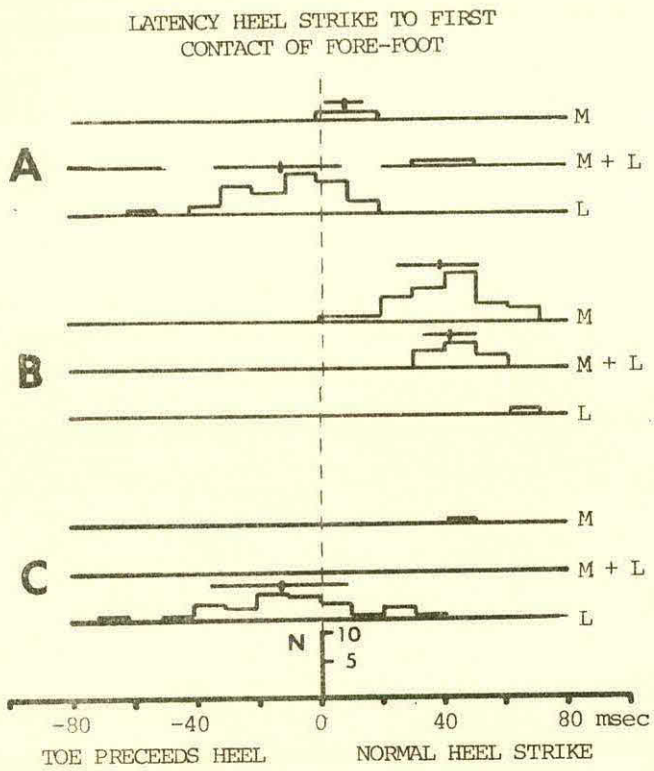
RESULTS

Results thus far for three children (ages 5-8 years) and one young adult (age 16 years) who have been tested after daily use of the FES, have been disappointing, in that no evidence of "carry-over" as defined above has yet been observed; this despite two subjects having received FES for nearly one year. Similar findings apply for an additional two children who because of their more recent entry into the program have been examined for effects of short term usage only.

In all cases an immediate and dramatic improvement in the patient's gait toward a more normal pattern is always seen when the stimulator is in use, but a reversion to the previous abnormal gait pattern always follows as soon as

the stimulator is turned off. Figures A and B demonstrate both the improvement in gait function that occurs with the FES and the absence of any carry-over of this improvement once the stimulator is turned off.

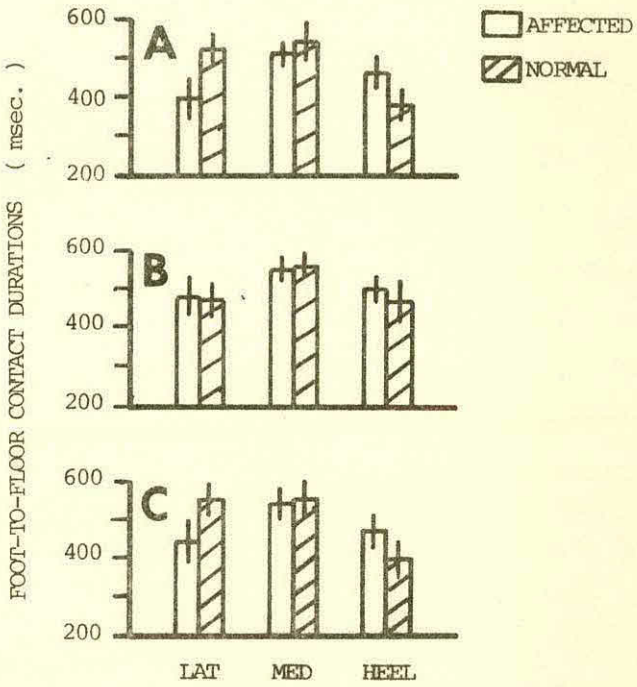
Figure 1



The histograms in Fig. 1 show for the stimulated leg the distribution of latencies measured from heel strike to the first contact of the forefoot. Data appearing to the right of the origin represent the normal foot-contact sequence of heel strike followed by forefoot contact. Steps in which "toe walking" occurred produce negative latencies since the forefoot contact preceeds the heel contact and these events are plotted to the left of the origin. With this graphical representation, more normal gait performance is indicated by a displacement of the data further to the right. The delay time between heel contact and forefoot contact is a direct function both of the elevation of the forefoot as heel strike is approached, and of the ability of the ankle dorsiflexors to resist the torque placed on the ankle at heel strike, which acts to produce plantar flexion. A further refinement of the display permits registration of which part of the forefoot comes into weightbearing contact first (i.e. lateral 'L' vs medial 'M' vs lateral and medial stimultaneously 'M+L')

Thus in 'A' which shows the subject's gait performance before using FES that day the majority of the steps showed "toe walking" with lateral forefoot contact occurring first and with the heel contact following later. The mean latency for this event was -11.8 msec as indicated by the short vertical tick mark (bin width=10 msec; horizontal bar indicates $\pm$ one std. dev. from the mean). The correction of the patient's gait when the FES is in use is apparent in 'B': All of the steps showed the normal contact sequence of heel strike followed by forefoot contact. The fact that the first part of the forefoot to contact the walkway was the medial side is a consequence of the slightly exaggerated eversion produced with FES of the peroneal nerve. 'C' shows the gait performance with the stimulator turned off and immediately follows the 20 minute period of FES shown in 'B'. The absence of carry-over is apparent as the patient's gait returned to the abnormal state seen in 'A'.

Figure 2



In Fig. 2 the mean foot-to-floor contact durations for the lateral and medial forefoot and the heel are plotted for the normal foot (open bars) and for the affected foot (cross-hatched bars) of a hemiplegic child. As in Fig. 1 above, the sequences 'A', 'B' and 'C' represent gait function before, during and after FES. The most apparent feature of the subject's gait abnormality is the desparity in the duration of lateral contact of the affected vs. unaffected foot. Use of FES in 'B' greatly improved the symmetry of function between the two feet. In the case of the lateral contact duration the affected foot showed some increase while the unaffected foot showed a decrease. When the stimulator was turned off at 'C' the abnormal, asymmetric gait was resumed.

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THE INFLUENCE OF ELECTRICAL STIMULATION ON MUSCLE STRENGTH AND FATIGUE IN PARAPLEGIA

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ABSTRACT

Isometric muscle moments induced by electrical stimulation were measured in ankle and knee joint of nineteen paraplegic patients. Knee extensors, ankle plantar and dorsal flexors were stimulated. The patients with upper motor neuron lesion were chosen, otherwise they were randomly selected regarding etymology, lesion level, time past injury, sex and age. Two paraplegic patients have passed the exercising program with chronic electrical stimulation. The muscle force has increased significantly in both patients, while the fatiguing of the stimulated muscle has been reduced. The comparison of the results obtained in training program with the results of the other patients shows that almost all atrophied muscles, which were measured, can be brought to nearly normal status. Such muscles can be efficiently used in paraplegic patient's standing, standing up, and even walking.

INTRODUCTION

Several reports describing applications of functional electrical stimulation (FES) to paraplegic patients have been published in the last decade. Wilemon et al. (1970) (1) have implanted the stimulators to T-5 patient with the aim to obtain locking of both knees. The patient was then able to stand and even walk with the extended legs. Specially interesting and encouraging are the results of knee extensors exercising with FES. Similar atrophied muscles strengthening has been reinvestigated by Kralj et al. (1973) (2,3). The paraplegic patient's muscles which were strengthened in a period of two to three months were strong enough to ensure the raising from the sitting to the standing position and few minutes of erect standing. A strong fatigue of electrically stimulated muscles was observed because of relatively high stimulation frequency (50 Hz). Mortimer et al. (1973) (4) and Peckham et al. (1976) (5) have demonstrated that muscle force and fatigue resistance increase with electrically induced daily exercise. The electrical stimulation alters the contractile properties of the muscle in the way that fast (glycolytic) muscle fibers are converted into slow - fatigue resistant (oxidative) fibers. The electrically reinforced muscles can therefore also after longer periods of no functional use remain in good condition. An important achievement regarding the technology of multichannel implantation and electrical splinting of the knee in paraplegia has been reported by Brindley et al. (1978-79) (6). Two patients with T-7 and T-12 complete lesions received five and six channel implanted stimulator respectively. Knee extensors were stimulated with T-12 and knee and hip extensors with T-7 patient. Both patients after being electrically exercised showed a significant increase of the force obtained in m. quadriceps muscles and good increase of fatigue resistance. They were able to stand and walk by means of crutches and electrical splinting of the knees. The T-7 patient was able to stand for 35 min and walk for 5 min, while the T-12 patient was standing for 75 min and walk for 5 min as well. Swing-to and swing-through walking modes were applied in these experiments. In 1979 we have reported (7,8) about the possibility of primitive biped walking pattern consisting of double and single stance phase. During double stance phase both m.quadriceps are stimulated, while during the single stance phase only one knee extensor is contracted. The swinging of the other leg is provided through excitation of flexor response in hip, knee and ankle joint. This gait was performed with two patients (T-5 and T-10) in parallel bars.

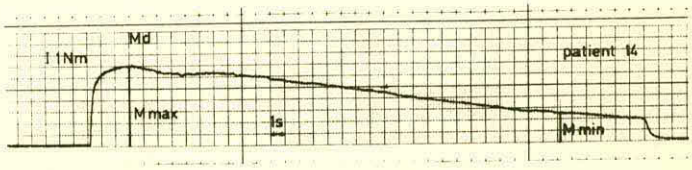

Fig. 1

Summarizing this brief review it is evident that the functional use of electrically retrained centrally deinervated atrophied muscles of paraplegic patient is possible. The questions of optimal procedure of muscle exercising, prediction of results, selection of patients who can benefit from FES remain open. An attempt to solve at least some of the mentioned problems is presented in this report.

SELECTION OF THE PATIENTS

The patients were examined during their summer vacations when almost all members of Society of paraplegics patients are brought together. All the patients asked were willing to participate in the measurements. Some general data of 19 patients are gathered in Table 1. Car accidents, different falls and tumors were the most frequent reasons of paraplegia. None of them was stimulated before this experiment.

Table 1

No.	Lesion	Year of birth	Year of injury	f (%)
1	T-10	1952	1970	17.1
2	T-9	1941	1969	27.3
3	T-3,7	1948	1968	79.1
4	T-5,6	1949	1964	50
5	T-12	1946	1962	59.5
6	T-8	1946	1968	41.7
7	T-4,5,12	1953	1977	12.1
8	T-8	1950	1974	71.9
9	T-11,12	1935	1978	1.8
10	T-5,7	1943	1960	-
11	T-8	1947	1977	2.1
12	T-12	1939	1960	14.3
13	T-12	1931	1966	5.9
14	T-4	1913	1956	97.8
15	T-12	1933	1973	10.7
16	T-4	1960	1975	0
17	C-7	1946	1971	2.8
18	T-5	1945	1966	1.8
19	T-6	1945	1962	7.7

METHODS

Surface stimulation electrodes fixed to the extremity with velcro straps were used in the measurement. A battery powered voltage output stimulator was providing stimulation pulses of 0.3 ms duration and 20 Hz repetition rate. The knee and ankle torque isometric measuring braces were transforming the joint torque into the voltage with the help of strain-gauge transducers. The torque response was recorded at different stimulation amplitudes from the threshold to the saturation level. Each stimulation amplitude was applied to the muscle for 5 s. The

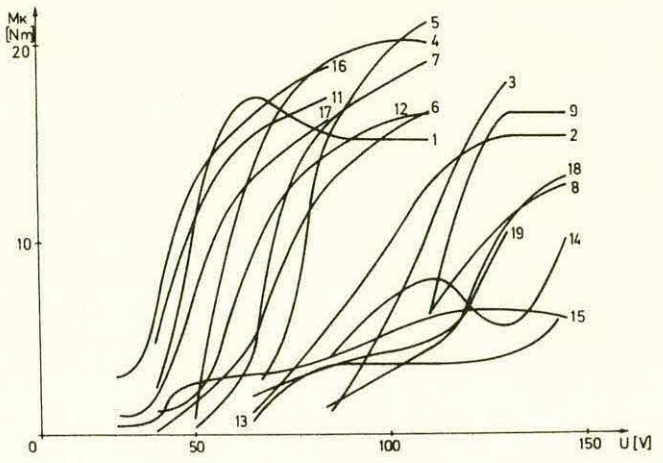

Fig. 2

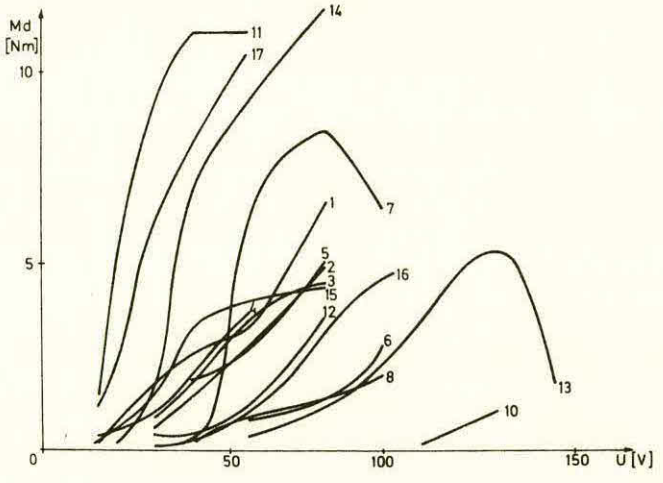

Fig. 3

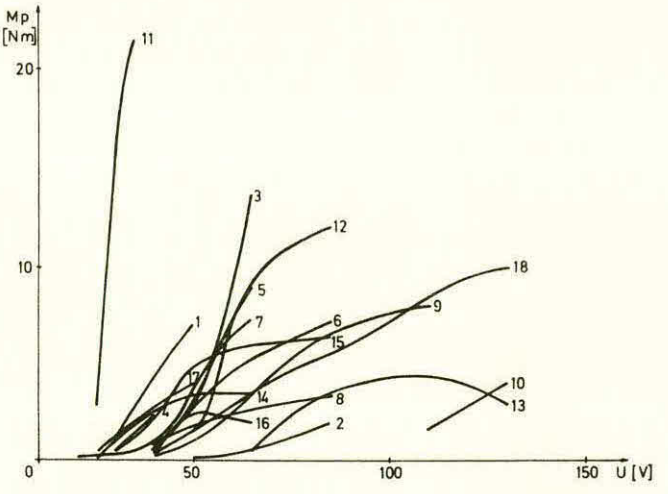

Fig. 4

pause of 20 s elapsed before increasing the stimulation voltage. When recording the muscle fatigue the stimulation sequence was prolonged up to 1 min. A typical record of knee moment is presented in Fig. 1. Mmin is the value of the joint moment occuring 30 s after the maximal joint torque Mmax. The measure of the fatigue f was defined as a torque drop in 30 s normalized with Mmax and expressed in %.

RESULTS

The measured characteristics of joint torque versus stimulation amplitude representing the gains of electrically stimulated muscles are presented in Fig. 2 to 4. Fig. 2 belongs to knee extensors, while Fig. 3 and Fig. 4 correspond to ankle joint dorsal and plantar flexors respectively. Each curve is marked with a number denoting particular patient from the Table 1. The fatigue coefficients f for knee extensors are also presented in the Table 1.

Two paraplegic patients who are not included in Table 1 have passed the muscle exercising program with chronic FES. In a month and a half of daily stimulation program the muscle force increased significantly. The muscle gain characteristics for a T-10 patient are presented in Fig.5 to 7 for knee extensors, dorsal and plantar flexors respectively. In the first week of the program the FES session lasted for half an hour while additional half an hour was added each week. Both extremities were trained isotonically with trains of pulses (4s of stimulation and 4s of pause). The increase of fatigue resistance was noticed as well. The coefficients f describing the fatigue rate of knee extensors before and after exercising are presented in Table 2.

TABLE 2

	31.1.79	15.3.79
right leg f (%)	16.4	3.5
left leg f (%)	30.8	16.7

In Fig. 8 the results of electrical exercising of the second patient (T-5) are given. The ankle joint plantar flexors were stimulated for three months. Five months after the training program the muscle characteristic was measured again. From the Fig. 8 it can be noticed that muscle force did not much decrease what supports the reports of Peckham (5) and Brindley (6).

CONCLUSIONS

From the measurements of the gain of electrically stimulated muscles in nineteen paraplegic patients with spinal cord lesions mostly in thoracic

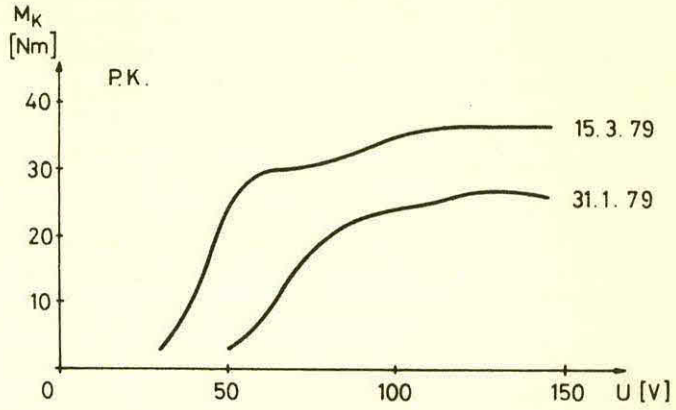

Fig. 5

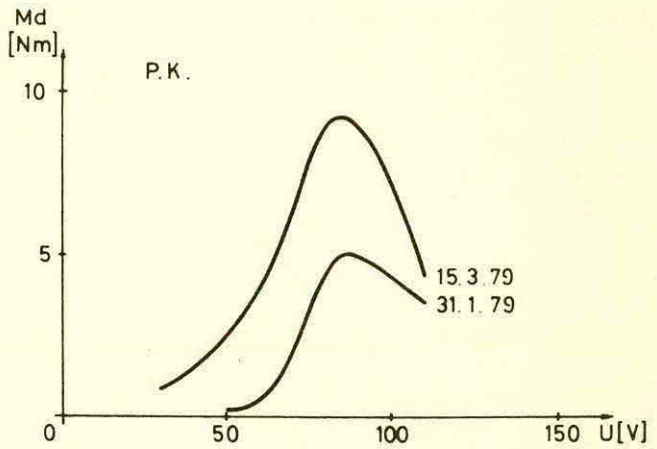

Fig. 6

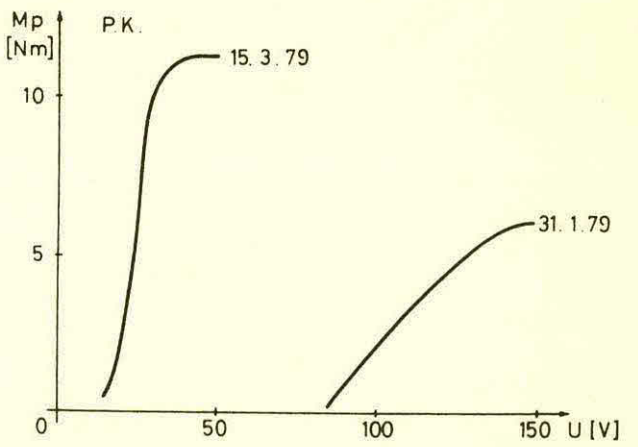

Fig. 7

part of the spine it can be concluded that the muscles are not only electrically excitable but in some cases even enough strong to perform functional tasks such as standing, standing up, and primitive walking. The results of exercising with chronic FES are showing that the muscles of the rest of the patients can be strengthen to the functional level in relatively short period of 2 to 3 months. The same is true also for the fatigue resistance of paraplegic patient's muscles. The results obtained are encouraging and the authors believe that further research for finding better exercising procedures is to be conducted.

ACKNOWLEDGEMENT

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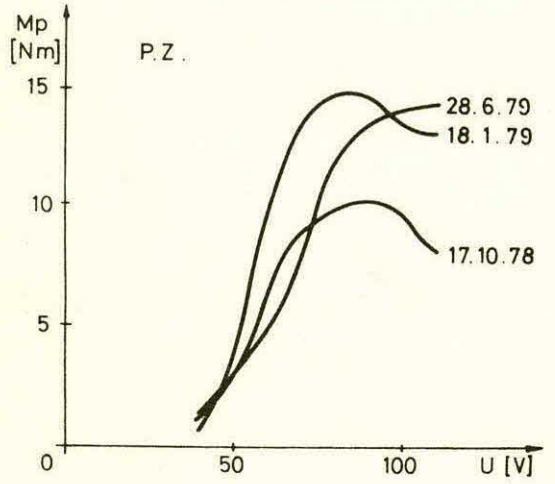

Fig. 8

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RESTORATION OF KEY GRIP AND RELEASE IN THE C5 AND C6 TETRAPLEGIC THROUGH FUNCTIONAL ELECTRICAL STIMULATION

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ABSTRACT

Electrical stimulation of selected paralyzed forearm and hand muscles in C5 and C6 spinal cord injury patients provides control of key grip and release.

Chronically indwelling percutaneous coiled wire electrodes were used to stimulate the muscles.

The patient controls the timing and the strength of the contraction from a single control signal. Three types of command controller are available; a shoulder position controller, a head position controller, or a myoelectric signal.

Five subjects are presently involved in the evaluation of this system. The function the electrical stimulation provides has been beneficial in performing tasks such as eating and writing.

INTRODUCTION

Injury to the spinal cord at the fifth and sixth cervical level results in severe motor paralysis of the hand. Electrical activation of paralyzed muscles is a means of providing controlled grasp and release in these subjects. In this study, electrical stimulation was employed to provide active contraction of the finger flexor and thumb adductor and extensor muscles.

METHODS

Restoration of motor function requires that there be 1.) a suitable means to excite the muscle and 2.) a means by which the patient can govern the strength of the contraction. We have developed a system for control of lateral pinch and release in the hand. In this system, a volitional command governs the stimulus applied to each of the muscles. Excitation of the muscles results in finger and thumb opposition/adduction (for lateral pinch) and in thumb extension (for release).

Electrodes

Percutaneous intramuscular coiled wire electrodes are used to apply the stimulus to the muscle. The electrode is a modification of that originally described by Caldwell (1).

Electrodes were implanted into the Flexor digitorum superficialis and/or profundus for finger flexion and into the Opponens pollicus and/or Adductor pollicus for thumb motion. For release, electrodes were implanted in the Extensor pollicus longus and Extensor digitorum muscles. A total of four active electrotes were used. Electrodes were implanted with hypodermic needles and the position of the electrode is identified by stimulating through the needle as the electrode is inserted. The electrodes remain chronically indwelling for extended periods (frequently months). The interface site is covered with a small surface connector which is taped to the skin, providing for protection of the electrodes and ease of electrical connection.

Modulation of Contraction Strength

The contractile strength of each electrically stimulated muscle is controlled both by recruitment and by temporal summation (2). To modulate force, the patient varies the single proportional input by the command control movement. This control input determines the output stimulus pulse width and frequency applied to each of the four electrodes.

Control of the contractile strength over the full range was achieved as follows. Minimal force was developed by activating each electrode with a just suprathreshold stimulus at the lowest allowable stimulus frequency. Demand for a stronger force could be continued to be met by pulse width modulation until all fibers were activated. Further force demands were met by frequency modulation with the pulse widths held fixed. Thus, we achieved an effective means for controlling force throughout its entire range while maintaining the lowest possible stimulus frequency at all times, consistent with fatigue-resistant performance.

Coordination of Muscle Action

We choose to use a single command signal for activation of several muscles. This required that these muscle groups be activated in a coordinated manner. The functional requirements was that the thumb contact the lateral aspect of the index finger. Coordination was provided by applying the command to the finger flexor electrodes prior to and at a faster rate than was that to the thumb adductor/opponens electrode. The result was that the fingers assumed a flexed position prior to thumb movement, so that the thumb closed consistantly against the fingers.

External Orthotics

Our objective was to minimize the need for external orthotics, thus freeing the patient of problems in donning. For C6 subjects, no external orthosis is used. This allows the subject full normal use of his hand when stimulation is not applied. Since these subjects are able to voluntarily extend their wrist, they achieve some grasp passively by tenodesis grasp. Electrical stimulation provides grasp over a full range of wrist positions.

For C5 subjects, external support of the wrist is required. Presently, this is supplied with a conventional dorsal wrist support. No additional bracing the thumb or fingers is required for the subject to achieve functional lateral prehension. Figure I shows the grasp and release with electrical stimulation.

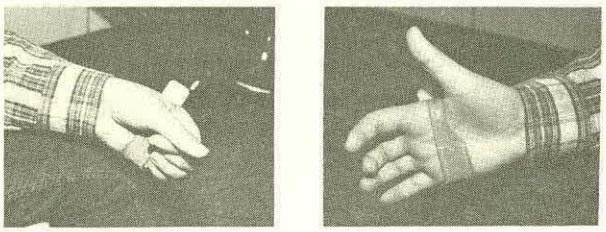

FIGURE I. Lateral Pinch and Release Achieved through Functional Electrical Stimulation in Paralyzed Hand of C5 Spinal Cord Injury Subject

Patient Control of Functional System

Three alternative command sources are available. They are shoulder position, head position, and myoelectric activity. A shoulder position transducer measures the relative position between the shoulder and sternum (3). The head position transducer measures the deviation of the head relative to vertical (3). The myoelectric control uses activity from a muscle such as sternocleidomastoid, which is processed and used to provide a command which is proportional in time (ie. exceeding a threshold level changes the command at a predetermined rate) (4). In the case that either position controller is used, control logic provides automatic "rezeroing" which allows the subject to respecify the initial position, and a "hold" function, which results in a fixed stimulus output regardless of command input.(5)

Patient Selection

Five adult male patients, all with spinal cord injury due to traumatic injury of the cervical vertebrae resulting in paralysis below C5 or C6 were involved in this study. The patients have been using functional systems from 1 week to 24 months. Details are given in Table I

SUBJECT	AGE	FUNCTIONAL LEVEL	ACCIDENT	TIME SINCE INJURY	OCCUPATION	TIME IN PROGRAM
TVHH	29	C6	Fell From Ladder	2 yr.	Student	2 yr
JHJ	30	C6	Swimming Accident	2 yr. 5 mo.	None (at home)	1 yr. 9 mo.
CHG	25	C6	Auto Accident	1 yr. 8 mo.	Student	11 mo.
MHH	21	C5	Diving Accident	1 yr. 6 mo.	Student	8 mo.
WHM	30	C5	Auto Accident	6 mo.	In-patient	3 mo.

TABLE I: SUBJECTS IN F.E.S. PROGRAM

Hardware Development

Special purpose instrumentation has been developed to be utilized by outpatient subjects for evaluation of the effectiveness of this technique. Figure 2 shows a subject with the stimulator unit on his lap. Briefly, the device will accept control commands from the three sources specified above, process the command, and generate the appropriate output stimuli. The device is 6 x 3 x 2 inches, powered by mercury cells, and has no external controls. Plugging in lead wires from the electrode and control sites turns on the circuitry. The stimulator is designed to be generally applicable to both the C5 and C6 subject. For each individual, appropriate controller gains, stimulus parameters, etc. are determined in testing, using a laboratory stimulator which functions identically to the patient devices. When the parameters are specified, they are then entered into the patient stimulator. This approach allows us to eliminate potentiometers from the device.

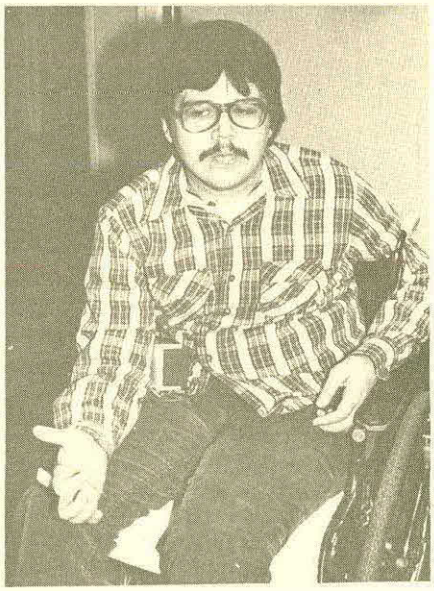

FIGURE 2. Subject in F.E.S. Program. EMG electrodes are on neck muscle (Sternocleidomastoid), and stimulator at waist. All lead wires are beneath clothing.

RESULTS

Each of the patients involved in the development of the muscle stimulation system demonstrated the ability to perform movements of lateral pinch and release. The tasks of lateral pinch and release the subjects performed with the muscle stimulation were performed without any additional orthotic aid. Such aids were necessary if the stimulation system were not active. These tasks generally were of a tonic nature, such as holding writing and eating utensils. In such activities, the utensil was generally held between the index and long fingers together. With C6 subjects, muscle stimulation allowed this grasp to be maintained over a full range of wrist position, thus allowing some versatility in the hand position when the tasks were performed. These results demonstrate that it is feasible to develop functional key grip and release in the C6 tetraplegic patient through electrical excitation of paralyzed muscles.

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MAXIMAL ELECTRICAL STIMULATION FOR URINARY DYSFUNCTIONS

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ABSTRACT

Short-term maximal electrical stimulation (MES) was used for treatment of urinary incontinence and retention in adults and children. Using self-stimulation technique with the stimulator designed for that purpose the patients were able to obtain a rather high intensity of stimulation without significant discomfort. 30 percent of the patients were improved or relieved of their symptoms. From the presented data and from data revealed in our previous studies, MES can be suggested to be accepted in the routine methods of treatment.

INTRODUCTION

Short-term maximal electrical stimulation (MES), applied once or several times, proved to be a useful method in treating urinary incontinence in adults (1). Several questions remained unanswered in our previous studies. How to reduce the discomfort of the patient during MES? Could MES produce therapeutic effect in enuretic children and in spinal cord injured patients having urinary retention? Is it possible to make a reliable selection of the patient for MES, i.e. obtain correct prognosis of the therapeutic effect of MES?

The objectives of the study were: a) to design the electrical stimulator for MES which would be suitable for the adjustment of stimulation intensity by the patient himself; thus the level of the discomfort due to stimulation is expected to be controlled by the patient's will, b) to test the efficiency of the stimulator on female patients with urinary incontinence, enuretic children and selected spinal cord injured patients having urinary retention, c) to test the reliability of the method of selection for MES based on the urodynamic measurements during the first application of MES, d) to quantify the post-stimulation changes in the bladder and urethral function by means of urodynamic measurements.

METHOD

One-channel unit generating constant current pulses was designed and used for MES (Fig. 1). The patient (adults) was asked to adjust the intensity of stimulation up to the level of tolerable discomfort. In children the intensity of stimulation was adjusted by the physician. Anal or vaginal plug electrodes were used for stimulation. Monophasic square current pulses of 1ms duration, a pulse rate of 20 cps were used, the current ranging from 15 to 100mA in adults and from 7 to 25 mA in children. The duration of one stimulation session was from 15 to 20 minutes in adults and approximately 10 minutes in children. Urodynamic measurements, i.e. urethral pressure profile (UPP) and cystometry (CMG) were performed before MES, during the first application of MES, and at the end of observation of the patients. The period of observation lasted from 6 to 24 months in adults and from 1 to 4 months in children, in the course of which MES was repeated 2 to 4 times.

FEMALE URINARY INCONTINENCE

143 female patients with the symptoms of stress incontinence, urgency and/or urge incontinence and mixed stress and urge incontinence were treated with MES. Using self-stimulation technique the patients were able to obtain a rather high intensity of stimulation without significant discomfort. 30 percent of the patients were improved or releaved of their symptoms. 68 patients underwent complete urodynamic tests. Among them there were all of the successful cases (43 patients) and 25 failures.

Almost all combinations of the bladder and the urethral responses to the first application of MES (as was previously observed,(2)) and the treatment with MES were found in clinically successful cases as well as failures. Only in the cases of motor urge incontinence the elimination of the uninhibited bladder contract-

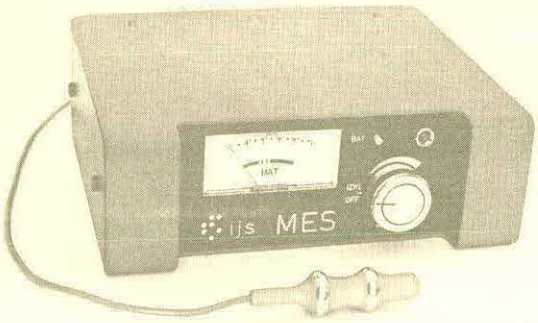


Fig. 1 Therapeutic stimulator MES with vaginal plug electrodes

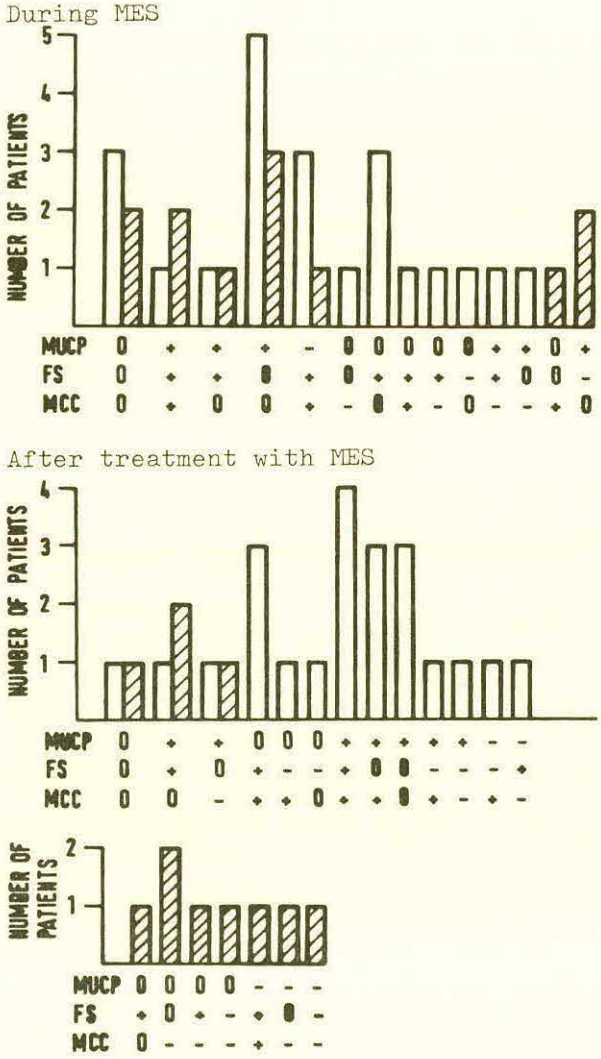
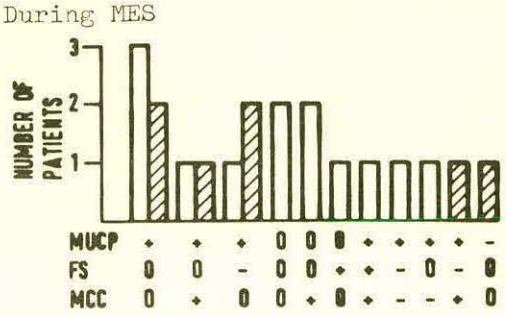


Fig. 2 Changes in CMG and UPP during and after treatment with MES in patients with stress incontinence - 22 successful cases (□) and 12 failures (▨).



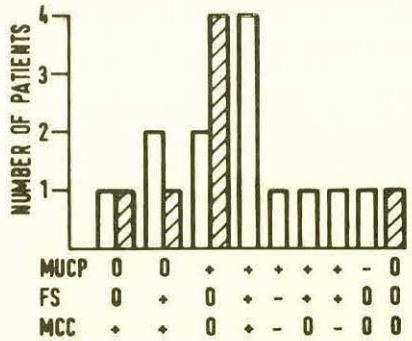
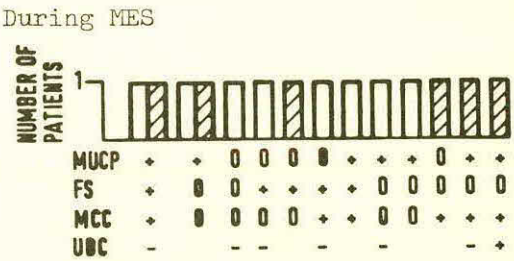


Fig. 3 Changes in CMG and UPP during and after treatment with MES in patients with mixed stress and urge incontinence - 13 successful cases (□) and 7 failures (▨).



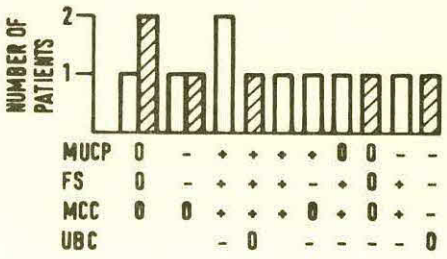


Fig. 4 Changes in CMG and UPP during and after treatment with MES in patients with urge incontinence - 8 successful cases (□) and 6 failures (▨).

LEGENDE to figures 2, 3, 4: MUCP-maximal urethral closure pressure, FS-bladder capacity at the first sensation, MCC-maximal cystometric capacity, UBC-uninhibited bladder contractions, n-number of cases, +, increase; -, decrease; O, unchanged.

ions during the first application of MES and after MES was correlated to the success of the treatment with MES (Figs.2,3,4). Patient's self-stimulation seems to be the optimal way of applying MES. 30 percent of the success could be expected from the results of this study.

The selection of patients for MES, i.e. prognosis of the effect of treatment, on the basis of measurement of UPP and CMG during the first application of MES proved to be of no value, except in the patients with motor urge incontinence. Thus, MES can be applied without urodynamic examination, since it never caused aggravation of the clinical state of the patient.

The changes in the measured urodynamic parameters in the clinically successful cases and failures do not seem to be a reliable estimates of the primary ethiological cause of incontinence, except possibly in the cases of motor urge incontinence and stress incontinence due to the weakness of the urethral muscles.

Whether other more sophisticated urodynamic methods and neurophysiological measurements would be a more proper tool for a reliable selection of patients for MES and for evaluation of the therapeutic effects of MES, is a question to be answered in the further studies.

INCONTINENCE IN CHILDREN

Fourteen children, in the age from 6 to 14 years, with the symptoms of diurnal and/or nocturnal enuresis and incontinence due to myelomeningocele were treated with MES using the anal plug stimulating electrodes.

Among 9 cases with nocturnal and diurnal enuresis treatment with MES revealed improvement of incontinence in 7 cases and 2 failures. In 6 of the successful cases CMG and UPP were performed before and after treatment with MES. Uninhibited bladder contractions with a small or normal bladder capacity and with normal UPP values were found in all 6 successful cases. After MES in 2 cases uninhibited contractions diminished and bladder capacity increased. In 4 of the successful cases uninhibited contractions as well as the bladder capacity remained the same as before MES. In 2 failures, uninhibited contractions and small bladder capacity were observed before as well as after MES. UPPs were of normal values.

Among 2 cases with nocturnal enuresis MES revealed 1 improvement and 1 failure. In both cases CMG and UPP were of normal values.

In 1 case with diurnal enuresis the improvement was observed, CMG and UPP were normal.

Two patients with incontinence due to myelomeningocele were treated with MES. In one patient, in whom overflow incontinence was diagnosed, MES was uneffective. In the other, hyperreflex bladder was diagnosed (uninhibited contractions with low bladder capacity). In this case improvement was observed after MES, as well as diminution of the uninhibited contractions and increase in bladder capacity.

URINARY RETENTION

Four tetraplegic patients with the spinal cord lesion at the level of C3 to C5 were treated with MES. They were all having complete urinary retention, thus using the permanent catheter. Hyperreflex bladder with significant detrusor sphincter dyssinergia was diagnosed in all cases by means of urological, neurophysiological and urodynamic examinations. They were all having high urethral closure pressures, CMG revealed maximal bladder capacities from 200 to 300 ml. In 3 patients, the uninhibited bladder contractions were observed at lower bladder volumes.

After MES 3 patients were able to provoke reflex micturition and are not using the permanent catheter for the period of 5 months from the application of MES. During MES the urethral closure pressure remained the same or was increased. Functional lenght of the profile was increased. Uninhibited contractions were diminished. Maximal bladder capacity was increased or remained the same. Urodynamic measurements, performed one month after the application of MES, revealed decrease in the urethral closure pressure, diminution of the uninhibited bladder contractions and the same maximal bladder capa-

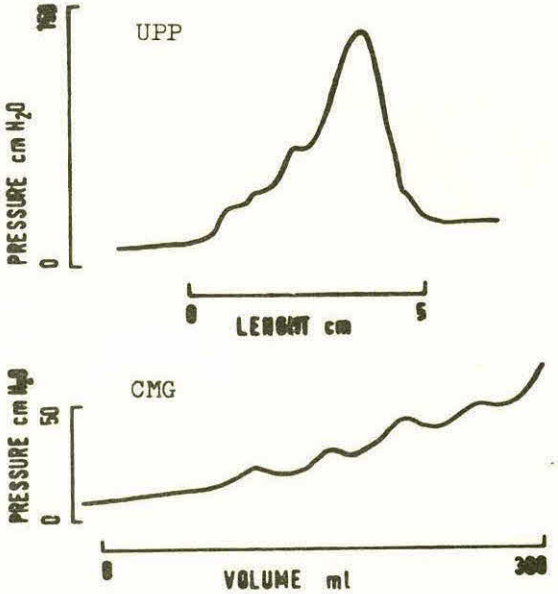


Fig. 5 Patient with urinary retention. UPP and CMG before MES.

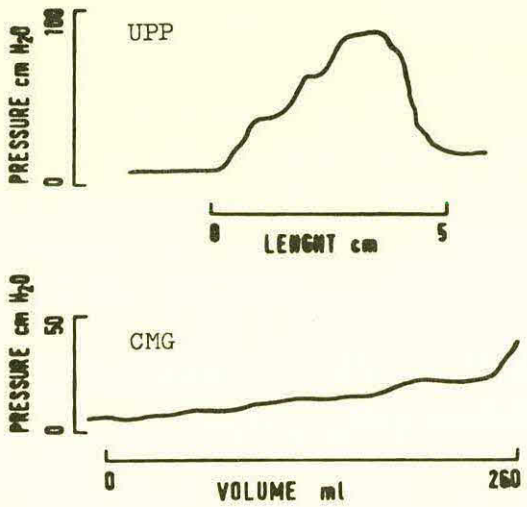


Fig. 6 Patient with urinary retention. UPP and CMG before MES.

city (Figs. 5, 6). In one patient MES was uneffective. UPP and CMG remained the same during MES as well as after MES.

DISCUSSION AND CONCLUSIONS

From the presented data and from data revealed in our previous studies (1), maximal electrical stimulation (MES) can be suggested to be accepted in the routine urological methods of treatment. Electrical stimulator MES can be used for treatment of different types of urinary incontinence, urgency and frequency of micturition. It can be used for treatment of urinary retention and faecal incontinence as well.

Indications: stress incontinence; urge incontinence due to neurological and unknown causes; enuresis in children after the age of six; reflex incontinence in patients with reflex neurogenic bladder and incontinence in other types of neurogenic bladder - in these patients MES is indicated only under the supervision of urologist; faecal incontinence in cases of showing no organic lesions; chronic retention of urine in the case of reflex neurogenic bladder.

Contraindications: different fistulas and ectopies; chronic retention of urine except in the case of reflex neurogenic bladder; before application of MES the urinary infection should be sanified, stone should be removed from the bladder etc.

Detailed anamnesis and clinical examination of the patient reveal the diagnosis in most cases. In evident cases of stress incontinence, urge incontinence and enuresis, MES can be used by the physician of general practice as a diagnostic as well as the therapeutic method. Examination of urine and urinoculture are necessary with the aim to sanify the urinary infection. In unclear cases detailed urological ginaecological, neurological and if possible urodynamic examinations should be done. The same is valid for the apparently evident cases in whom the treatment with MES was not effective.

Electrical stimulator MES is intended for treatment in clinics and ambulances and exceptionally for the use at the patient's home. Treatment must be supervised by the physician. The stimulation intensity is wished for to be adjusted by the patient himself up to the level of tolerable discomfort. Suggested duration of stimulation is up to 20 minutes.

First control is usually one month after the stimulation. Therapeutic effect is excellent in 10 to 30 percent of the patients. It is dependent upon the type of incontinence. In cca 50 percent of the patients different degrees of improvement of the clinical state can be expected. The duration of the therapeutic effect after the first stimulation varies from several days to several months. Therefore, the stimulation should be repeated in periods depending on the duration of the therapeutic effect. Therapeutic effect is observed mostly after the first or after the second application of MES.

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FIVE-YEAR EXPERIENCES IN TREATMENT OF FEMALE URINARY INCONTINENCE BY NONIMPLANTABLE FES

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ABSTRACT

Five-year experiences of treating female patients with urinary incontinence using functional electrical stimulation is presented. Most frequent kinds of incontinence in women as well as their pathogenetic mechanisms are described. The development of functional electrical stimulation used for treating urinary incontinence is illustrated. The effect of FES upon the pelvic floor muscles and detrusor of the bladder is described. The application of the acute maximum electrical stimulation (AMFES) as well as the urodynamic changes involved are described in detail. It has been found that the results of treatment using FES depend on the appropriate selection of female patients. A chronical treatment using FES is successful in 82 % of the female patients with stress incontinence. Urge incontinence treated with AMFES will show recovery in 91 % of the cases.

INTRODUCTION

Population Urinary incontinence is a frequent inconvenience in women. In approx. 5 % of all female population it is so serious that treatment is needed. Stress incontinence is the most frequent type of incontinence in female, urine leaks, when they are making any effort, when coughing or sneezing. The second most common type is urge incontinence when urine leakage takes place immediately after the first feeling for micturition. Less frequently we find incontinence of the neurogenic bladder which is observed after injuries or damage of the spinal cord or the central nervous (CNS) system. Each of the three types demands its own special treatment. But before choosing a type of treatment we have to perform all the neccessary examinations in order to determine the type and degree of incontinence. Before we decide which of the methods to use we must take a thorough patient´s history, perform the clinical examination and incontinence test as well as gynaecological, urological, urodynamical, neurophysiological and x-ray examinations.

Stress incontinence can be treated surgically, with exercises after Kegl and by functional electrical stimulation (FES). The results of the surgical treatment are not quite satisfactory, since a successful outcome can be found in only 80 to 90 % of the cases. Exercises after Kegl are successful only in cases of light stress incontinence. Urge incontinence is treated by bladder training, sedatives, spasmolitics and drugs acting upon the vegetative nervous system. The results of treating urge incontinence were even worse than those in stress incontinence (approx. 50 %).

As a result of a rather unsuccessful treatment of female patients with stress incontinence with the existing methods and because of contraindications in some patients, treatment with functional electrical stimulation has been introduced. A method for the treatment of patients with urge incontinence using FES has been developed as well.

PATHOGENETIC MECHANISMS IN TREATMENT WITH FES

The neurophysiological research showed that FES applied to the pelvic floor muscles acts mainly reflectively via the nervous sacral center of the spinal cord (Trontelj et al., 1974). That means that the reflex arch and the sacral center for micturiction in spinal cord must be preserved if FES is to be applied. From the urodynamic point of view it has been found that FES applied to the pelvic floor muscles performs contraction of the pelvic floor muscles and simultaneously the inhibition of the bladder detrusor. Whether or not FES will have an effect on both mechanisms or the contraction of the pelvic floor muscles depends on the current amplitude used. By changing the stimulation current amplitude the etiological mechanisms could be effected. Stress incontinence results from rather weakened pelvic floor muscles which will contract following use of a lower current amplitude. The pathogenetic mechanism in urge incontinence are spasms of bladder´s detrusor and these can be inhibited by applying a higher current amplitude of FES. By applying FES we can provide for an etiological treatment, or at least effect both pathogenetic mechanisms

causing both types of incontinence. Research have shown that reinnervation of the pelvic floor muscles following FES is quicker than usually which has a very favourable effect upon the treatment of urinary incontinence. The treatment of stress incontinence seems not only to result in strengthening of the pelvic floor muscles but also to provide a better reflex arch reacting quickly and adequately upon stress conditions.

TREATMENT OF URINARY INCONTINENCE WITH FES

Implantable electrical stimulators

Around 1960 already functional electrical stimulation was used for treating urinary incontinence. The implantable radiofrequency stimulators were used (Caldwel 1965).

Shelley reported in 1972, very critically from the physician´s and the patient´s standpoint about 380 implantations of radiofrequency stimulators being carried out for the treatment of urinary incontinence. The results reported and our experiences show the success in treatment just above 50 % which can hardly justify further implantations. Those unsuccessful cases are the cause that they are hardly in use more.

Nonimplantable electrical stimulators

Hopkinson and Lightwood described the intravaginal and intraanal possibility of stimulating the urinary tract´s closing muscles in 1967. The variation of a vaginal stimulator was presented by De Soldenhoff and McDonel in 1969, and the anal one by Glen in 1969.

Electrical stimulators for the treatment of urinary incontinence with a vaginal or an anal plug are nonimplantable and they consist of external unit generating the impulses and of the electrodes attached to the vaginal or the anal plug.

For the purpose of treatment of the urinary incontinence by way of the vaginal or the anal aperture, several stimulators have been worked out. The common scientific research done by physicians and engineers established the fact that exactly determined values of electrical current in duration and in the form are extremely important.

The following parameters were determined for optimal treatment of urinary incontinence: a rectangular impulse, frequency around 20 Hz, the current varying according to the effect desired. The current amplitude of around 30 -40 mA will only provide for the contraction of the pelvic floor muscles. The urethral closure pressure profile (UPP) increases. The current over 60 mA will cause contraction of the pelvic floor muscles and relaxation of the detrusor (Kralj, Šuhel, 1979). So we will use a stimulator generating current up to 40 mA for correcting stress incontinence. Its use is chronic. In the treatment of urge incontinence we will use a stimulator that will generate higher current, over 60 mA. Its use is short-lasting (Godec et al. 1975).

Automatic vaginal electrical stimulator

It was soon found that the therapy value of nonimplantable stimulators was the same if not even better than that of the implantable ones. The female patients especially are bothered by the external unit of the nonimplantable stimulators. This imperfection led the researchers to construct stimulator without external unit

An automatic vaginal electrical stimulator has been constructed. It is integrated in the plug, consisting of the electronic circuitry and the battery. This stimulator is called VAGICON-X.The stimulator starts operating as soon as it gets in contact with the tissue (Šuhel, 1976).

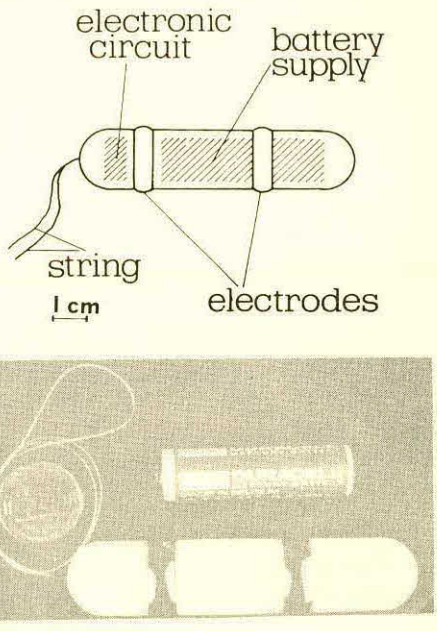

Fig. 1 The outward appearance of the housing of VAGICON-X.

Acute maximal electrical stimulation

For the treatment of the urge incontinence we use the acute maximal functional electrical stimulation (AMFES). Three pairs of electrodes are used, the electrodes in a form of the vaginal plug, the anal plug and two needle electrodes, that are pierced into the pelvic floor muscles, particularly into the m.levator ani. The voltage and the current on each electrode are raised to the level, where the pain can still be born by the patient.

A monophase and rectangular impulse is used also for the AMFES, the duration

1 msec and the frequency is 20 Hz. The average current used on all three electrodes is 90-120 mA, the stimulation lasts for 20 minutes. One single stimulation is sufficient for the treatment. In the recidive cases of the urge incontinence we stimulate once more after 1 to 3 months.

The characteristic parameters of the urethral closure pressure profile measured during and after AMFES were significantly higher than those measured before the application of AMFES.

Group results In our group of patients the bladder capacity from the point of the first feeling of urge to the point of imperative urgency increased during AMFES for 80 to 110 ml. This enlargement in capacity diminished slightly after termination of AMFES, but the bladder capacity still remained higher, although there was no further stimulation. This constituted the essential point in treatment of urge incontinence.

The increased intravesical pressure and the detrusor spasms disappeared during AMFES, as was expected from the inhibitory influence of FES on detrusor muscle.

The mechanism of urge incontinence improvement after AMFES is not clear. The possible alternatives are the facilitation of the supraspinal neuron or the direct action upon the lower motor neuron. After FES the voluntary control of the lower motor neuron improves.

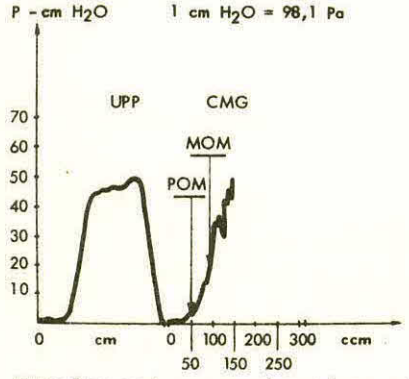

Fig. 2 Urodynamic examination of a patient with urge incontinence

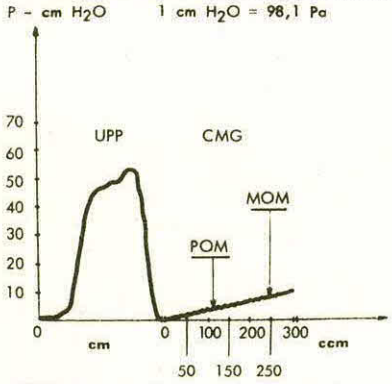

Fig. 3 Urodynamic examinations following application of AMFES.

RESULTS

Rather favourable results of the treatment of urinary incontinence using FES could be expected in only properly selected female patients. The choice is based upon the above mentioned analyses and a test application of FES to the pelvic floor muscles. In the urodynamic lab the evaluation of a test application is rather simple and reliable. From the urodynamic point of view the change of UPP when applying FES can be seen as well as a change of the cystometric curve.

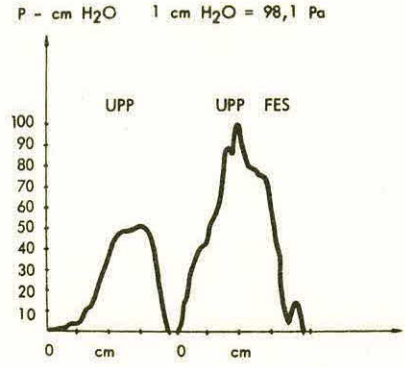

Fig. 4 Urethral closure Pressure Profile: left before and right after the application of FES.

Test application of FES Test application of FES indicates either a successful or unsuccessful treatment of stress incontinence with FES. In cases when treatment of stress incontinence in females requires either a vaginal or anal plug for stimulating the pelvic floor muscles, a test application of either type will still be needed. With respect to a more satisfactory response when a plug is applied a corresponding type of the stimulator will be chosen.

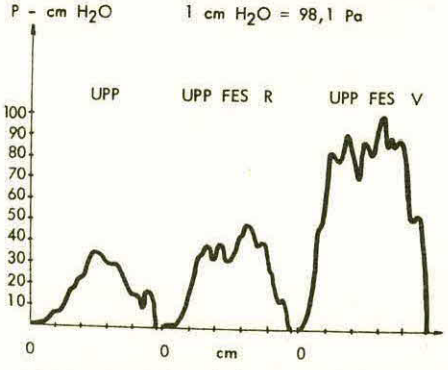

Fig. 5 Changes of UPP in patient with stress incontinence following application of FES: left - before stimulation, middle - during use of anal plug, right - during use of vaginal plug.

Results of the treatment of stress incontinence by FES

Recovered	40	59,7 %	82,1 %
Improved	15	22,4 %	
No effect	12	17,9 %	
Total	67	100 %	

Results of the treatment of urgent incontinence by AMFES

Recovered	35	62,5 %	91,1 %
Improved	16	28,6 %	
No effect	5	8,9 %	
Total	56	100 %	

CONCLUSIONS

Functional electrical stimulation is a successful method for treating urinary incontinence in females. A proper selection of patients is a prerequisite for a successful outcome of the treatment. The selection of patients requires a detailed history, clinical, gynecological and urological examinations as well as urodynamical and neurophysiological analyses. A test application of FES is of particular significance. With such a selection of female patients treatment of stress incontinence by means of a chronical FES can be successful in 82 % of the cases. The treatment of female patients with urge incontinence using AMFES can be successful in 91 % of the cases treated.

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THE USE OF ELECTRICAL STIMULI FOR REHABILITATION

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Electrical stimuli in the form of Bone Growth Stimulation (BGS) accelerates the healing time in the fractured tibia. It also helps approximately 60% of non-union fractures of tibia, some of which previously came to amputation. Bone Growth Stimulation (BGS) is a definite asset to early and complete rehabilitation.

Rehabilitation takes many forms, both mental and physical. This Paper presents a recent method called "Bone Growth Stimulation" (BGS) and its uses in dealing with rehabilitation.

Across Canada, 100,000 people are involved in accidents every year. These include home, industrial, car and motorcycle accidents which result in fractured limbs. Approximately 10,000 of these cases involve the long bones (tibia) in the legs.

Standard medical/surgical treatment of these fractures has been either by plaster casts or by insertion of pins or plates. Unfortunately, a number of these factures fail to heal, and are eventually classified as non-union joints in which no healing takes place. There were 311 such cases in the Province of Ontario in 1979. Several of these cases go on to amputation. These patients then have to go through a long and costly rehabilitation period. Para-bone growth stimulation (BGS) first gained prominence in the 1970's by Dwyer (1) followed by Bassett (2) and Patterson (3), all Orthopaedic Surgeons. It was discovered that by the use of electrical stimuli, new bone could be grown more rapidly. This resulted in the healing of approximately 60% of all non-union fractures. Each of these investigators uses his own special methods as listed below:-

(1) Non-invasive
(2) Semi-invasive
(3) Totally invasive

All of these methods helped in the cure of the patient with non-union joints and have enabled most of them to be rapidly rehabilitated and to lead a normal and useful life without limitations.

Today's success rate is now 90%. The increase in positive results lies in the fact that in most Centres, only

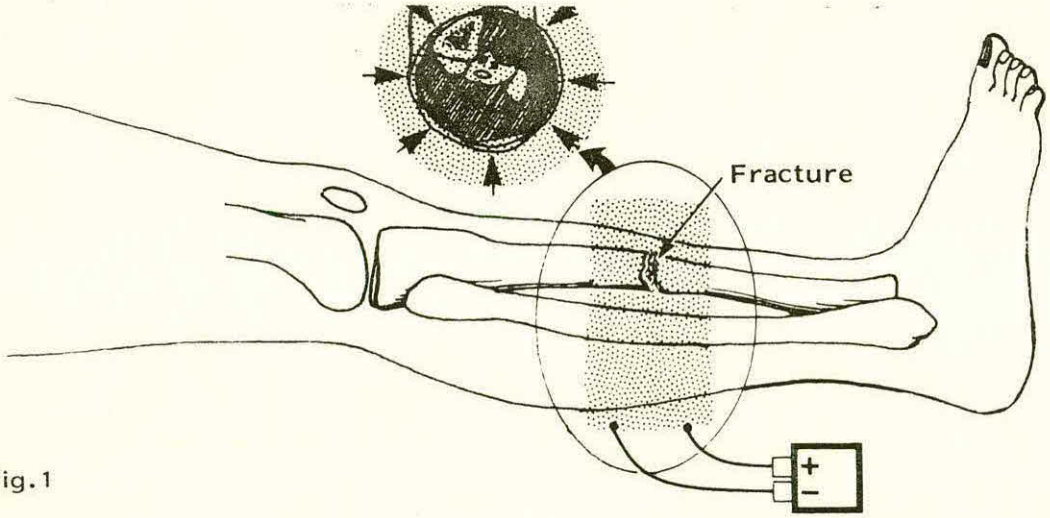

Fig.1

Fig. 1. The non-invasive technique involves placing coils around the skin in the region of the fracture and electromagnetic pulses are transmitted to the non-union site. This method has its limitations, as aiming the pulses to the precise site is difficult.

the true non-union fractures were originally treated by BGS. As experience was gained, it became apparent that other advantages were to be gained by BGS. For example, some cases where drugs did not appear to help an infection, BGS did. This has not been clearly documented. However, there is no doubt that the healing of fractures is accelerated and many Centres now use BGS in conjunction with standard methods. It has been found that by these means, the healing process has been shortened, and rehabilitation has thus been accelerated. The saving in funds to the Health Care Programme far outweighed the cost of the BGS Unit (approximately $400.00) even though the unit is non-reusable.

At Sunnybrook Medical Centre (4), we are using modified BGS in a study in cases of non-union of the carpal scaphoid that can result in a loss of mobility of the hand, resulting in adverse effects in patients who require full flexibility of the hands, as in typing, etc.

CONCLUSION

Present results with BGS in the healing of non-union fractures are very encouraging. Many patients have had a speedy rehabilitation who otherwise, would have faced limited activity or possible amputation.

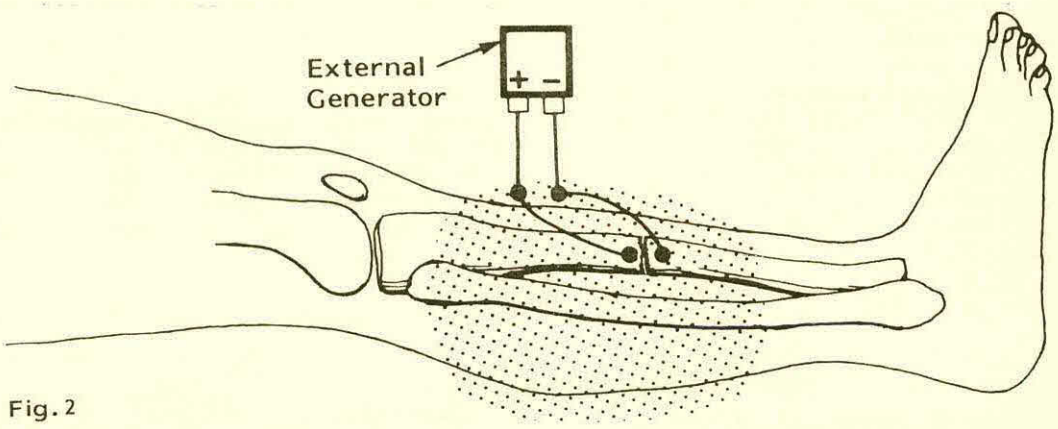

Fig.2

Fig. 2. The Semi-Invasive technique involves an incision and electrodes are attached to the bone immediately adjacent to the fracture. The electrodes are exposed and connected to an external generator which is attached to the outer limb and covered by a cast and bandage binding. The main disadvantage of this method is that the electrodes being external and present for a minimum of 3 months, increase the risk of infection.

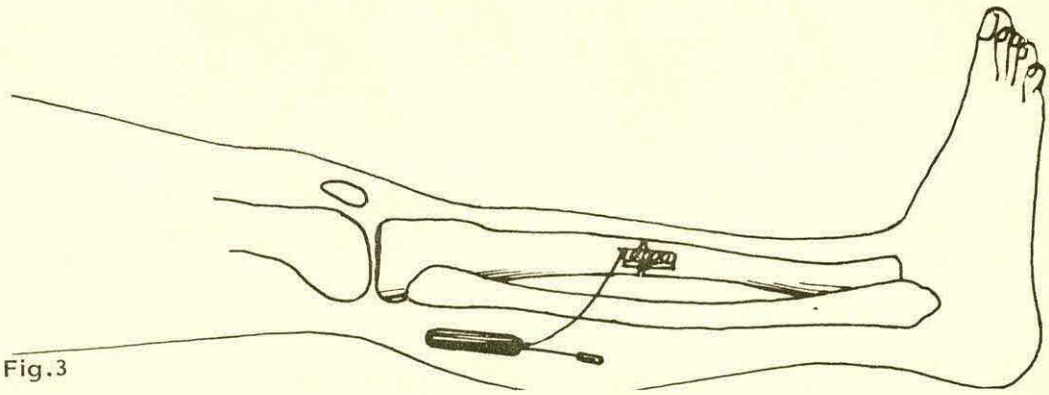

Fig.3

Fig. 3. The third technique, the Totally Invasive Method, combines the generator and electrodes as a single unit and is surgically implanted in toto, overcoming the risk factors of the two previously mentioned methods.

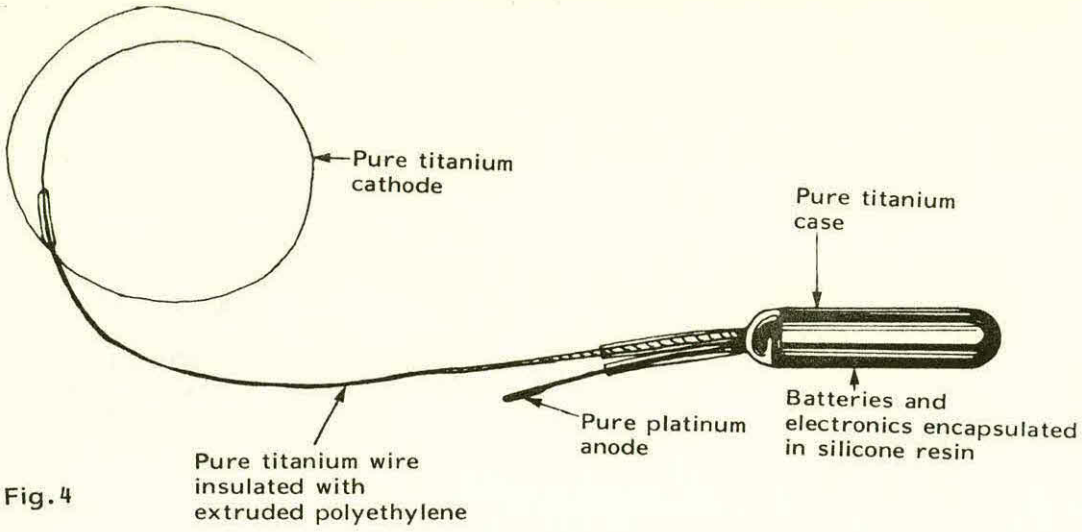

Fig.4

Fig. 4. This final method, developed by Sir Dennis Patterson (3) in Australia, has been used by our Orthopaedic Dept. at Sunnybrook Medical Centre, University of Toronto, for a Study on the "Effect of Electrical Stimuli on Non-Union Fractures". This unit which is a constant DC generator providing an output of 20 microamps over an impedence range of 1-100 Kohms. It is powered by two silicon oxide alkaline zinc cells. The battery life is calculated to last 9 months. The generator is encased in a silicon-resin capsule and sealed in an outer casing of pure titanium weighing 14 gms. and is 11 X 45 mms. in size. The electrodes are constructed of different materials. The anode is platinum and the cathode is titanium.

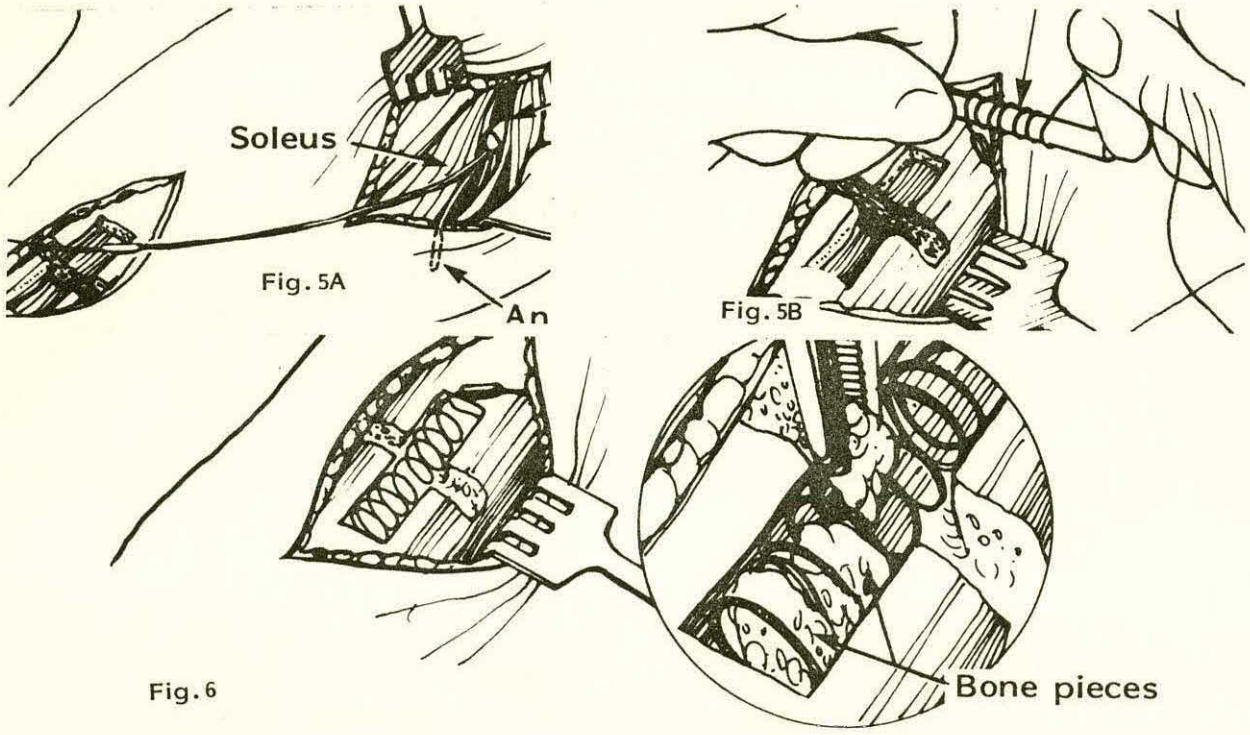

Fig.6

Fig. 5 & 6. At surgery, under general anaesthetic, an incision is made to expose the fracture site. The fracture site is cleaned to viable bone and a small slot prepared on either side of the fracture. The cathode is then formed into a helix coil and placed into a 1 X 1 X 3 cm. slot (1.5 cm.) of equal length on either side of the non-union. Cancellous grafting is recommended where large gaps are involved.

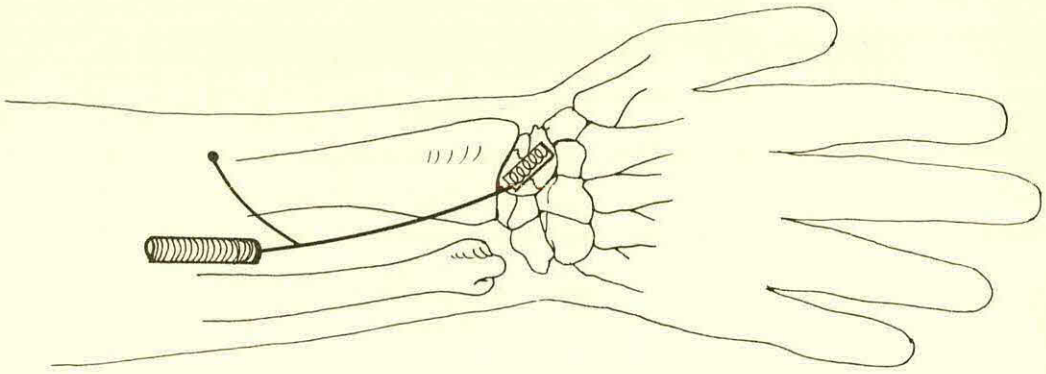

Fig. 7

Fig. 7. It was with the experience gained from long bone fractures that the problem of the fractured non-union scaphoid was investigated. After some preliminary tests, special units were designed. They were of a lower output, 5 microamps, and approximately half the original size, although all other specifications were the same. This is the first time that three documented cases of bone growth stimulation (BGS) in the scaphoid has been reported. The units were left in situ for 5 months. One of the three has proved successful.

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ELECTRONIC DEVICE CONTROL USING THE RETRO-REFLECTIVE CONCEPT

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ABSTRACT

The retro-reflective approach addresses the need to provide direct access control of electronically controlled devices for individuals that lack adequate hand function. The concept utilizes an infrared light source, a simple light-weight plastic retro-reflector mounted on the user's eyeglasses, and a sensor device to capture the position of the reflected light. When combined with an L.E.D. feedback panel, improved keyboard-type of control can be affected.

BACKGROUND

Many severely handicapped individuals either through lack of hand function or deficiencies in their vocal mechanisms, need to be able to control a variety of electronically controlled devices. Once control of these devices has been achieved through on-off switches, or X/Y scanning, the challenge that usually remains is to provide control concepts that increase the speed of access and resulting output of the devices being controlled. Obvious examples are secretarial workstations for high level quadraplegics, or for those lacking hand function due to other neuromuscular deficiencies such as poliomyelitis or muscular dystrophy. Others require access to computer terminals related to vocational pursuits; and still others that are non-verbal due to cerebral palsy require access to communication aids: in addition to output devices such as printers or video displays that can enhance their involvement in the educational process.

Many of these devices are now becoming available. The major deficiencies that remain are lack of portability to permit usage in multiple environments; and the need for increased speed of access so users may be more competitive in vocational settings, or engage in spontaneous communication with more people in their daily lives.

The UT-REC pioneered the development of the Direct Access L.E.D. head mounted sensor for controlling a communication aid and a powered wheelchair. The D.A.L.E.D. control system is now available in a commercial model as a wheelchair tray mounted communication aid. The output can be either printed on paper strip or displayed on a video monitor. The remaining deficiency in the D.A.L.E.D. system is that, due to the optics required, the head mounted sensor is large (2.5 cms. dia. by 12 cms. long) and requires that a cable be run from the display unit to the sensor mounted on the user's head.

Other approaches being investigated in several research facilities is to monitor eye and head positions and use the computed line-of-gaze as a method of control. The development at the Denver Research Institute monitors the position of the eye by use of a light sensitive reflector/sensor device mounted on the rim of the eyeglasses. When the eye position is known and combined with a feedback panel mounted in front of the user, rapid control of a variety of devices can be demonstrated. The deficiencies noted in the ocular monitoring approach is the high cost of the components required to monitor and compute the position of the eye; the difficulty in providing secure mounting on the rim of the glasses, and the need to run a multi-conductor cable to the head.

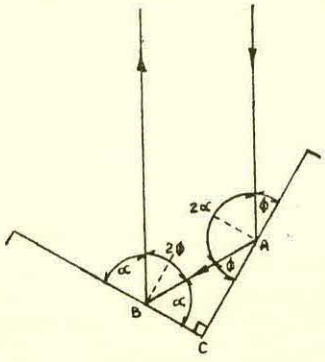
RETRO-REFLECTIVE APPROACH

Concept

The retro-reflective concept is a development currently in process at the UT-REC, that will sense head position without wires or bulky apparatus being run to the head of the user.

Since commercial image sensing devices are too expensive for practical use in the immediate future, the retro-reflective concept utilizes the same image sensing device developed by Dr. Rinard at the Denver Research Institute. The Denver approach is to use a dynamic memory I.C. as an image sensor. This is accomplished by removing the metal lid from the ceramic package and mounting a lens in its place. The retro-reflective

concept requires that the lens be flanked by several infrared emitting diodes which illuminate the area of the user's head. The user wears a small inexpensive retro-reflector, the reflection from which is seen by the image sensor. The retro-reflector is a commercially available array of corner cube mirrors made of plastic. They are very efficient reflectors, and are used routinely in photoelectric controls. A single corner cube in the retro-reflector returns light to its source. (Fig. 1)



RETRO-REFLECTIVE PRINCIPLE

As long as the light source is close to the sensor, the reflected light is seen by the sensor. If the retro-reflector/sensor system is combined with a L.E.D. panel to permit feedback and selection of desired alpha-numeric functions; communication, powered wheelchair, or computer terminal control, can be accomplished by the user. (Fig. 2 & 3)

Technical Details

The memory works by storing charge on 1024 very small capacitors according to the data bits re-written back at least once every two milliseconds or the charge will leak from the capacitors. The charge will leak faster if light strikes the capacitors. The capacitors under the image of the reflector will discharge sooner than the others. If the sensor's position is fixed it can be used to detect the position of the patient's head. An interference filter rejects ambient light from the system. Image enhancemay be accomplished by subtracting the image seen without the I.R. emitters from the image seen with the emitters. It is hoped that such enhancement will not be required.

Current Status

The Retro-Reflective System is currently in the bread board model stage. It is hoped that a working model will be available for presentation by June, 1980. The initial application intended for the system is for use in the control of a micro-processor controlled wheelchair in combination with a communication aid.

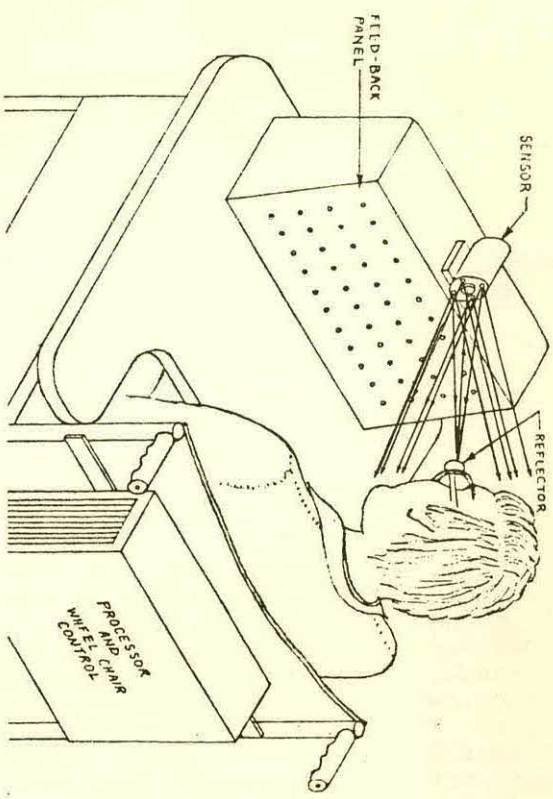


Fig. 2

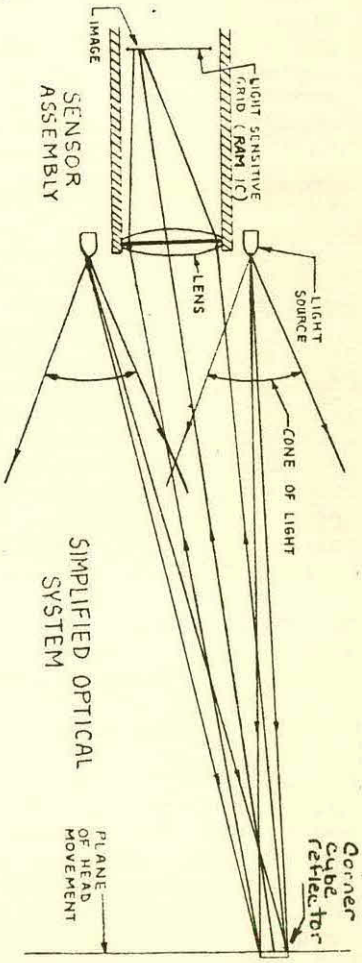


Fig. 3

SHOULDER POSITION CONTROL, AN ALTERNATIVE CONTROL TECHNIQUE FOR MOTION IMPAIRED INDIVIDUALS

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ABSTRACT

A control technique has been developed which utilizies shoulder position information as a command signal. Developed primarily for the high-level spinal cord injury population, the shoulder position controller allows the user to control assistive devices requiring up to two simultaneous proportional command signals. The command signals are derived from a transducer measuring the relative position between the shoulder and the sternum during voluntary movement. A general purpose signal conditioning/interface package allows the system to be easily interfaced to a variety of devices, such as powered wheelchairs and hand orthoses.

INTRODUCTION

Individuals with upper extremity paralysis rely on assistive devices to enhance their independence in daily activities. Greater degrees of motion impairment necessitate larger numbers of and more complex devices to satisfy their needs. These devices typically demand a more involved man/machine interaction which unfortunately is more difficult to achieve as the paralysis claims more control sites.

In the case of spinal cord injury, individuals with injury levels as high as C4 retain some degree of shoulder motion. The concept of the shoulder position control technique is to utilize this voluntary motion as the command signal source for the control of assistive aids. Feasibility studies of shoulder position control were conducted by the Cyberbetic Systems Group, now the Applied Neural Control Laboratory, at Case Western Reserve University (1,2) as early as 1972. Later Peckham (3) utilized a single axis of shoulder motion to control stimulus parameters of a hand orthosis employing functional electrical stimulation (FES).

The hardware for shoulder position control has been redesigned and the new system has been incorporated in systems for wheelchair control and FES. These systems are presently under evaluation by handicapped individuals in our Rehabilitation Engineering Center (REC).

Criteria were developed through a process of interative design by REC staff and evaluation by patients. They are:

1. Ability to proportionally monitor two axes of shoulder motion with minimal impairment of motion;
2. Interchangeability;
3. Non-invasive with ease of mounting;
4. Maximal cosmesis;
5. Low power consumption;
6. Minimal cost.

This meets many of the design features desired by both consumers and clinicians (4).

SHOULDER POSITION TRANSDUCTION

A miniature two orthogonal axis proportional position transducer measures the relative position between the shoulder and the sternum. The shoulder position transducer is mounted on the user's chest with a harness or double-side adhesive tape. A position-sensing arm extends from the transducer body and its end is secured to the shoulder with tape (Figure 1). The transducer's outputs are proportional to shoulder elevation/depression and protraction/retraction, relative to the sternum.

To accomodate device mounting, postural changes, and fatigue due to maintaining a particular position for a length of time, the shoulder position

control system incorporates a Resettable Automatic Zero Point (RAZP) function. With this, the user can define his shoulder position upon activation of a switch as the zero reference or null position. Shoulder movement from this null point, which can be anywhere in the individual's useable range of motion, will generate an appropriate output. The zero point is redefined each time the switch is activated.

Figure 1. The shoulder position transducer.

HARDWARE

The shoulder position control system is comprised of two basic elements, the position transducer and the signal conditioning/interface electronics.

Transducer

Electromagnetic coupling is used to realize a small, rugged, two axis position transducer which will exhibit negligible wear with normal use. Briefly, the transducer consists of a matrix of sensing coils (fixed to the transducer body) and a single driver coil. The driver coil is contained inside the ball of a ball and socket in the transducer body. The position sensing arm is attached to the ball. The driver coil is excited by a low-level high frequency current. As the position sensing arm is moved with reference to the body of the transducer, the energy coupled into the sensing coils varies. The differential output voltage of a symmetric pair of coils is a measure of single axis motion. The transfer characteristics for a single axis are shown in Figure 2.

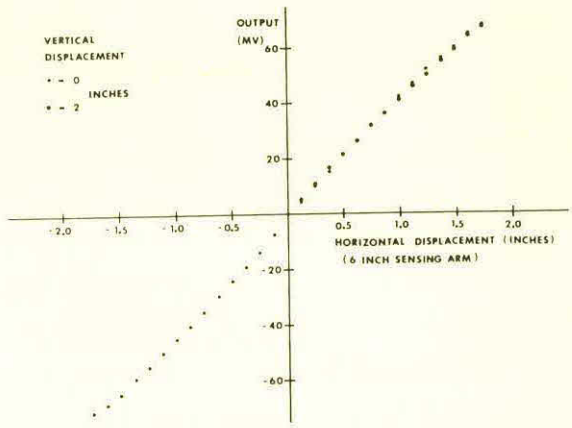

Figure 2. Transfer function of the shoulder position transducer for horizontal movement.

The complete transducer weighs 45 grams including the cable and connector. The transducer body measures 1.5 cm x 1.5 cm x 2.5 cm. The telescoping position sensing arm will accommodate a 7 cm change in length and has a lateral range of 20 degrees in any direction from its center position.

Electrically, for a driver oscillator circuit input of 8 volts D.C. at 1.75 mA, the output signals are on the order of 10 mV/degree D.C. Noise is minimal and has presented no problems to date.

Signal Conditioning/Interface Electronics

The electronics package has been designed with flexible signal conditioning and several operating modes to allow the shoulder transducer to be easily interfaced to various devices. Figure 3 illustrates the signal conditioning for one of the two channels. First the transducer output is amplified and low pass filtered. The polarity for each axis can be reversed by simply moving one resistor. The signal is then RAZPed. A variable width deadband around the RAZPed zero reference follows. This allows small movement around the null position without initiating an output. Variable gain is next which adjusts the transducer sensitivity to position change. Finally the signal is clamped with a variable limiter to allow maximum outputs to be set. Several outputs are available. Adding two jumpers make the system compatable to Everest and Jennings powered wheelchair systems.

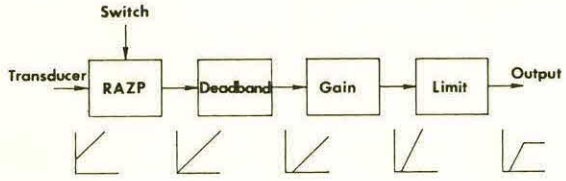

Figure 3. Shoulder position transducer signal conditioning for a single axis.

Two switching modes are possible and also jumper selectable. In the toggle mode the system switches back and forth between an idle and active state each time the mode switch is activated. In the idle state outputs are zero independent of the user's shoulder position. Upon entering the active state the RAZP function redefines the user's current shoulder position as null. The output is then dependent on shoulder movement away from the null position. Operating the switch again causes the system to return back to an idle state.

In the momentary mode the signal is RAZPed each time the mode switch is activated and the output is always available.

The mode switch circuitry is capable of detecting open or shorted cables and up to four redundant mode switches can be used to alleviate possible problems due to mechanical switch failure.

The electronics are packaged in a 6 cm x 15 cm x 15 cm enclosure. The transducer and signal conditioning/interfacing electronics consume 30 ma at 12 volts not including indicator lights and power relay which are used with the wheelchair application.

APPLICATIONS

Two systems successfully controlled with shoulder position control are presented as well as some tenative applications.

Shoulder Position Control of Powered Wheelchairs

The primary application of the shoulder position controller by our center has been in the area of powered wheelchair control for C4 level spinal cord injury persons. The joystick of a conventional proportional powered wheelchair (e.g. the Everest and Jennings "3P" series) is replaced by the shoulder system as described previously. The system is "plug compatable". That is to say the system simply plugs into the joystick control cable and utilizes the existing wheelchair electronics without modification. This lends itself to convenient interchangability of controllers.

The mode switch which controls the idle/active state of the system can take the form of a headrest switch, a chin switch or an EMG switch for example. On the prototype wheelchair it is a headrest switch 10 cm high and as wide as the backrest. The user need only exert a small force anywhere along its length to activate it. The shoulder position wheelchair control system is shown in Figure 4.

Figure 4. Application of the shoulder position controller for control of a powered wheelchair.

The wheelchair control system operates in the toggle mode. The system powers up in the idle state. As the operator activates the headrest switch the system RAZPs and toggles into an active state. Voluntary shoulder movement from the defined null point will cause the wheelchair to move. Elevation/depression of the shoulder controls the forward/rearward speed and protraction/retraction controls the turning rate (into the direction of the shoulder motion). The mode switch is again operated, at which time the system will toggle back into the idle state. The zero point is redefined each time the system enters the active state.

To date a total of four C4 level quadriplegics have used the shoulder position controller to control their powered wheelchairs with no difficulties. It has been well accepted by these individuals because it is cosmetic and does not interfer with head movement and associated activities. It alleviates relative motion problems (such as chin poking) inherent to the VAPC chin controller on rough terrain.

Data from these individuals show an increase in both range of motion and fatigue resistance with use.

Control of Upper Extremity Orthotic Systems

Progress in our REC's core research area of upper exremity functional electrical stimulation has resulted in the development of small stimulators for control of hand function in high-level spinal cord injury persons. The two systems that have been developed provide preshension and release in the C5 quadriplegic hand and lateral pinch and release in the C6 quadriplegic hand.(5).

A single axis version of the shoulder position transducer is used as a proportional control source. Shoulder elevation controls the finger flexor stimulus and depression the finger extensor for prehension and release respectively.

The stimulator contains the electronics necessary for the rezero function as well as a hold function. The hold function allows the user to maintain a fixed stimulus output while performing a desired task. The RAZP, hold, and on/off function are controlled by a two level EMG signal obtained from a site with some remaining voluntary activity.

We are currently investigating the feasibility of using the second axis of the shoulder transducer as a second proportional control source for multifunction stimulation or as the control logic signal source for the RAZP, hold, and on/off functions.

Future Applications

Plans are underway to use the shoulder position controller as an input to a video game. This should serve two purposes. First it provides a simple and inexpensive training aid for clinicians to instruct and evaluate prospective users of the shoulder position control technique. The game not only would provide motivation to learn, but also be a viable recreational activity in the long term.

Other applications are to environmental controls and communication aids. Shoulder position control offers the severely motion impaired individual an alternative to more conventional control schemes available.

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SPAR-OCCC TELEMANIPULATOR FOR POWERED WHEELCHAIRS
SOME PRELIMINARY FINDINGS

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ABSTRACT

Studies have been conducted at the Ontario Crippled Children's Centre (OCCC) to demonstrate the practicality of using wheelchair-mounted telemanipulators as rehabilitation aids for some of those patients with major loss of upper extremity function. Two prototypes were developed; the first at OCCC (I) and the second at SPAR Aerospace (II) with OCCC participation. User experience with these manipulators to date has been short, but the initial results have been of great use in more clearly defining the possible benefits and technical requirements. A formal evaluation protocol has been established. However, the question of practicality remains unanswered.

INTRODUCTION

In Canada there are about 1000 quadriplegics(1) who have high level lesions and an equal number of Muscular Dystrophy patients in their early teens and twenties who could benefit from remote manipulator technology.

Manipulator aids can be grouped into the following categories:

1. Orthotic Aids: There are many relatively simple technological aids, such as handsplints, mouth sticks, reaching tools, adaptive handles, and other mechanical devices which can extend a patient's residual physical abilities. In general, these are the most adaptable, simplest to use and least expensive. Their use should always be attempted first, with the addition of more sophisticated devices only if absolutely warranted.
2. Environmental Controls: There are a number of electronic aids on the market which, through custom designed interfaces, connect the user to a display panel which provides access to a number of pre-programmed functions. These systems are useful when critical tasks require switching operations and can be conducted within a local fixed environment.
3. Telemanipulators: Several types of manipulator systems are under development and these can be classified into six categories based upon their mode of operation as seen in Table I. In this table [specific] relates to "a device designed to perform specified tasks such as page-turning or feeding" and [general] relates to "a device designed to provide a level of manipulative skills to be applied for tasks of the user's choosing."

Table I Classification of Telemanipulators

Purpose	Mode: Fixed Base	Portable	Self Propelled
[Specific]	A	B	C
[General]	D	E	F

4. Functional Electrical Stimulation: Research at several other centres is now being focussed on the possibility of providing an artificial neural system to stimulate paralysed muscles in appropriate sequences in order to regain functional upper extremity movement.(2) This work is generally in its early phases an it is not expected that practical devices or systems will be widely available for some time.

Telemanipulator Systems

A decision was made to pursue the development of a general, portable manipulator system (see Table 1), as this provided potential users with the widest range of functions and could be modified for use by any of the pertinent disability groups.

Successful utilization of telemanipulator systems hinges on appropriate technology which is considered available at this time. The essential issue for address relates to the definition of pertinent design criteria.

In these authors' view there are four main subsystems in any telemanipulator system:

i) the manipulator arm assembly (MAA)
ii) the terminal device (TD)
iii) the control system (CS)
iv) the telemanipulator/teleoperator interface (TTI)

These subsystems for the two prototypes alluded to earlier are described in Table II.

Telemanipulator (I) was designed for a specific patient. This individual suffered from spinal muscular atrophy and, although severely handicapped, led an intellectual and active lifestyle. The goal of this pilot project was to gain as much independence as possible for this person.

Because there was only a slight amount of movement in the right hand available for input signals to the control system, an interface was designed to exploit this limited motion by using ultra-sensitive microswitches which were activated by a specially designed reaching aid. The same aid was used to control the powered wheelchair and to provide some manipulation of objects resting on the surface of the wheelchair tray. The telemanipulator was located on the forward corner of the tray and was able to position the TD within a two foot radius spherical operating envelope.

Telemanipulator (II) was intended to be adaptable to many of the commercially available powered wheelchairs as it was designed to be mounted and stored on the side of the wheelchair frame when not in use. This format provides compatability with either desktop or tray use. The entire system was interfaced to the operator through the joystick of an Everest and Jennings model 3P wheelchair, and was controlled through logic gating circuitry which routed power from the chair's own proportional controller to the drive elements.

Thus, the user had to be capable of operating the regular wheelchair joystick. For this reason three quadriplegic subjects who had lesions at the C5-C6 level were chosen for the initial evaluation. They were not representative of the most needy user group but were able to cooperate in this initial test of the system and its evaluation protocol.

EVALUATION TECHNIQUES

The recipient of the first system became familiar with its operation during its development and participated in a series of informal bench-mounted trials as the device was finalized. In the first stage of use in the home environment, the telemanipulator was mounted on a small workstation cabinet and set up close to the subject's bookshelves. There followed several weeks of unsupervised use during which the operator worked with the system as an assistive device for reading activities. In the final stage of use, the telemanipulator was mounted to the wheelchair tray and was powered by the wheelchair batteries.

Within the first week of use the subject overloaded the system and the resulting technical difficulties (burnt-out motors) led to a halt in the testing. Interviews were then held to ascertain the subject's impressions of the range of possible applications and limitations inherent in this design.

The evaluation of Telemanipulator (II) was extended to include a formal experimental protocol based upon clinical psychology techniques. Different task forms were used to record and measure the efficiency of the telemanipulator system in completing set tasks. Objective parameters such as completion time, number of errors and number of mode changes were recorded. The observer's impressions of the efficiency with which the system was used and degrees of frustration and satisfaction experienced by the subject in carrying out a trial were noted. The subject's own impressions were then assessed through a structured interview form completed at the end of each task.

Training began with practice movements of the TD, followed by a targeting routine. Standard tasks were then performed in subsequent training sessions. These were: Task 1; a) pick up a glass, b) bring it to the drinking position; Task 2; a) pick up a book, b) bring it to a reading position. The final stage was to have involved the subjects having unsupervised use of the system in their home environments, but this stage was not reached due to technical difficulties.

RESULTS

Although the time available for subject use of the prototype systems was limited, the results and experience obtained were valuable. The progress of the research programme was marked both by continuing technical development of the telemanipulator system and deeper understanding of the possible benefits such systems have to offer.

The subject using Telemanipulator (I) surprised the investigators with the innovation of tasks considered important from the user's point of view but not anticipated by others. One such situation was the manipulator assisted movement of the flaccid hand to provide relief from the discomfort of immobility. The interviews with this user highlighted frustration around the inefficient control system and operator interface.

The formal evaluation of the SPAR-OCCC prototype also provided data on the ability of subjects to develop skills at operating the system and an opportunity for the observer to monitor user acquisition of efficient control strategies. All three subjects were able to improve their task completion times during their training sessions in an approximately exponential fashion (see Fig. 2). Maximum performance was not reached within the relatively short trial period available during this project. All subjects were able to complete the first task (picking up a glass of water), with the best times between 2 and 3 minutes. Although this task time is several orders of magnitude slower than that required for an able-bodied person, it is a great improvement over not being able to do it at all. Each subject tended to require fewer changes of motor

Table II Subsystem Descriptions

SUBSYSTEM	OCCC PROTOTYPE (1)	SPAR-OCCC PROTOTYPE (II)
MAA	-anthropomorphic -10 lb. payload -electromechanical 12 VDC -direct gear motor drive -counter sprung joint at elbow flexible cable drive for wrist -wrist, 2 DOF custom built	-folding extensible -5 lb payload -electromechanical 12 VDC -gearmotor, chain and band drive -ball detente clutch at elevation, extension and rotation (incomplete) -wrist, 2 DOF Otto Bock components
TD	-modified electric hook, finger plates	Otto Bock 12VDC hand, finger plates
CS	-on/off microswitch control -limit switches at maximum limit of movement	-analogue input signal -pulse modulated control of motors -mode selection and motor control combined -reset switch for mode select
TTI	-compound micro switch interface, several independent actuating levers -wheelchair control independent of the telemanipulator system	-proportional joystick with custom reset micro switch, head activated -LED display of operational status, indicating the activated pair of motors selected for coordination by the operator (including the wheelchair)

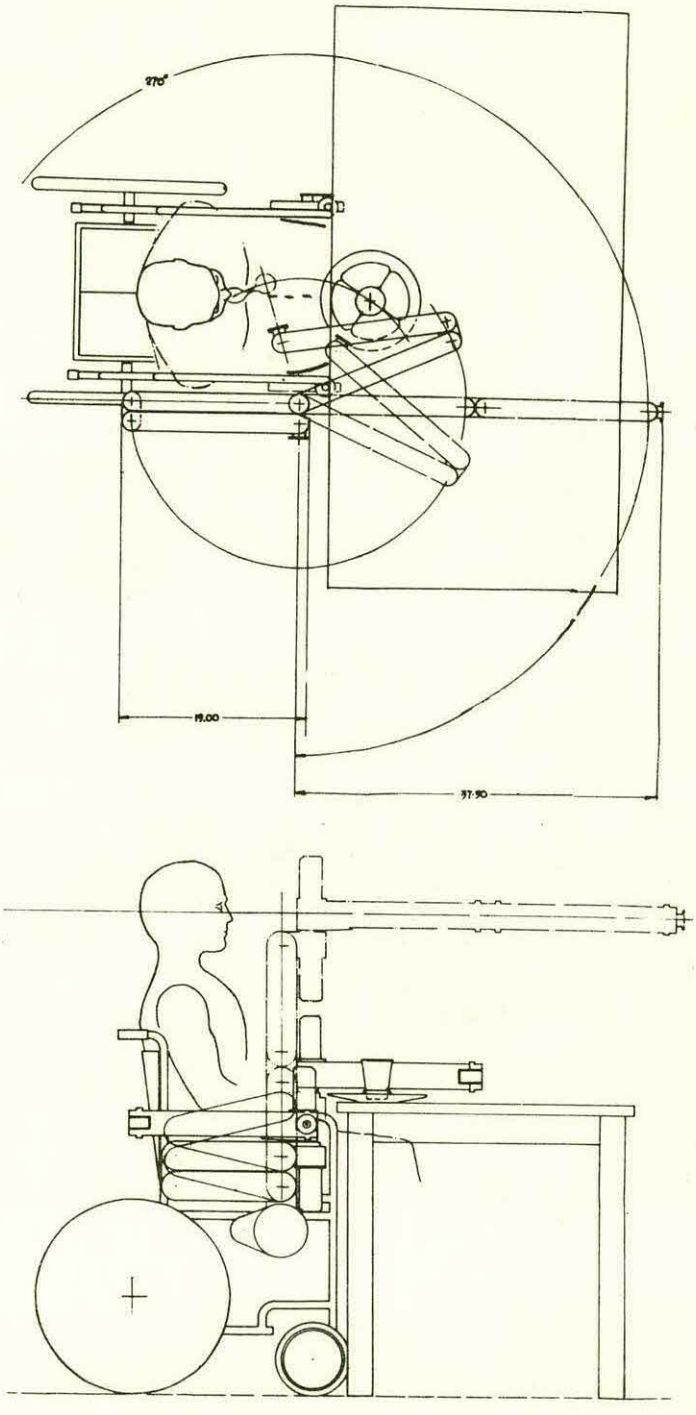

Fig.1 Operating Envelope
SPAR-OCCC Telemanipulator

pair selections as their proficiency improved, suggesting that their choice of control tactics as well as their skill at operating the interface improved with experience. None of the subjects reported experiencing great frustration during the trials.

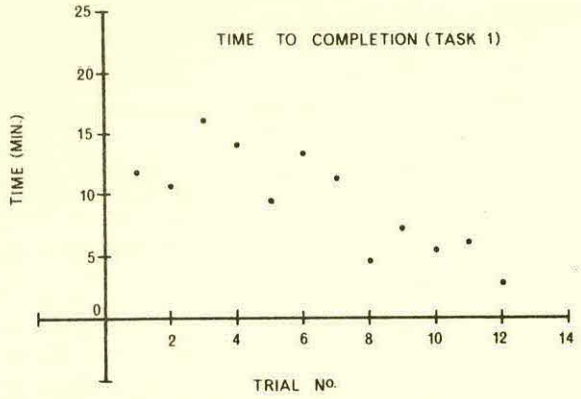

Figure 2 Improvement in task completion time over trials for subject A. The learning curves for the other two subjects followed a similar pattern.

DISCUSSION AND CONCLUSIONS

Two aspects of research must be fully investigated in order to determine whether telemanipulators can play a significant role in the rehabilitation of the severely handicapped patient.

First, a safe and reliable telemanipulator system must be completed. The improvements presently needed are: (1) complete clutching of the drive system for mechanical protection of both the manipulator and the operator; (2) design development of a wrist and prehensile device. Our experience with the prosthetic components now used has demonstrated the need for a high degree of compliance in grasping irregularly shaped objects. This same problem is present in the design of prosthetic hands, but the user of a telemanipulator is at a relative disadvantage as he or she must rely on a more remote control system. This area represents a tremendous design challenge with potential benefit in the prosthetic field.

Second, further analysis of trials is needed in order to estimate the full impact of this technology on those disability groups which stand to benefit from it. The routine of training and individual task evaluation followed by extended individual use of prototype systems will help to generate this information. Further basic research into the problems of interfacing is necessary in order to determine how efficiently information can be exchanged between operator and machine. This area of research has broad applications within the areas of communications and mobility in rehabilitation. There are many technical hurdles yet to be overcome before a practical system can be made available, nevertheless, it remains an exciting possibility.

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PREHENSION ORTHOSES FOR QUADRIPLEGIC PATIENTS

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ABSTRACT

Motorized prehension orthosis are still suffering a high level of rejection by the quadriplegic population, who frequently prefer to use simple splint adaptations. This situation may appear paradoxal when one considers their prehension deficit. There are many causes for this rejection which generally cannot be credited to the lack of patient motivation, but more on the inability of the present orthosis to fulfill the quadriplegic's expectations.

In this project our objective is to provide the quadriplegic with C5-C6 lesions with an adequate and useful prehension function.

This paper covers firstly the driving mechanism for a motorized orthosis and secondly, the structural mechanism of the orthosis. Two structural approaches are presented. The first uses a refined version of the classical tridigital prehension. The second uses a new prehension mode offering a pattern of motion similar to the "palmar prehension".

DRIVE MECHANISMS FOR PROSTHESES AND ORTHOSES

It is probably superfluous to mention that the driving mechanism is a critical element in any externally powered prosthesis or orthosis. To qualify this assumption, it may be worthwhile to remember that a good actuator for all externally powered prostheses or orthoses should be light and small and be able to produce smooth, fast and yet powerful movements while minimizing the energy consumption.

A drive mechanism utilizing a permanent magnet D.C. motor followed by a straight gearbox cannot meet such requirements. The addition of a mechanical "feature" which would relate both the speed of displacement and the force of prehension should represent a solution to the problem. Adding this unit to a drive mechanism does the following:

1) It minimizes the energy required to perform a given task since, assuming a low coefficient of friction in the moving parts, little energy is demanded for the entire working range of the unit (i.e. from high speed with low torque to low speed with high torque). This means that the motor size and weight can be appreciably reduced.

2) It maximizes the prehension parameters such as the grasping force, opening and closing speed and opening range.

3) Combined with a good control system, it can improve the quality of motion particularly if continuous changes in speed and torque are provided.

Referring to the biomechanical model one can see that the work performed versus the energy required for a given task is continuously optimized by the following:

1) By the biomechanical properties of the muscle alone.

2) By the conjugate action of two or more muscles acting around a given joint.

3) By the bone and joint geometry.

One could analyze the biomechanical property of the muscle alone, using the well known curve derived from Hill's Equation. This curve, represents an equilateral hyperbola in which the velocity is plotted against load during an isotonic contraction. The relation between velocity and load represents on its own, an interesting example of energy optimization, which takes advantage of the fact that as more force is required to do work less speed or displacement is needed. The reason for introducing the biomechanical model is not that we believe that we can reproduce it, but we believe that the concept relating speed and force or torque about an axis in proportion to the load is not yet sufficiently exploited in prosthetic or orthotic actuator design. Ideally such design should relate continuously both the speed of displacement and the force of a given body segment, to the resistance opposing the motion (1). This approach will then contribute to a smooth and more harmonious motion of the activated body segment.

In the industry, the above concept is widely used, at least in the steo-by-step gear-shift approach The continuous version is not as widespread but it is found on some machine tools and occasionally on motorized vehicles.

PROSTHETIC AND ORTHOTIC STATE OF THE ART

The continuous approach as described above is not to our knowledge, used at present in any prosthesis or orthosis. A simple torque converter with a limited working range may represent an exception to this usage. This unit has been developed at the Rehabilitation Institute of Montreal (2) and will be described briefly in the following paragraphs. On the other hand the two speed approach has been gaining interest since its introduction on the old valduz hand.

In this hand, the gear shift from low to high force was obtained only when the finger reached the object to be grasped at a given minimum velocity. This represents a serious drawback since it reduces the real effectiveness of the mechanism and restricts the control to the on-off mode.

More recently Otto-Bock introduced its version which is still in use in their hands. The gear-shifting method used in this hand does not depend on inertial forces as on the Vaduz model. It occurs following the coiling of a spring which begins at a predetermined load. This approach has proved to be efficient despite the delay introduced by the coiling and uncoiling of the spring, especially when flexible objects are grasped.

An interesting alternative to the gear-shifting method has been developed and described under the name of "synergetic prehension" by D.S. Childress of Northwestern University (3). It requires two sets of motor and gear-boxes. The first set drives the fingers to produce rapid excursion at low force. The second set drives the opposing thumb and produces high force with a small excursion.

ACTUATORS DESIGN

The activating mechanism together with the orthotic structure present a challenging design problem. In fact the realisation of an orthotic device, and particularly a prehension orthosis, requires design efforts so as to maximize the prehension parameters while minimizing the bulk and weight factors, which of course are more critical than in hand prostheses.

Our investigations of actuators includes several attempts in both the continuous and the gear-shift approaches. All have been built with a D.C. motor coupled to a speed reducer gear-box and followed by a torque producer mechanism which, is an attempt to simplify the following description, has been termed a "torque converter". In the continuous approach a few concepts have been tried. Their miniaturized versions have failed to offer an acceptable level of efficiency or working range. As previously mentioned, a limited range torque converter unit has been produced. This unit was installed on a prehension orthosis made for a C5 quadriplegic patient. The mechanism and its working principle are rather simple. Supplied with a constant force from the drive mechanism, the torque converter mobilizes the finger segments through a preloaded spring located along the proximal phalange of the index finger. This spring elongates as the force of prehension increases. The elongation is then used to increase proportionally the leverage i.e. the moment of force acting on the metacarpophalangeal joint. The cosmesis of the finger splints was somewhat affected by the spring dimensions, but it's elongation has proved to be a valuable source of visual feedback of the force exerted. This fitting and the analysis of the effects on the prehension parameters have permitted us to determine that a torque gain of approximately 5 will probably be adequate for a prehension orthosis. Consequently, a new design combining the mechanical advantages derived from levers and gear-shift mechanism was initiated.

A first prototype was constructed in December 1976 (4) and a second in 1977-1978 and, finally a refined version was completed in 1979 (5). This last version, including the motor and gear-box, is now self contained (see fig.I)

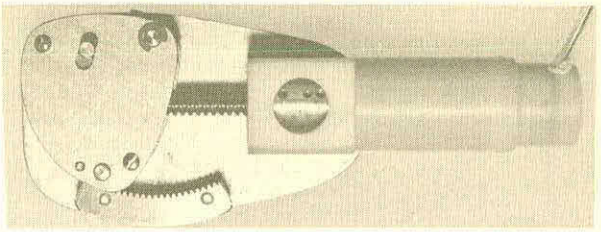


Figure I. New drive mechanism and torque converter assembly weighing 103 gm.

A small production series has been started in order to proceed with a clinical evaluation.

WORKING PRINCIPLE OF THE CONVERTER

Referring to fig.2 Fi represents the force coming from the motor and speed reducer assembly and Fo represents the output force activating the fingers. The working principle of the torque converter can be summarized as follows:

In the unlaoded state, the high torque lever L2 is locked with respect to the low torque lever L1 and the planetary gear G1 rotates freely over the internal gear segment. These conditions cause L1 to pivot around 0 (fixed axis) and this yields an output velocity and displacement which are twice as great as at the input.

In the loaded state, L2 unlocks and gets linked to the gear G1 which now rolls over the fixed gear G2 and drives L1 at a reduced velocity but with higher force.

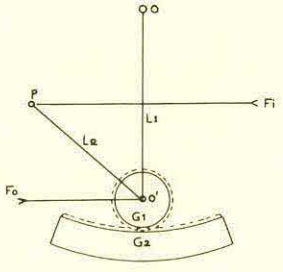


Figure 2. Working principle of torque converter and layout of the various mechanical parts.

Neglecting the relative contribution of the angular changes, the gain in force from low to high can be approximated by the following equation:

$$\frac{Fo\ high}{Fo\ low} = \frac{L1}{r}$$

where: r is the radius of G1

L1 is the distance 0-0'

STRUCTURAL MECHANISM OF THE PREHENSION ORTHOSIS

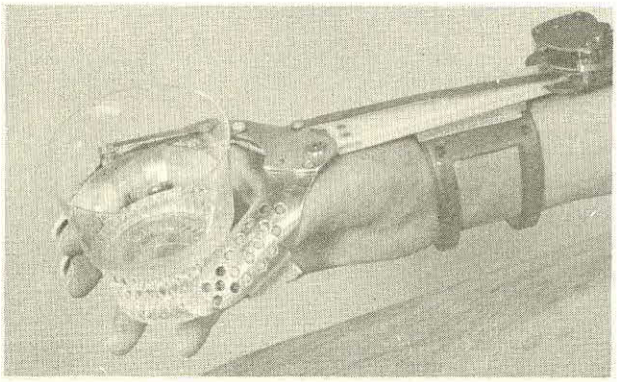
Before dealing with the orthotic structure it may be interesting to remember that complex synchronized musclo-skelatal actions take place

during all prehensile activities. These actions are the results of the large association between cortex and hand. Taking an object is not just grabbing but taking it for the purpose of using it (6). A certain sequence of actions and decisions takes place during a normal prehensile activity:

1) positioning the hand over the object
2) choosing the mode of prehension and opening of digits
3) closing the digits and manipulation
4) opening the digits to release the object

For a paralyzed extremity a prehension orthosis cannot meet all of these requirements.

Today's hand orthosis provides just a single mode of prehension which belongs to the prehension grip function and is termed bi or tri-digital pinch. The precision grip using Napier's concept includes three prehension patterns: palmar, tip and lateral pinch. Anyone of these three patterns can be selected for the design of a prehension orthosis. But this selection requires a careful evaluation of the functional gain and the level of mechanical hindrances pertaining to each pattern. This evaluation is difficult, since each approach should be tested on hands representing many types of deformities. In each case, it should be possible to align and stabilize some specific joints while mobilizing some others.



The major attributes of orthoses as listed by T.J.Engen (7) and confirmed by others (8) and by our own experiences (2) can be summarized as follows: An orthotic system should be simple, cosmetically acceptable, provide a good functional efficiency, a passive structural support and achieve a near-normal musculoskeletal alignment while utilizing the skeletal articulations as a necessary component of the powered system. Two other attributes are, we believe, also important for a prehension device. The first is the patient's ability to put on and to remove the prehension device without outside help (6). To our knowledge, this ultimate goal has not yet been achieved. The second is the need to design an orthotic structure which will obscure minimally any residual sensation which may prevail at the level of the thumb and index finger (9).

In our actual orthosis efforts are made to incorporate the above-mentioned requirements.

DESCRIPTION OF THE PREHENSION ORTHOSIS

The present prehension orthosis is the result of several trials and modifications which have been implemented during the project. In the last version (see figure 3) the force from the drive mechanism activates simultaneously the thumb and the finger mechanical structure. The force of activation is applied to the thumb. The thumb movement is then used to activate the mechanical structure of the index finger at both the levels of the metacarpophalangeal joint and of the proximal interphalangeal joint. To the index mechanical structure are fixed specialized ring attachments. Their role is to couple and secure both the index and major fingers in a stable and functional position. This arrangement allows an efficient prehension function based on the tri-digital pinch. Simultaneous mobilization of the thumb and of the finger joints affects the prehension in the following ways and allows:

1) To increase the orthosis opening range. The maximum opening amplitude is now 11cm between finger tips.

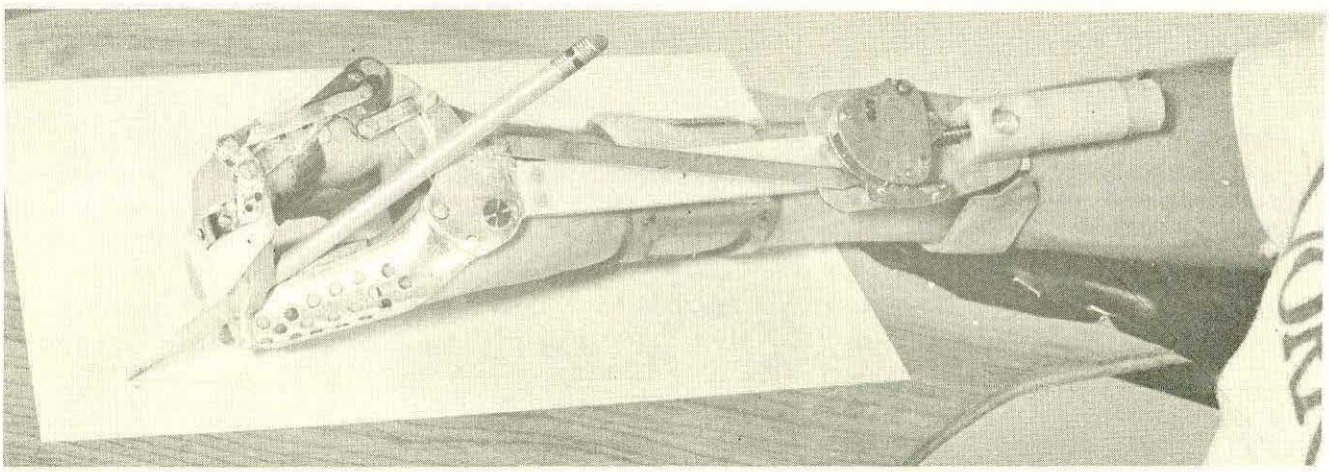


Figure 3. Prehension orthosis fitted to a C5 quadriplegic patient.

2) To increase the maximum closing and opening velocities. Measured at the finger tips, the velocity is now 20cm/sec.
3) To obtain a better transfer of force from the drive mechanism to the finger tips pinch function. A force of 180N from the drive mechanism corresponds to a pinch force of approximately 45N.

This prehension orthosis has not yet been clinically evaluated. The fitting made on an experimental basis on two patients are, so far encouraging. This fitting indicates that some of the inherent problems of the tri-digital pinch approach are as follows: Firstly, the tendency of the fingers is to slip back as the pinch force increases (10). Secondly, related also with the amount of pinch force available is the formation of pressure sores where the orthosis is attached to the fingers or hand. Both of these problems are usually solved by precise fittings which require many custom adjusted parts and much time and expertise on the part of the orthotist. In the light of the above considerations, alternative fitting solutions and prehension approaches have been explored. The results of this exploration has directed our interest toward the experimenting with a new mode of prehension for hand orthosis.

NEW APPROACH TO PREHENSION ORTHOSES

The new mode is very close to the palmar prehension as defined in the Atlas of orthotics (11). In order to fully take advantage of this prehension pattern a new structural approach to the orthosis has been decided upon. In a conventional orthosis the prehension is obtained by moving the fingers from an orthotic structure attached to the lateral side of the hand. In the proposed orthosis the fingers (except the thumb) are mobilized from the inner arch of the palm. The thumb flexes approximately in the same plane as explained for the palmar prehension.

A first prototype is presently being made and the results obtained, so far, as promising. Further developments and refinements are still required before any definite conclusions are drawn.

The following is a list of the advantages expected from this new approach:

1) The orthosis does not require a precise fitting.
2) The fingers do not need to be strongly tied up.
3) It decreases the effect of the variations between the anatomical joints and the mechanical joints.
4) It helps to prevent finger slipage when pinch force increases.
5) It should allow a standardized fabrication of nearly all parts, which, would be made mostly out of light plastic material.
6) The patient may be able ultimately to "put-on" his orthosis himself. This final goal should include, of course, the controlling apparatus.

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A MOTORIZED SOLID AND LIQUID FOOD FEEDING MECHANISM FOR THE INDEPENDENT FEEDING OF THE HANDICAPPED

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ABSTRACT

A motorized mechanism for the independent feeding of the severely handicapped is presented. The mechanism is easily made to operate in two different modes using the same size spoon to feed solids in one mode and liquid food in the other mode. The spoon holds the food at the eating position in dwell for a certain period of time, or as long as the eater requires. In the latter case a signal from a usable limb of the eater starts the cycle. The spoon scoops the solid food at an inclined position as it passes through the food bowl which is automatically indexed, then it holds the food at the horizontal eating position. In case of the feeding of liquids the spoon always remains horizontal. The simple method of synthesis of the mechanism is discussed; and a sample working mechanism, which is very economical for industrial production, is shown.

INTRODUCTION

One of the major goals of rehabilitation of the physically handicapped is to obtain functional independence. Some physical disabilities are often cured through physical and occupational therapy techniques applied in conjunction with other therapeutic measures. Some handicapped have severe physical limitations due to fixed congenitial deformaties which do not respond to physical or occupational therapy. In such instances a therapist looks for solutions through the design of special equipment to attain independent or semi-independent functioning of the handicapped. There are many areas of intervention for the design of special equipment, such as seating, communication, mobility, feeding, hygiene, play, reading, writing, dressing, cooking and homemaking, and recreation [1-4].[1] For the cerebal palsied whose upper limb incoordination may always prevent getting food to the mouth and for other permanently or temporarily handicapped, who cannot use upper limbs, independent or semi-independent feeding becomes a very important activity and its solution is a very worthy goal for the mechanism and machine designers. Although there are some mechanical semi-independent feeders requiring sequence of hand operations for mounting and dismounting, there is no commercially available feeder for independent feeding [3, 4].

The mechanism presented in this article feeds solids as well as liquids such as soup. When feeding solids, the spoon is entered to the food bowl at an inclined position to scoop food, is raised, and brought forward and retained at dwell remaining in horizontal position for eating. The spoon either waits at this dwell position until a signal (from a timer or by a switch actuated by a usable limb of the handicapped) is generated, or it waits at this position through a preset dwell time, then it returns to pick food, feeding continuously. The solid food bowl is indexed during each feeding to maintain a spoon full of food. By disconnecting a pin and inserting it into another bearing joint, the feeding mechanism is altered to feed liquid food. In that case the spoon always remains parallel to the ground. The level of the food bowl is also adjustable.

THE MECHANISM GENERATING THE SPOON MOTION FOR FEEDING SOLID FOOD

The motion of the spoon is of a rigid-body in planar motion. The planar four-bar mechanism can generate the spoon motion in the simplest way. Although a computer-aided optimum synthesis of the four-bar mechanism may be performed to guid the spoon as a rigid body [5], the Hrones-Nelson Four-Bar Linkage Coupler Curve Atlas [6] can be used to determine the four-bar mechanism that generates an acceptable spoon motion. The latter case is used to simplify this design. Mechanical engineers who took undergraduate kinematics courses are familiar with the Hrones-Nelson Coupler Curve Atlas. Each page of it presents a four-bar mechanism and ten coupler curves traced by ten coupler points on a line parallel to the coupler link line AB. See Fig. 1. The mechanisms are shown at the reference positions in which AB is horizontal and AO_2 and O_2O_4 are in line. The position of each coupler point at the reference position of the mechanism is shown on the curve it generates. Each coupler curve is shown in 72 dashes defining the true displacement of the coupler point as the input crank O_2A rotates with 5° increments through the cycle. The four-bar mechanism O_2ABO_4 in Fig. 1 is shown at the reference position. It also shows the connection of the coupler point S that traces the egg-shaped coupler curve used to define a spoon

[1] Numbers in brackets designate References at the end.

path for the feeding mechanism designed.

To synthesize a spoon carrying mechanism one must choose a desirable spoon path first, and find a page in [6] where such a coupler curve is generated. Since the Coupler Curve Atlas [6] does not furnish information on the angular positions of the coupler link, the designer must test to see if the angular displacements of the coupler link provides the desired inclination of the spoon as the spoon travels through the food bowl. This is easily performed by moving a transparent coupler link--which has the spoon geometry centered at the coupler point--to follow the chosen coupler point path as the mechanism moves. An egg-shaped curve having wider portion at the top is preferable to displace the spoon forward and pick food during the travel along the lower portion of the curve. Having wider portion of the coupler curve at the top causes the spoon to tilt very little backwards after feeding. Having the narrow portion at the bottom assures that the spoon experiences large angular displacements when scooping food. Also to be observed is that the coupler link must not experience large angular displacements during the return of the spoon so that the food does not scatter when the eater fails to consume the food. Choosing the wider protion of the curve at the top assures small angular displacements of the coupler link during the upper portion of the return cycle. Thus, Fig. 1 shows the mechanism chosen for this design and the spoon positions it generates from the page 265 of [6]. After deciding the feeding position (horizontal level) of the spoon, preferably close to the upper and wider part of the coupler curve, the frame of the mechanism is defined relative to this position of the spoon. In Fig. 1 the mechanism is shown at the feeding position where it will experience dwell.

The spoon motion generating mechanism shown in Fig. 1 is driven to provide continuous rotation of the input crank O_2A with a dwell at the feeding position shown. The dwell is accomplished by driving the crank O_2A by a coupler curve generated by another four-bar mechanism, where the coupler curve must have a cusp and the input shaft at O_2 of the spoon carrying mechanism of Fig. 1 must be inside this driver coupler curve to guarantee complete rotation of the crank O_2A; and the actuated extension of the crank O_2A must lie along the tangent of the cusp to assure spoon dwell [7]. The two four-bar mechanisms are positioned relative to each other and are connected such that the dwell is generated at the feeding position of the spoon, and that it is the position corresponding to the middle of the spoon dwell motion. The Coupler Curve Atlas [6] also presents many mechanisms generating coupler curves with cusps. The driving coupler curve should be chosen such that the coupler point travels along the cusp to provide dwell of sufficient duration when feeding is done continuously providing independent functioning of the eater. This dwell is also used to activate a control circuit to cause dwell indefinitely until the cycle is restarted by a signal. Figure 2 shows the spoon carrier four-bar mechanism of Fig. 1 connected to a four-bar mechanism generating the driver coupler-curve with a cusp, where their positions correspond to the middle of the spoon

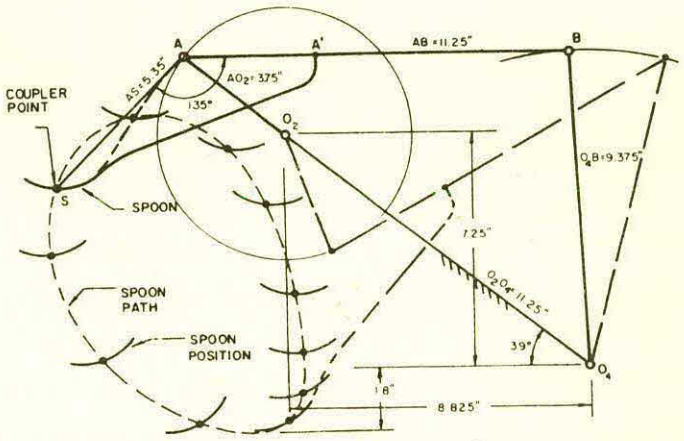

Figure 1 Spoon Carrier Mechanism for Feeding Solid Food, the Spoon Path, and Its Orientation

dwell motion at the feeding position. This driver mechanism is chosen from page 15 of [6]. The cusp provides dwell during 1/8 of the feeding cycle. The crank O_5C of the driving loop is driven at the speed desired. In the working model of the mechanism designed and shown in the photographs, and will be demonstrated during the presentation of the article, a 3.6 rpm electric motor drives the input crank O_5C through a gear reduction unit generating a speed ratio of 1/3.6, providing one continuous feeding (with a dwell at the feeding position) per minute. At E in Fig. 2 is a revolute-prism pair that permits E to slide along O_2F meanwhile rotate relative to O_2F.

The spoon can be connected to the coupler link AB in any convenient manner using an extension normal to AB to provide space between the spoon and the mechanism. It could be connected to some point A' on the coupler link as shown in Fig. 1. In this design as seen in the photographs of the mechanism, the spoon is connected to an extension connected to the coupler link at A. The spoon connection can slide on the spoon support in this design so that one mechanism and one spoon can be used for feeding both solids and liquid by the demonstration model designed as shown in the photographs in Figs. 4 and 7. The food bowl is positioned under the coupler path traced by the spoon on a wooden bowl support. The bowl support is supported by a threaded shaft which passes through a slotted nut that can be loosened or tightened, so that the level of the bowl support can be adjusted as desired according to the bowl used and food being fed. The slotted nut is connected to the indexing disk. See Fig. 3. Indexing is accomplished by means of a slider-crank mechanism displacing a rachet pawl activated by its own weight. The slider crank mechanism is driven by the output crank O_4B of the spoon motion generating four-bar mechanism. The slider-crank mechanism has 1 in. output stroke to displace the indexing pawl performing 20 indexings to cause one rotation of the bowl in the sample design. The stroke of the rachet pawl may be increased to increase the

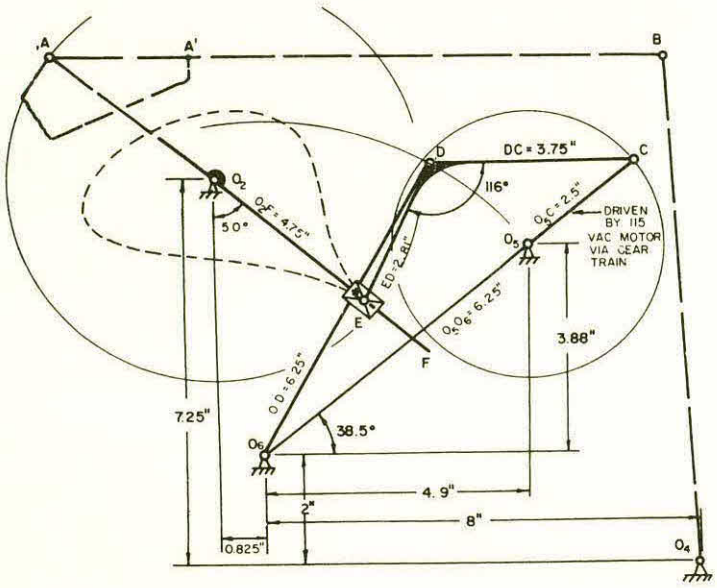

Figure 2 Driver Four-bar Mechanism Providing Dwell of the Spoon

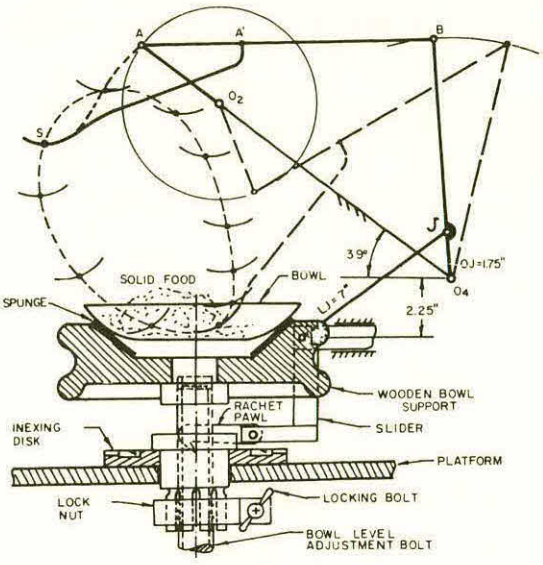

Figure 3 Bowl Holder and the Indexing Mechanism

angular indexing stroke of the bowl. This can be accomplished by moving the connection of the slider-crank coupler link at J on O_4B towards B. This connection can be put below O_4. This requires that the teeth on the indexing disk are reversed to cause indexing by pushing the rachet pawl as the spoon returns.

Figure 4 shows the mechanism at two positions of the spoon when feeding solid food. The switch at G has three positions: the middle position stops the mechanism; the lower position causes the mechanism to feed continuously bypassing the switch H; the upper position causes the cycle to become dependent on an external signal to continue the cycle after the spoon dwells. This signal may be generated by a timer causing an extended dwell period or the eater having partial use of his limbs (hand, leg, foot, shoulder, knee, elbow, etc.) actuates a switch to start the cycle. The switch K provides such a signal in this mechanism.

THE MECHANISM GENERATING THE SPOON MOTION FOR FEEDING LIQUID FOOD

The spoon must remain at horizontal position throughout the feeding cycle when feeding liquids, such as soup, so that as the spoon travels outside the bowl the food remains in it. Also when the eater fails to consume the food, it must be returned back to the bowl without spilling it around. The coupler point motion of a parallelogram mechanism becomes well suited for this purpose. A parallelogram mechanism is formed by attaching link O_7M to the coupler link AB of the spoon carrying four-bar mechanism used for feeding solid food as as shown in Fig. 5, where a double joint is formed at A. When feeding solid food, a threaded pin shaft at A connects link portions AM and AN, thus, connecting O_7M to the coupler link rigidly. No

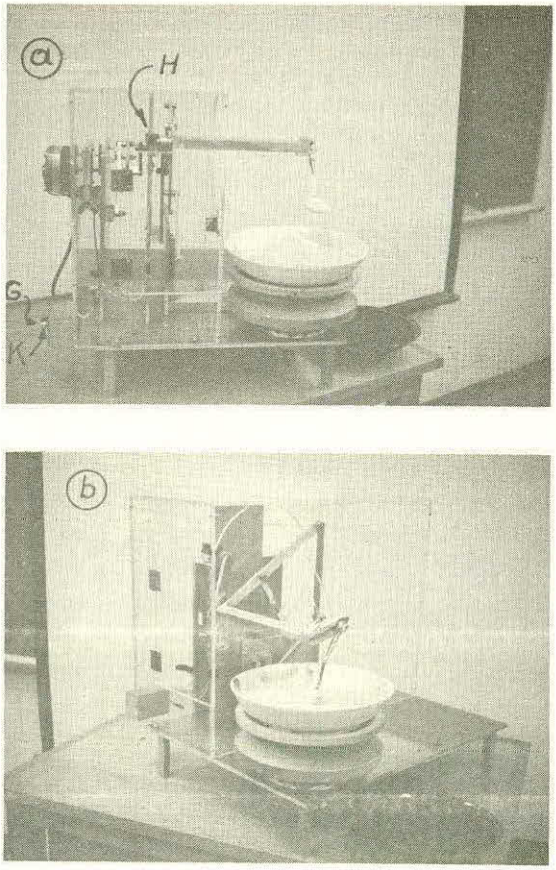

Figure 4 Sample Mechanism Designed Feeding Solid (Dry Rice); (a) Eating Position, (b) Spoon in Scooping

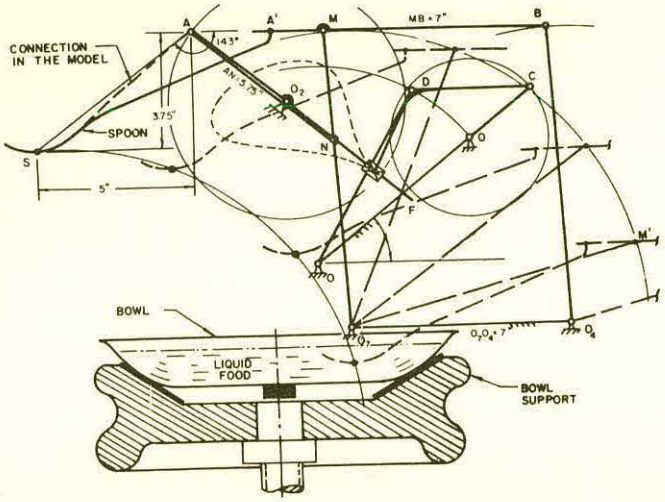

Figure 5 The Liquid Food Feeding Parallelogram Mechanism (O_7MBO_4), its Driving Loop (O_2ANO_7), and the Positions of the Spoon as it Feeds Liquid Food

shaft exists at O_7 then. When feeding liquid, the threaded pin shaft at A is removed and used as the shaft at O_7. This makes the revolute pairs at M and N active, forming the O_7MBO_4 parallelogram loop driven by the same input crank O_2A through the coupler link AN. Hence, the O_2ANO_7 is the driving loop of the parallelogram loop O_7MBO_4. The spoon path in this case becomes a circle of radius O_7M and any point on the coupler link AB can be used to center the spoon. In this design the spoon for liquid feeding is also connected to the spoon carrying extension at A as in the case of feeding solid food, except that a longer soup spoon is needed as shown in Fig. 5. Shown in Fig. 5 is also the path of the spoon when feeding liquid. The mechanism is shown at the dwell position of the spoon, which is the same dwell position used for feeding solid food. When feeding solids, the rigidified link $AMNO_7$ floats as a rigid body uselessly as seen in Fig. 4.

Figure 6 shows the mechanism when feeding liquid food. The control of the spoon motion follows the same circuit sequences as in the case of feeding solids. When feeding liquid, however, there is no need for indexing the bowl. So, the rachet pawl may be lifted to stop indexing. For a long-term liquid feeding, the crank of the indexing slider-crank mechanism may be disconnected at J (see Fig. 3). O_7 may be positioned closer to O_4 to increase the transmissivity of the driving loop O_2ANO_7.

The spoons in this mechanism for both cases of feeding are rigidly attached to the coupler link extension. However, an additional cam-follower mechanism may be used to rotate the spoon slightly when it is picking liquid food. After the spoon is lifted from the bowl, it is rotated to assume its horizontal position. This may be desirable to maintain partially full spoon when it is transferring liquid food. This spoon rotation can

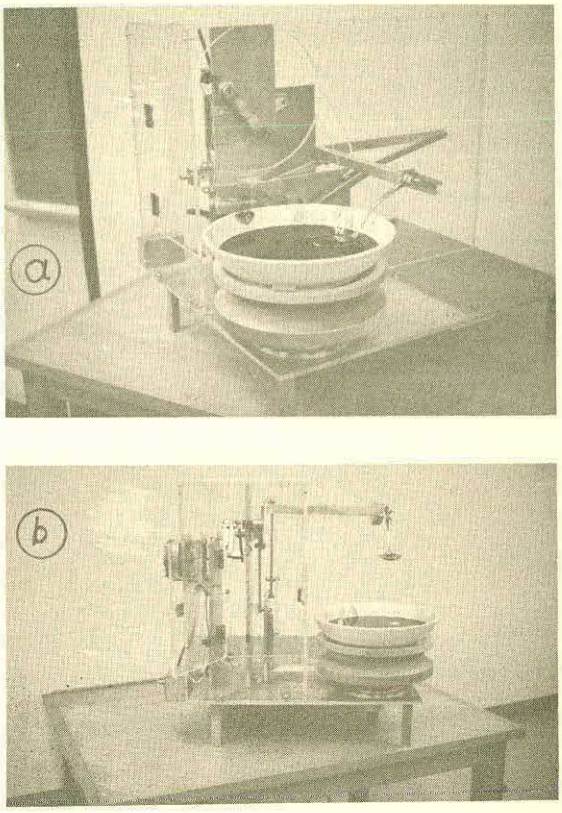

Figure 6 Sample Mechanism in the Liquid Food Feeding Mode; (a) When the Spoon is in the Food, (b) When the Spoon is in the Feeding Position

easily be achieved by connecting the spoon on a rotating head which is connected to a small four-bar mechanism. The coupler link of this four-bar mechanism functions as the follower to rotate the spoon holder 3^o to 5^o as the spoon travels through the end of the bowl during which the input crank O_2A rotates through 45^o. See Fig. 7. The follower coupler link is acted upon by a compression spring to retain the spoon at its unrotated position. The cam that may provide 1/8 in. stroke for the follower coupler link can be mounted on the link AN within 2 in. from N. This addition may be costly but it provides cleaner liquid food feeding.

CONCLUSIONS

Illustrated in the foregoing is a simple design of an automatic feeding machine in demand for the physically handicapped to attain his complete functional independence or semi-functional independence when feeding himself. Simplified design approach using the available Four-Bar Linkage Coupler Curve Atlas leads to a low-cost system for marketing. The full size working sample model designed and shown in Figs. 4 and 6 is not optimized for industrial mass production. Even then, including covers, it weighs 20 lbs., where all the links, floor, and the frame are made of aluminum. It is portable and can be placed on

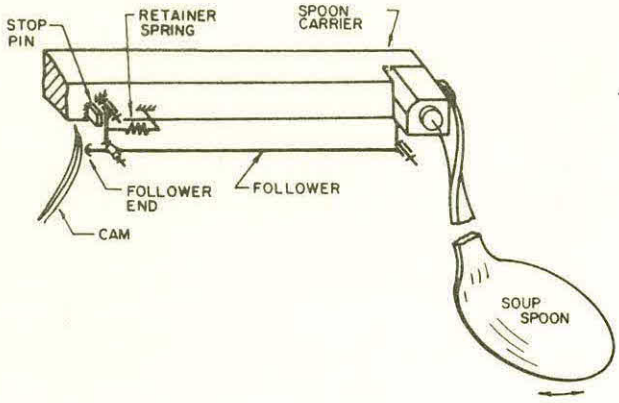

Figure 7 Spoon Rotating Attachment Mechanism to Maintain Partially Full Spoon When Transferring Liquid Food

any feeding table. Its manufacturing cost as a single unit stands about $400 which certainly can become much lower in mass production.[2]

This design example is an illustration of how industrial mechanism and machine designers can solve the problems of the physically handicapped providing them functional independence and fun. As stated in the introduction, there are many problem areas of the physically handicapped where the industrial machine designer can put his talent and creativity to use.

ACKNOWLEDGMENT

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[2]Patent for this feeding mechanism is pending.

MODULAR, PREFABRICATED ORTHOSIS FOR TREATMENT OF ELBOW FLEXION CONTRACTURES

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ABSTRACT

A simple and adjustable orthosis have been designed and prototypes built to aid in increasing the limited range of arm extension available in disability involving flexion contracture at the elbow. The orthosis, positioned 5 degrees larger than the angle of maximum passive extension applies a small extension moment to the arm, slowly lengthens the musculature and leads to a gradual lessening of the contracture. When used in conjunction with therapy it helps maintain the benefits of daily stretching exercises.

Key design features include modularity with discrete, easy to fit and adjustable components, compatible with anatomical constraints without left or right sided parts, simplicity, cosmesis and low cost.

INTRODUCTION

As the result of research and development on upper limb orthoses in the past two decades, restoration of useful function in upper extremity motor impairment has been much improved. Efforts at Texas Institute for Rehabilitation and Research, Rancho Los Amigos Rehabilitation Center and elsewhere have resulted in the development of the modular concept with emphasis on lightweight finger prehension orthosis. However, despite the many available functional devices an acceptable orthosis for the prevention of upper limb deformity due to contracture has not yet been achieved. Flexion contracture at the elbow in particular is disabling in itself and is a difficult problem to correct (6). In certain spinal cord injuries this contracture may lead to reduced patient mobility by preventing reach to the handrim of the wheelchair.

This paper presents the design and application of a simple adjustable orthosis which is to prevent and help correct contractures at the elbow. Prototypes have been fabricated in three sizes for application by therapist and clinical trials are underway.

Methods of treatment of elbow flexion contractures can be surgical or non-operative. Surgically, tenotomy involving the lengthening of the biceps tendon has been successful in improving arm function in paralytic diseases other than spinal cord injury (5). Another surgical technique, using percutaneous electrodes, has also been successful. Mooney (4) implanted electrodes to stimulate the extensor muscle group and reduce the muscle imbalance that caused the flexion contracture at the joint.

Among the non-operative methods, serial or wedge casting of the patient's arm in forced extension has been used with success, although, the weight and bulk of the cast, the pressure sensitivity and inaccessability of the skin are the real shortcomings of this method. Traction is another non-invasive treatment that has been tried (7). Unfortunately, it requires stationary positioning of the patient in a bed or chair which, in spinal cord injury cases, may be inductive for pressure sores and thus is seldom used.

Orthotic devices to correct elbow flexion contractures have been described as well. Goller and Enders (2) have used a dynamic plastic elbow-extension orthosis in five patients with moderate success of an average of 15 degrees reduction in flexion contractures. More recently Green and McCoy (3) have described a turnbuckle orthosis which accomplished in 12 patients an average reduction in deformity of 37 degrees in an average treatment period of 20 weeks. This custom-fitted orthosis has been tried in existing short term contractures of traumatic fracture origin where little or no impairment of skin sensation was present.

EXPECTED BENEFITS

The orthosis at the University of Virginia Rehabilitation Engineering Center was developed with the expectation that this modular system will make the correction and prevention of an elbow flexion deformity easier to be resolved by the therapist. It is hoped that the availability of three sizes in stock components will make possible the application and fitting of the orthosis by the therapist immediately upon need in the clinic.

It is assumed that this orthotic system will be worn by the patient during non-therapy sessions and, thus, maintain the increased elbow extension achieved in therapy. The orthosis is expected to benefit any disability involving a flexion contracture at the elbow with particular emphasis on

C4-6 quadriplegia, brain injury, burns, and some forms of rheumatic diseases.

DESCRIPTION

The modular orthosis illustrated on Figure 1 consists of four components: a biceps cuff with velcro strap, a leaf hinge, and two "book binder" screws. The cuff type can be distinguished by its shape and the number of velcro straps. Also, the biceps cuff has its only strap centrally located while the two straps of the forearm cuff are fastened to each end. Each cuff type is available in three sizes, small, medium, and large.

Figure 1
Elbow extension orthosis used by a C5-6 tetraplegic individual.

The leaf hinge is a thermoplastic material also available in three sizes. The large leaf hinge (46 centimeters long) with four screw holes at both ends is made of polycarbonate and is the strongest. The other hinges having three holes at each end are made of acrylic and available in two lengths of 38 and 30 centimeters. The illustration on Figure 2 shows an assembly and the interchangeable three sizes of the components. The "book binder" screw is a two piece fastener consisting of a slotted threaded screw and its base nut.

Initial designs were using the three point principle for application of correcting forces to the upper limb with the olecranon area as the central point. The principle was abandoned because the excessive pressure applied at the center could result in skin break down.

The current design utilizes force couples (four point principle) acting on the arm segments and produces a constant extension force from the spring action of the plastic hinge. This force application maintains elbow extension by applying slight pressure to the arm.

To prevent tissue ischemia the applied pressure should be less than the capillary pressure of 30 mm Hg. When the angle of the brace is larger than the angle of the contractured arm the brace must bend when applied. The resulting deflection of the leaf hinge will generate the applied force necessary for the extension moment across the elbow joint. For a brace angle of 5 degrees greater than the contracture angle, the applied force, using medium size components, was measured to be 4 lbs. and was calculated to be less than 20 mm Hg of pressure applied to the skin.

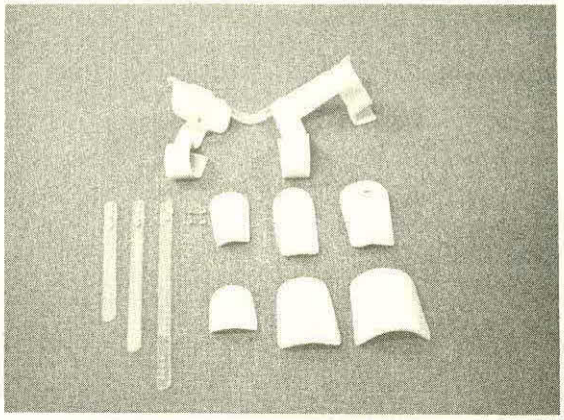

Figure 2
Three sizes and interchangeable components of the modular adjustable orthosis.

APPLICATION

1. Patient Selection. The orthotic system is an adjunct to the patient's therapy treatment of stretching and will assist in the maintenance of the newly acquired range of motion. This orthotic system will be an aid to those patients with flexion contractures at the elbow of 60° or less as defined by the American Academy of Orthopedic Surgeons (1).

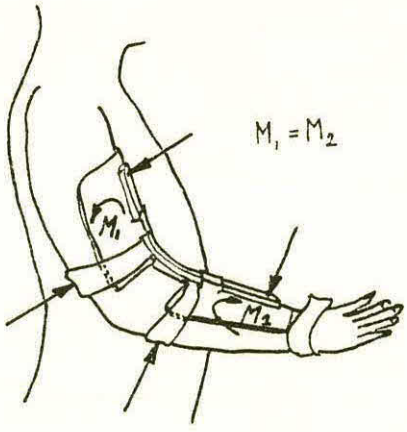

Figure 3
Four point principle of elbow extension orthosis, showing moments M_1 and M_2 maintaining extension. The orthosis can be adjusted by heating the hinge strip with a heat gun.

Because this system utilizes direct pressure from the cuffs to the underlying skin, the therapist must take precautions to periodically check the skin for pressure sores. It is recommended that the therapist apply this device for ½ hour at the first application and gradually increase the wear time if no problems are apparent.

2. Patient Measurement and Components Selection. The proper cuff size is determined from four circumferences and length measurements of the fore and upper arm. The measurements are to be made with the elbow flexed at 90° with the wrist rotated to its neutral position. A conversion table then allows the therapist, to choose the small medium or large cuffs for the orthotic assembly.

Hinge length and stiffness selection are determined by the summation of length measrue-ments.

3. Flexion Contracture Angle Measurements. Measure the flexion contracture angle passively under the effect of gravity. This measurement is used to establish the extension angle of the orthosis. Once this procedure has been accomplished, add approximately 5° to the extension angle to compensate for the leaf hinge flexibility. The result will be the initial extension angle of the orthosis.

4. Setting The Leaf Hinge Angle. While the elbow is passively extended to the contracture angle, place one end of the leaf hinge on the volar surface of the arm proximally to the radial styloid. The other end of the hinge should be lateral to the anticubital fold of the elbow. Mark the leaf hinge in line with the anticubital fold. Once this has been accomplished, mark the hinge 3 cm proximally and 3 cm distally to the fold line.

Heat the leaf hinge between the marks with a heat gun until the plastic is soft and pliable in this region. Place the heated hinge flat on a table surface. While holding the distal (longest) end on the table proceed to lift or bend the other end until the orthotic extension angle is achieved. Hold this angle until the hinge cools off (about 3 minutes).

5. Assembly. First, slide the distal end of the hinge into the forearm cuff tunnel. Next slide the proximal hinge end into the biceps cuff such that the cuff tunnel is near the hinge bend. Position both cuffs on the hinge to fit the contour of the arm. Align the screw holes in both cuffs with the nearest screw holes in the hinge. Mark the position of the aligned holes on the hinge. Remove the orthosis and fasten the cuffs to the hinge with the book binder screws.

6. Fitting And Check Out. Reapply the assembled orthosis to the patient. Test for excessive pressures and cuff misalignments. Make any necessary adjustments. Finally, trim the velcro straps to the patient's need. Again, remove the orthosis and apply the protective sleeve to the hinge.

7. Readjustment. Periodic readjustment of the hinge angle is required with changes of the contracture angle of the elbow. It is recommended that the hinge angle be changed with each 5° change of elbow contracture. To achieve a change in the hinge angle, remove the protective sleeve from the hinge and reheat the area with a heat gun. It is necessary to disassemble the orthosis to accomplish this task.

SUMMARY

An orthosis have been designed to aid in increasing the limited range of arm extension available in disability involving flexion contracture at the elbow. When used in conjunction with therapy it helps maintain the benefits of daily stretching exercises.

Important contributions of the orthosis are modularity with interchangeable components in three sizes, simplicity, cosmesis and low cost. Prototypes have been fabricated for application by therapists and clinical trials are underway.

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A MULTI-POSITION ELBOW DIAL

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ABSTRACT

The report presents an improved method of stabilizing the arm in a balanced forearm orthosis, by providing an elbow dial that moves in synchronization with arm motion.

The components used in building the system are described, and alternate means of construction are suggested. An analysis is made of the design principle, which is a four bar linkage of the double rocking lever type.

The work is of significance in the treatment of any disability requiring balanced forearm orthotic assistance.

MOTIVATION BEHIND THE WORK

The effectiveness of a balanced forearm orthosis to assist the arm depends on the arm remaining fixed in the trough; so that once the pivot point has been set to give mechanical advantage to the weaker muscles, this advantage is not lost by the arm slipping out of position. Straps alone can not secure the arm, so the well known elbow dial is included in the system. This leads to the dilemma which arises when the stationary elbow dial is angled to hold the arm when the elbow flexes. The arm is then restricted in any forward motion.

The purpose of this paper is to report on a system which moves the elbow dial in co-ordination with the motion of the arm: moving it up to a right angle with the trough as the elbow flexes, and allowing it to fall back toward the 180 degree angle as the elbow extends. Figure 1 introduces the system and its component parts.

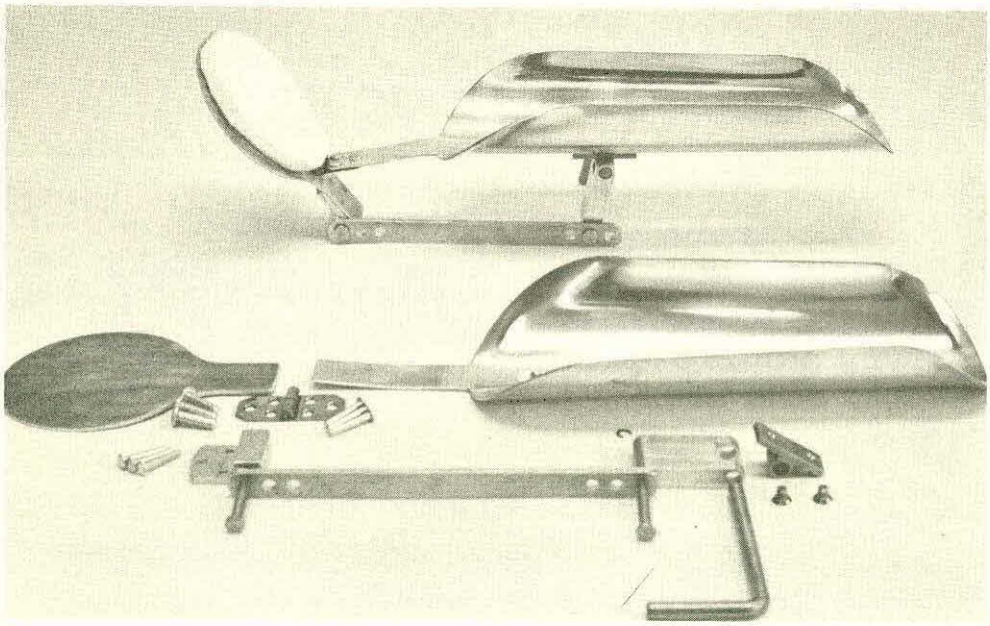

Figure 1

A completed assembly shown with individual components.

The basic concept was invented at the Roosevelt-Warm Springs Rehabilitation Center in Warm Springs, Georgia, by the late Jack Conry. Modifications to that system were added by Bill Crowder, also of Warm Springs. We had wished to duplicate their device, but were not equipped with the supply of stainless steel and monel metal, or the welding capability of their laboratory. This led us to develop a system that can be fabricated of aluminum. In the design process, we found it advantageous to create an adjustable linkage that allows for reassembly possibilities, should it be necessary to change the pivot point of the arm.

COMPONENTS OF THE SYSTEM

Figure 2 shows the individual components of the elbow dial system, which can be identified by the letters as follows:

A. A commercially available balanced forearm trough with the conventional elbow dial removed, and the connecting tongue flattened.

B. An aluminum elbow dial with an extension. The round portion may have the edges shaped and rounded over a lead block before assembly.

C. A commercial hinge which will be riveted to the circular portion of (B) and the tongue of (A). This hinge is point 1 of a four bar linkage system.

D. A channel for a connecting pin with extension to allow riveting the channel to the back of the elbow dial extension.

E. An adjustable link shown with(F). The adjustment holes are the same distance apart as the holes in the BFO trough.

F. Two connecting pins. The pins form points 2 and 3 of the linkage system, and have grooves cut in one end to accept (G).

G. A Tru-Arc ring that facilitates assembly and adjustments.

H. The non-moving link of the four bar system. One end has a channel for the connecting pin; and the other end has both a channel for (J) and a saddle type notch in which (K) will pivot around the rod. This pivot point is number 4 of the linkage.

J. 90 degree offset rod. The short arm positions the entire unit in a standard distal arm of a wheelchair attachment.

K. BFO attachment unit and pivot.

Once the offset rod is inserted in (H) and (K), a small hole is drilled through the wall of the channel of (H), into the rod; and a small pin is inserted to create the immovable condition of the linkage.

After assembly, the elbow dial is padded and can be further upholstered with leather for a finished appearance.

Dimensioned drawings of these components will be available to anyone interested.

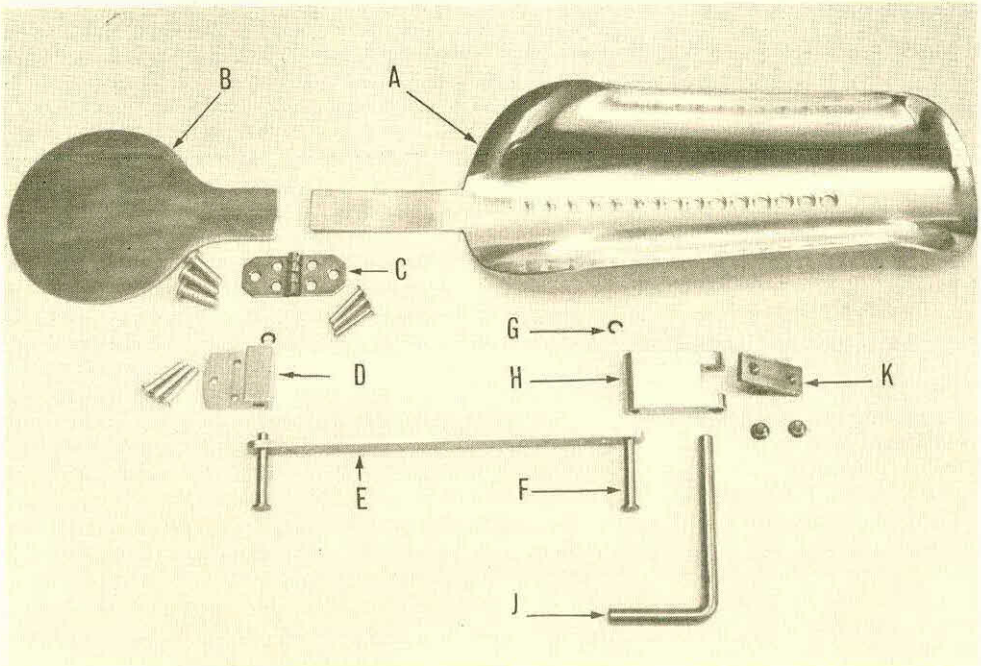

Figure 2

Identification of Components

This system is only one way to create a disappearing elbow dial. The Warm Springs approach is to weld channels onto flat components. A rod of fixed length is then bent to fit into the channels, the end being hammered over to keep it in place. In the system first presented, Tru-arc rings could be replaced by a system of screws in tapped holes. Other possibilities can be developed with materials at hand, conceivably using plastics for some components, as long as the principles governing the action are understood. For this reason, it is necessary to take a few minutes to understand what is happening in the system.

ANALYSIS OF THE SYSTEM

The mechanism is a four bar linkage, of the double rocking lever type. The mystery of these linkages is their obvious simplicity while their analysis is complex. For this reason, the best approach to analysis is a graphical one as seen in Figure 3. Again, point 1 is the hinge, 2 and 3 are the pins, and 4 is the pivot point of the BFO trough. The driving crank is link (1-4), the arm trough powered by the motion of the patient's arm. The follower crank is the adjustable link (2-3). The fixed link is (3-4) which establishes the centers of rotation for the driver and follower cranks. Link (1-2) is the coupler of the system which undergoes both rotation and translation. Our interest is centered on this coupling link. It is an extension of the elbow dial (represented by dotted lines), and its position directly determines the angle of the elbow dial.

The uppermost representation of the linkage (A) shows links (1-2) and (2-3) in nearly a straight line. It is important that they not reach the dead center position. The curve in link (1-4) provides a stop to prevent this. Once the assembly is complete, the tongue of the arm trough can be curved to effect the necessary stop. This uppermost position represents the situation where the weight of the arm rests entirely on the trough, and the elbow needs no control. The dial must be out of the way so that the arm can go through range of motion exercises in this plane and be able to reach for objects if this is within the patient's capability. Not only is the elbow dial out of the way, but the arm can not pivot too far forward and slip out of the trough. The angle at which this stop position is reached can be predetermined and put into effect by graphing in this way and designing components in proportion to the drawing.

Position B shows the configuration of the linkage when the elbow is flexed to bring the hand up, as in eating or grooming activities. The coupler linkage has now brought the elbow dial up securely behind the elbow. Two intermediate positions of the coupling link are also shown.

Figures 4 and 5 show a laboratory situation with the arm in two positions, illustrating the effectiveness of the system in actual use.

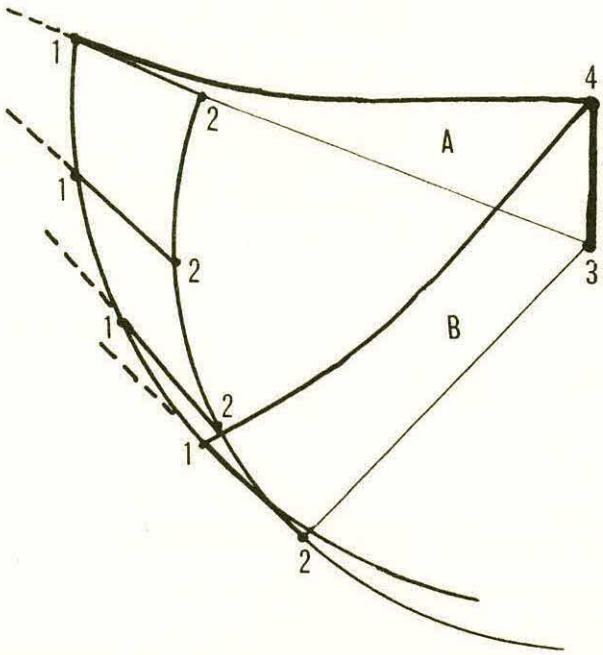

Figure 3 Graphic Analysis of Four Bar Linkage

IN CONCLUSION

We appreciate this opportunity to present the preceding information to the rehabilitation field. We are all too aware of the frustrations that previously existed as we tried to apply BFO's to our patients.

We are indebted to H.G. Bowden and Bill Eason of Warm Springs, Georgia for their co-operation in showing us their design and giving us the historical background. We also wish to thank Barbara Greenfield for her photographic contributions to the paper.

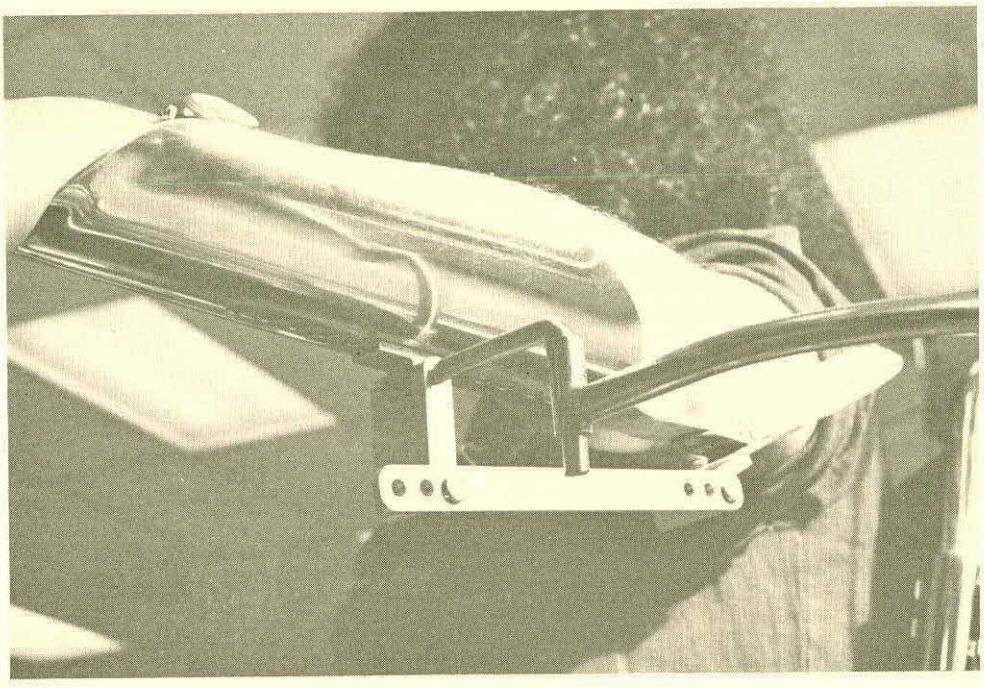

Figure 4 Elbow and Dial in Extended Position

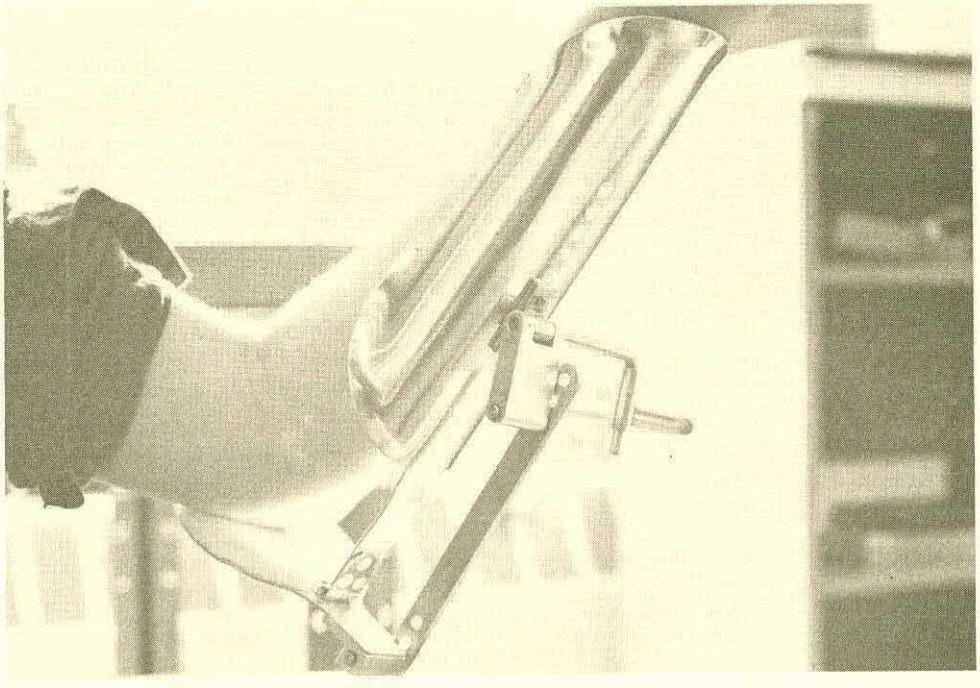

Figure 5 Elbow and Dial in Flexed Position

ELEMENTAL ORTHOTICS

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ABSTRACT

The shape of the human body presents a challenge to the designers of orthoses, prostheses and special devices. Containment of this complex shape can be achieved by use of small standardly shaped repeating elements linked together to form an orthosis. Elemental orthotics presents advantages in manufacturing, clinical practice and user comfort in an elegant manner.

INTRODUCTION

There are six fundamental aspects of measurement in Rehabilitation Engineering (1). These encompass most of the devices and processes that exist or are under development. They are motion, force, shape, properties of materials, neural control and social interactions. Of these, shape is the most basic to prosthetics and orthotics. It determines the interface of the device with the patient. Stumps, limbs and trunks are complex, three-dimensional irregular shapes that need individual care in their approach. Standardization has been achieved with some success in sockets and spinal orthoses, for example, but generally these shapes need a customized approach. However, the shapes have several common factors in terms of the criteria and constraints they impose.

REQUIREMENTS

The basic requirement of an orthosis or prosthesis is that it should provide containment of the body segment so that motion and force can be controlled. The motion requirements might best be viewed as degrees of freedom. A tibia and femur junction, for example, would have three rotational and three translational movements if it were not constrained by ligaments, tendons and other tissues. The orthotic control of this joint could be expressed in terms of how many degrees of freedom should be permitted. This is a function of containment and is achieved by the shape of the container.

The force requirements can be viewed in terms of weight-relief (e.g. ischial bearing A.K. sockets) or in terms of force re-distribution (e.g. the C.A.R.S.-U.B.C. Knee Orthosis (2)). An imbalance of forces results in motion and provision of force to control motion is dictated by the shape of the orthosis. The six aspects of Rehabilitation Engineering are inextricably linked, with the tissue quality, neural control and social aspects (e.g. cosmesis), all limiting how motion and force can be controlled by shape.

DEVICES

A closer examination of the body shape by such techniques as Shadow Moire will reveal that there are many standardly shaped areas on the body surface. These consist of circles, ellipses and even straight lines in some areas. Attempts have been made with modular orthoses to take advantage of these standard shapes, with some success. However, other parts of the body, such as below-knee stumps, are quite complex and irregular in shape. The cladding of such complex areas in nature has been achieved by use of small, similar repeating elements such as feathers, scales and platelets. Application of dynamic group problem solving techniques (3) to this analogy has brought forth the concept of Elemental Orthotics. Elemental orthotics is a system that contains the complex body shape by small similar repeating elements that are linked together to form an exoskeleton. An illustration of this concept is given by Figure 1. This conceptual model shows a two-component system of indoor golf balls and plastic rods, continuously repeated to clad the limb. Refinements of this model are shown in Figures 2 and 3.

We have progressed past the modelling stage and have tested solutions to the problem of seating severely disabled children, as shown in Figure 4. (After all, a seat is just an orthosis that you sit on.) This shapeable matrix for seating the disabled child (4) consists of cylinders and balls strung onto a wire frame. When the wires are slack, the matrix can be adjusted to conform to a required shape, e.g. to provide a central pommel to abduct the hips. When the required shape is achieved, the wires are tightened and the seat is formed. Eight children have been seated in such a manner, the only additional component needed being a thin foam cover and perhaps a restraining waist strap. In one case it was possible to fit a child in his home in two hours.

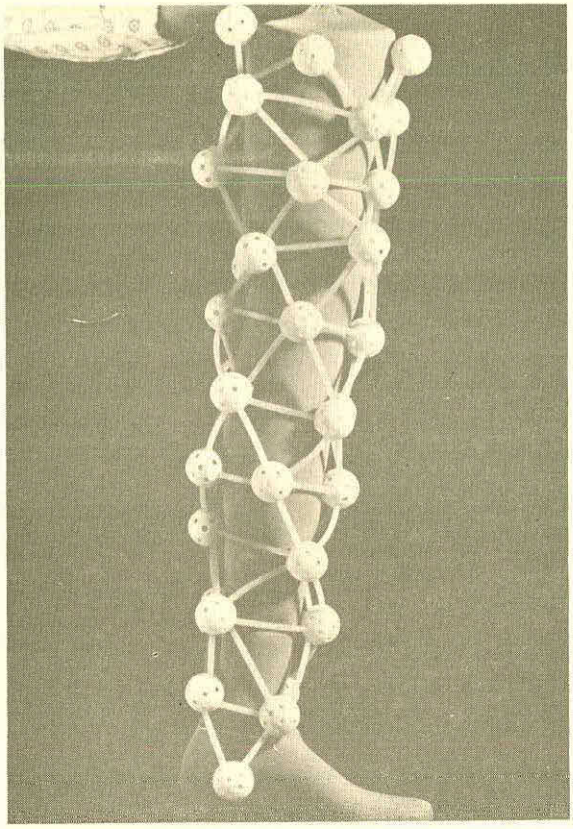

Figure 1: Conceptual model of a two component system to clad the limb.

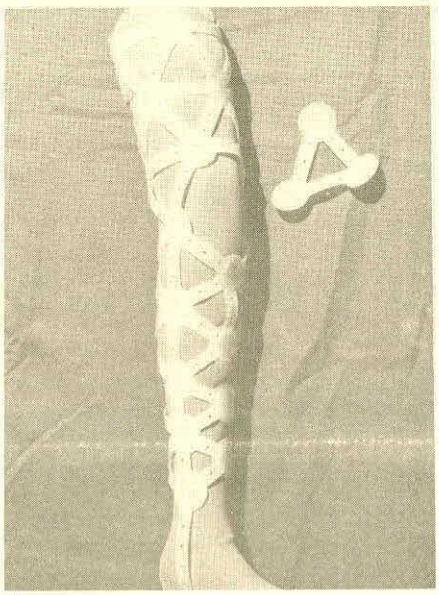

Figure 2: Refinement of the model in figure 1 showing the repeating unit.

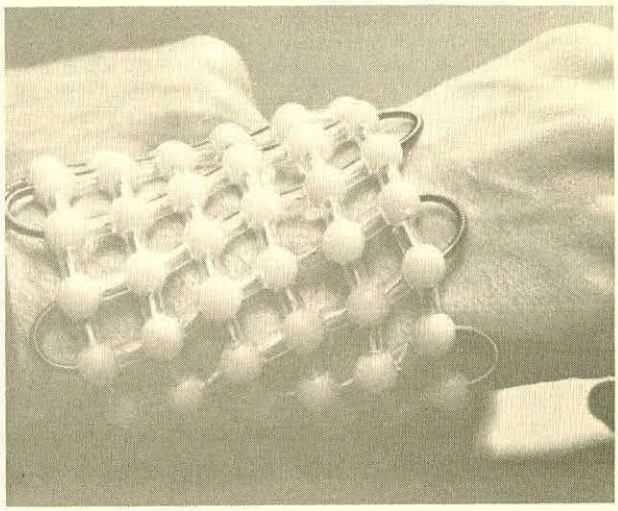

Figure 3: The shapeable matrix on the wrist, a further refinement of the models in figures 1 and 2.

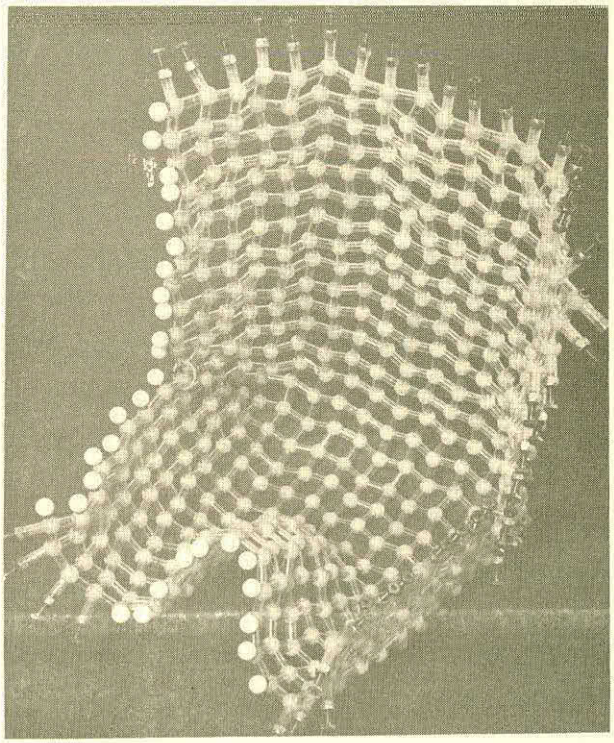

Figure 4: The shapeable matrix as a seat for disabled children minus the thin foam cover.

ADVANTAGES

The advantages of this approach to seating and other orthoses are obvious and numerous.

Elemental orthoses can be produced at low cost since the components can be mass-produced using such techniques as injection moulding or extrusion. The orthotic "elements" (in the case of the seat, a tube and a ball) can be produced in large numbers from the same mould. These orthotic elements can then be assembled to custom fit almost any shape on the body. A custom fit is achieved by standard components.

Secondly, the fitting process of an elemental orthosis is reversible, i.e. adjustments can be made to the device. For example, if muscles waste due to disuse; or swelling subsides; or the load-bearing characteristics need to be changed; addition or removal or re-positioning of the elements can accommodate the changing needs. This type of adjustment can only be achieved with difficulty by plaster or by systems that emulate plaster.

Solutions to other current problems soon become apparent: skin hygiene; heat; weight and inspection continue to be problems with plaster, for example. Elemental orthoses provide opportunity for bathing, air circulation, removal of elements for inspection and would enhance x-ray examination.

RELEVANCE

The importance of this approach to orthotics lies in the cost of delivery of present devices: cost in terms of dollars and time. Long-leg orthoses, spinal orthoses, seats and prostheses are becoming increasingly expensive to obtain and maintain. Considerable handwork and skill is needed to fabricate an orthosis. This fabrication should be left to industry. The orthotist should be free to spend his time in the clinic with the patient.

At present, devices must be made before their usefulness can be estimated. As a result, perhaps 50% of the long-leg orthoses end up sitting in the closet. The resulting cost to the institution is high, but the cost to the patient is often higher. He or she may have to wait six weeks for a long-leg orthosis, perhaps two months for a prosthesis and months !! for a specialized seat. This is unreasonable today, when a muffler can be changed in fifteen minutes and a hamburger produced in fifteen seconds.

With elemental orthotics, it will be possible that the evaluation and fitting process will become one. The device will be tried on a patient at first sight and if it meets her needs, she will be able to go home with the finished device. We now do this with the C.A.R.S.-U.B.C. Knee Orthosis; we can seat some children at first sight within two hours; and we can foresee that we shall be able to do the same with prostheses. So, the elemental approach will enhance service. It will also blur the divisions of prosthetics, orthotics and special devices.

Modern prosthetics employ modules which can be quite complex, e.g. knee modules with swing-phase control. These modules cannot be modified or used in an alternative position on the prosthesis. The Elemental approach, however, employs the simplest repeating component possible, adaptable to many and varied locations. The shapeable matrix, for example, was originally tested on a wrist, found its first clinical application as a seat, and may eventually find application as a femoral fracture orthosis. The application is not specific to the device and as a result, a single system may find many uses, just as plaster of paris does now.

By breaking the shape problem down into the smallest standard elements, the end result, be it a prosthesis or spinal orthosis, is simply a device that provides containment of shape.

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ELECTROGONIOMETRY & SYMMETRY: AN APPROACH TO CLINICAL GAIT ANALYSIS.

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ABSTRACT

The application of objective gait analysis techniques is not widespread in hospitals. Reasons for this include a lack of a suitable equipment and a lack of a suitable format for result presentation in the clinic. The C.A.R.S.-U.B.C. Electrogoniometer is discussed as a viable method of obtaining relative angular data. The interpretation of this data via symmetry of motion is illustrated with examples of hip, knee and ankle pathological gaits.

INTRODUCTION

Gait analysis techniques are not used extensively in hospitals. The reason for this lies in the equipment available and in the biases towards expediency of the clinical practitioners. Responsibility for the equipment and techniques rest with the research and development gait analysis laboratories. The preoccupation of these groups with normals, fundamental studies and constrained clinical studies has resulted in equipment unsuitable for general clinical use. Either the equipment has been tailored to a particular study or there has been an over-pre-occupation with the equipment itself, with little thought given to eventual clinical application.

The application of clinical gait analysis requires the meeting of some severe criteria. Uppermost is that of speed. Speed is required for both the measurement process and for the presentation of the data. In a clinic this usually has to be accomplished in ten or fifteen minutes if the gait analysis is not to impede the clinical process. This severe criterion of speed precludes many elaborate instrumentation techniques and puts a severe strain on the data manipulation and presentation process.

The clinicians using clinical gait analysis techniques require immediacy and are often concerned with the short term results. The eye can provide this and is well-established as a gait evaluation tool. Any new clinical procedure has to perform better than the eye. However, the eye has limitations in that it does not guarantee an accurate permanent record, has difficulty in detecting rapid reversals of movement and has difficulty in keeping track of three-dimensional movements. The criteria and constraints on clinical gait analysis techniques, therefore, can be summarized in terms of low cost, little time, simple technique and compatibility with personnel and patients.

The solution we offer is two-fold. We feel that the most clinically viable method of obtaining gait data is electrogoniometry. The secondary problem of data interpretation can best be met by examination of symmetry of motion of the limbs.

The following will describe the C.A.R.S.-U.B.C. Electrogoniometer (1) and the method of result interpretation based on symmetry of motion.

ELECTROGONIOMETRY

Frequently of interest in clinical gait analysis is the relative movements of the lower limb segments at the hip, knee and ankle joints. The C.A.R.S.-U.B.C. Electrogoniometer, as shown in Figure 1, can measure relative movements in three dimensions at each of these three major joints of the lower limb bilaterally and simultaneously, if desired. The device consists of thigh shank and foot attachments, with measuring modules positioned at each of the joints. The measuring modules consist of three potentiometers mounted orthogonally so that their axes of rotation meet at a common point. These potentiometers give relative angular movements of limb segments in terms of a varying voltage. For example, at the hip is recorded flexion/extension; internal/external rotation of the femur with regards to the pelvis; and adduction/abduction. The potentiometers are connected to the attachment frames through a parallelogram and slider mechanism which absorbs translational movements without altering angulations. Mal-alignment and changing centres of rotation of the joint are compensated for. This translation absorption capability allows the measuring modules to be positioned approximately at the hip (greater trochanter), knee (joint line) and ankle (lateral malleolus). This approximate positioning allows the device to be put on quickly, making it clinically viable. Recordings of three-dimensional motions at the hips, knees and ankles bilaterally can be made in fifteen minutes with practice. Use of moulded plastic parallelograms, frames and attachment straps enables the overall weight of a complete limb attachment to be kept down to 10 ounces (284 grams). This light weight, together with freedom of the joint movement, results in an unencumbering device that has little effect on the gait to

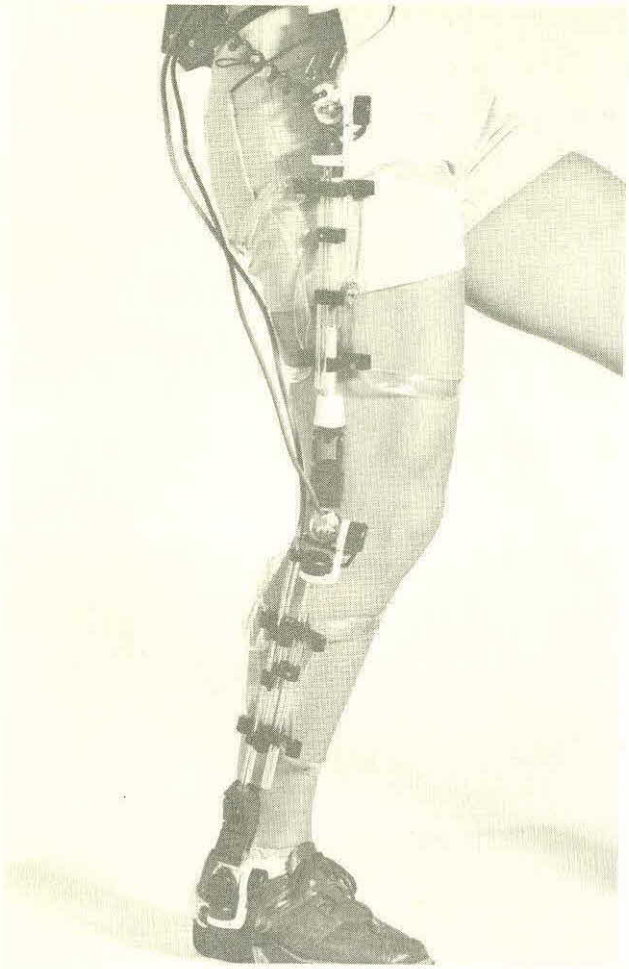
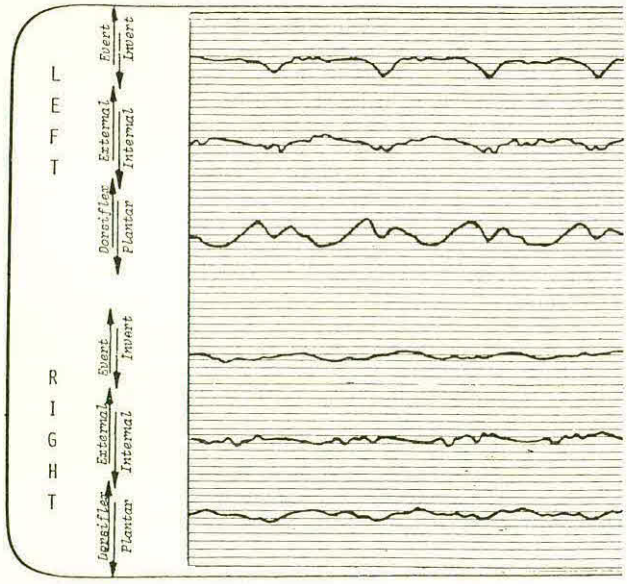


Figure 1: Above. CARS-UBC Electrogoniometer positioned at the hip, knee, and ankle. Measurements are taken bilaterally.



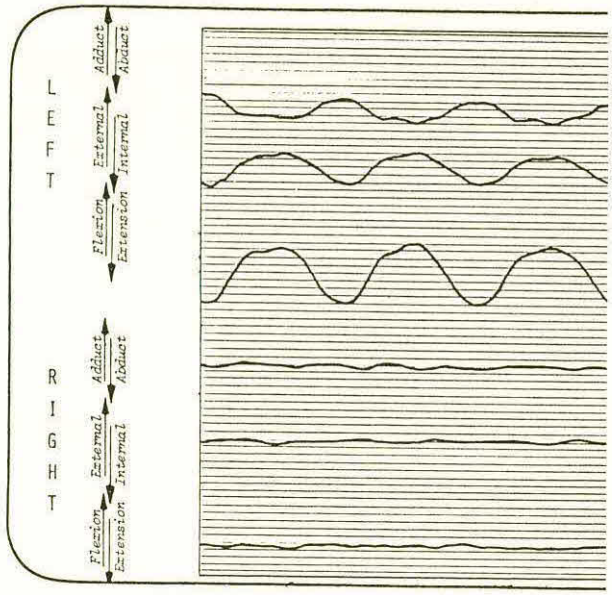


Figure 2: Above. Arthrodesis of the right hip. Demonstrating the stability of the electrogoniometer on the lower limb. Each division equals four degrees.

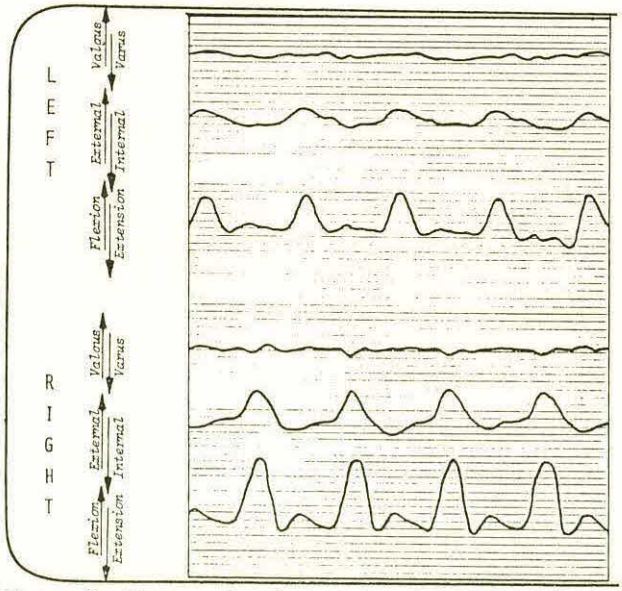


Figure 3: Above. Gunston arthroplasty on left knee; note the lack of symmetry between the left and right limbs.

Figure 4: Opposite. Right Newton ankle arthroplasty; note the limited dynamic motion of the right ankle/foot.

be measured. It is a viable way of obtaining data.

SYMMETRY

Interpretation of gait data is at least as large a problem as obtaining it in the clinic. Gait is complex, many variables are present and clinicians are often not receptive to a complex presentation of the results of a gait analysis. The standard approach of comparison with normals is disrupted due to the large number of variables encountered in the clinic.

These variables may be grouped under six distinct headings: measurement method; parameters; test conditions; population; result format; and economics.

Different measuring methods may give differing values of the same parameters. Or choice of a particular method may result in the loss of some information. The type of information required, e.g. force parameters, relative or absolute angular parameters, may bias the result in a particular way. The test conditions, such as: walking speed, footwear, surface type, etc., vary from locale to locale and even between successive measurements of the same patient. The patient population introduces many variables such as: age, sex, neurological status, etc. There are so many "normal" differences between us that wide variations in gait are to be expected. The format of the results can influence their interpretation. For example, a bias in a statistical method or a pattern recognition technique may lead to false assumptions. The overall constraint of economics will influence all of the previous factors. Not every institution can afford a designated gait laboratory space or armies of personnel.

In summary, a complex picture is presented, with many variables influencing the gait measurement and result interpretation procedure.

A re-examination of the gait of the disabled will reveal that there are two major distinguishing factors from normal gait: a) the disabled usually walks at a slower speed, and b) the disabled person limps, i.e. has an asymmetrical motion. Speed as an indicator of disability has been clarified by several authors.

The usefulness of asymmetry of motion via Electrogoniometry is well demonstrated by Figure 2. This patient had a fusion of her right hip, a hip arthrodesis. As a result, there should be no movement between the femur and the pelvis. As can be seen in Figure 2, there is a small amount of motion recorded, especially in the medio-lateral direction. This evaluation of the stability of the Electrogoniometer on the body, or conversely of the stability of the arthrodesis, is made entirely on the basis of asymmetry of motion of the two hips. No reference to what is normal under these particular test conditions is needed.

Surgery to the knee can also greatly influence gait. The patient shown (see Figure 3) has a Gunston Polycentric knee prosthesis for the left knee. The differences between the left and right knee are immediately apparent. The left knee has reduced flexion during both swing and stance phase, compared to the right. To compensate for this loss of motion, there is an increased rotational movement of the contralateral (right) knee. Significant statements on the performance of different designs of artificial knee joints have been made using this analytical approach.(2)

Currently under study is the performance of the Newton Ankle prosthesis by the second author. Figure 4 shows a typical gait trace measured at the ankles, with a unilateral loss of all three motions for the right ankle prosthesis. Although the study is still in progress, it is emerging that a patient with an ankle prosthesis uses all passive range when walking. From this we conclude that the fullest range of motion be an essential result of surgery.

Symmetry of motion has been employed in other studies by the group to demonstrate: the influence of alignment changes on amputee gait; the proprioceptive feedback effect of flimsy elastic knee supports; the freedoms and constraints of differing knee orthoses; and the effects of walking aids, shoe modifications and physiotherapy. Many of these findings would have been difficult and laborious to obtain by other analytical methods. Symmetry of motion has fared well as a clinical analytical tool.

The statements are useful clinically and are relevant to that particular patient in the particular circumstances of the measurement process. Little need is apparent to refer to standard ranges, effects of velocity, walking surface, age, etc., as all these factors are peculiar to that particular patient. It is necessary, of course, to have some knowledge of what motion is desirable and what is not, to be aware of the effects of velocity, etc.; but these effects are minimized as they affect both limbs similarly.

The limitations of symmetry as an analysis tool include the presence of equal bilateral disability; and over-riding factors such as pain. A symmetrical gait may be the most desirable in appearance but not at the expense of comfort.

SUMMARY

In summary, an approach to clinical gait analysis is discussed that employs electrogoniometry and symmetry as measurement and analytical tools. The advantages and limitations are defined. These techniques have been employed extensively by the Medical Engineering Resource Unit of the University of British Columbia in an attempt to break the log jam of clinical gait analysis, in the hope that these techniques will bear fruit in research, education and clinical practice.

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BASIC INTERPRETATION OF PEDOBAROGRAMS

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A pedobarogram, or foot-ground contact pressure map, carries information regarding both the structure and action of the foot and the posture of the whole body during standing and walking. Examples of the use of the Pedobarograph to obtain such pressure maps is described, and basic interpretation of the results is made with reference to pedobarograms from normal subjects.

INTRODUCTION

A map of the distribution of pressure between the soles of the feet and a supporting surface, the 'pedobarogram', can reveal specific information about both the structure and action of the foot and postural control of the whole body. The type of observation which can be made vary from the detection of anatomical features such as abnormal bone structure or bearing which relate to overall postural balance, to detailed dynamic patterns of loading which may indicate the action of particular muscular groups during voluntary standing manouvres or walking.

To extend the use of pedobarograms to their limit, ideally a recording device is required which gives a continuous map of the instantaneous pressure distribution with sufficient spatial resolution to delimit the important anatomic details of the foot. To date, many devices monitoring the vertical components of force have been developed. For example, the ridged, inked mat used by Morton (1) gives a single, integrated load distribution for a complete step; techniques employing a number of small force-sensitive pads, either in the form of insole transducers or matrix force-plates, produce a series of force/time records which can be processed off-line to build up a pressure map, (2); and a dynamic technique developed by Arcan and Brull (3) based on optical interference patterns gives displays of instantaneous pressure. Each of these methods is based on a different physical principle and has its own advantage of low cost, accuracy or immediacy, but all suffer from a lack of spatial resolution. Another device, the 'pedobarograph' combines acceptable accuracy of registration with good spatial resolution and instantaneous, immediate pressure maps. This device was designed by Chodera (4) in Czechoslovakia, with a parallel development in the U.S.A. reported briefly by Hertzberg (5), and has been developed more recently at Roehampton (6,7) and other U.K. centres (8). The pedobarograph follows Elftman's barograph (9) and is based on a simple optical principle of total internal reflection.

This paper demonstrates the use of the pedobarograph to obtain foot-pressure maps, and the type of information which can be read from pedobarograms.

TECHNIQUE

The pedobarograph consists of a mainframe structure with associated video-electronics. The mainframe comprises a top glass plate lit into two opposing edges by fluorescent strip lights, and a supporting structure. The subject stands on top of the plate with a foil, which may be a thin sheet of textured PVC, polyurethane foam or similar, between the foot and plate. Where the soles press the lower surface of the foil into close contact with the plate, light normally conducted across the plate by total internal reflection is bounced from the foil downwards, and on a macro scale the intensity of light escaping from the lower surface of the plate is proportional to the local pressure applied to the foil. Pedobarograms can be viewed directly beneath the top plate of the mainframe as light intensity-modulated pictures, with a continuous conversion of pressure-to-greyscale. Alternatively, the direct picture can be picked up on a video camera, fed through a colour analyser to convert the image into a zonal map between preset pressure levels, and finally displayed as a colourised zonal map on a video monitor. A video tape recorder can be used to capture pedobarograms during stance or gait for later inspection.

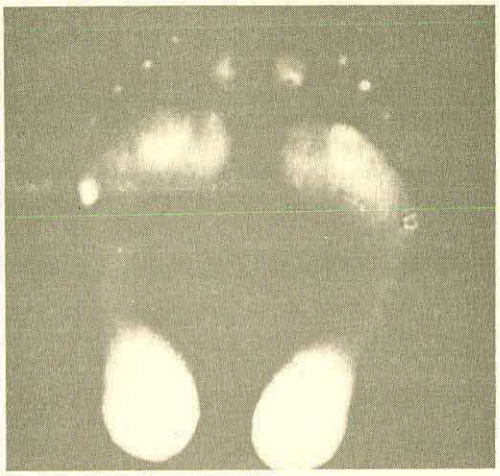

Figure 1. Direct pedobarogram.

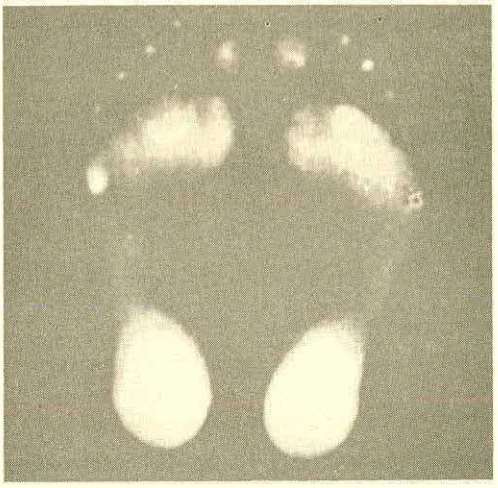

Figure 2. Zonal pedobarogram.

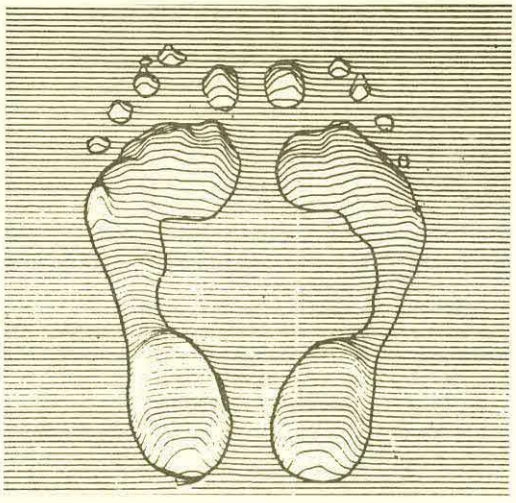

Figure 3. Quasi three-dimensional pedobarogram.

EXAMINATION OF THE STANDING SUBJECT

The distribution of the pressure in soles during standing is dependent on the position of the feet and posture. It can be changed by volition easily. It is therefore important to standardise the posture, position of the feet and exclude deliberate interference from the subject. We ask the person to do a small exercise (say stretch the arms forward 3-4 times with closed eyes) and then fix his eyes on a target 30^{o} below eye-level. We examine the person in three basic positions. In the first, the subject stands naturally. Next standing with toes close together and heels apart emphasises the pressure on the lateral side of the soles - the foot goes into inversion. The third position is with heels together and toes apart with 60^{o} included angle. The foot is now in eversion and details of changes are emphasised on the medial edge of the sole. Two manoeuvres are employed; while standing with feet close together the patient is asked to depress the toes firmly. This will in the normal foot lower (or remove) the pressure under the 1st metatarsal head and raise the longitudinal arch. The pressure in the midfoot should diminish. In the second manoeuvre we ask the patient to raise the toes whilst keeping the rest of the sole on the plate; the pressure increases under 1st metatarsal heads and the lift of the longitudinal arch should be well expressed. Both manoeuvres cause a complex muscular reaction and display early changes in clinical conditions like rigidity of the big toe, lack of muscular coordination in control of foot arches, rigidity or hypermobility of the small tarsal joints etc.

EXAMPLES OF PRACTICAL PEDOBAROGRAMS

Figures 1 to 3 show examples of different display techniques of pedobarograms. All three are derived from the same footprint of a man aged 46 with chronic sciatica L4-S_1 of 13 years duration. He has his feet close together with toes depressed.

Figure 1 is a direct picture in black and white as the television camera would see from underneath the glass plate of the instrument.

Figure 2 converts the grey scale of the previous figure into 7 zones of grey (or 7 contrasting colours), limiting on the one hand the detailed information but giving a better impression of the overall distribution of pressures.

Figure 3 takes each fourth line of the 625 PAL standard television picture and displays it as an amplitude analogue curve rather than the standard intensity modulated line. It produces a three-dimensional spatial impression of the pressure distribution and offers a lot of detail for clinical purposes.

Note the high pressure under the 5th metatarsals; on the right there is a patch of hardened skin - a corn. It is possible to compare the three display modalities with reference to these details under the metatarsal heads.

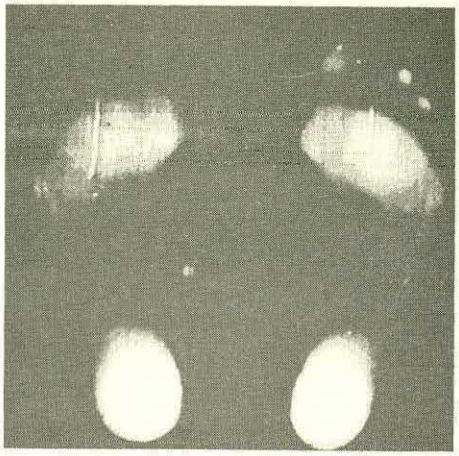

Figure 4. Zonal pedobarogram. Natural stance.

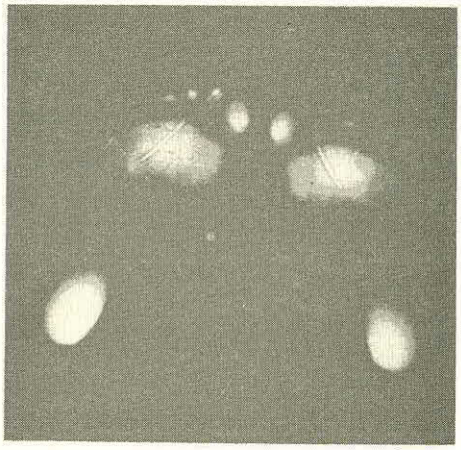

Figure 5. Zonal pedobarogram. Convergent stance.

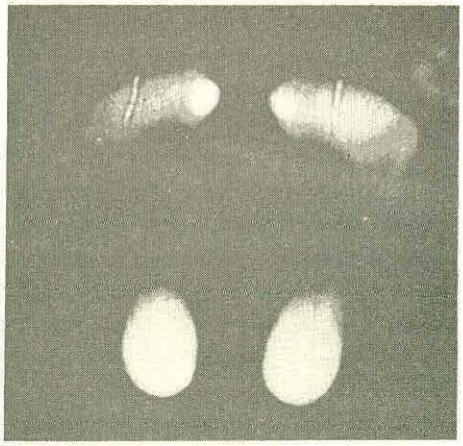

Figure 6. Zonal pedobarogram. Feet together, toes raised.

The plateau under the first metatarsal head on the right foot displays the relief of pressure normally achieved by depression of the toes. This fails on the left foot. The expected lift of midfoot is well expressed on the right, but the pressure under the base of the 5th metatarsal confirms the lowering of the proximal transversal arch.

Figures 4 to 6 demonstrate the PBG manoeuvres, with zonal pedobarograms. Figure 4 shows an indifferent position of the feet in a 35 year old woman with relatively normal pressure distribution. She has occasional pain in the back since a delivery five years ago. Figure 5 shows the convergent position and Figure 6 a well expressed outcome of raising the toes.

FURTHER APPLICATIONS OF PEDOBAROGRAPHY

Pedobarography has many uses in addition to the examination of barefoot standing man just described. The pedobarogram in shoes reveals the resulting shoe-to-floor pressure distribution with surpringly many details of parts of foot in action (big toe, metatarsal heads etc). Hence the technique can be used to check on redistribution of pressure by insoles, to quantify the corrective pressure, position and function of rockers etc. The influence of orthotic bracing is reflected in a changed pressure distribution, and its action can be evaluated. With some additional processing to add a centre-of-foot-pressure marker on the television display the technique can be used in rehabilitation of posture in children and for instance restoration of balance in sport injuries and following cerebral accidents. It has been proved to be useful in assessment of the function of the foot-shoe complex for the shoe manufacturing industry. The PBG has also special value in surgical shoemaking, assessment of prosthetic replacements of the lower extremity, and assessment of the function outcome of orthopaedic surgery.

CONCLUSION

Pedobarograms can be processed in a variety of ways to yield both scientific and clinical information. Examples of interpretation of normal pedobarograms have been given. These are in no way to be considered a comprehensive guide to interpretation, but merely demonstrate the type of features which the critical eye may wish to investigate further.

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AN INSTRUMENTED GAIT MAT

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ABSTRACT

An instrumented mat is described which uses electronic methods to obtain measures of step length, cadence, velocity, and swing/stance times during human locomotion. The device requires no attachments to the subject, is easy to use in the clinical setting, and provides a printed record of the measured gait parameters.

INTRODUCTION

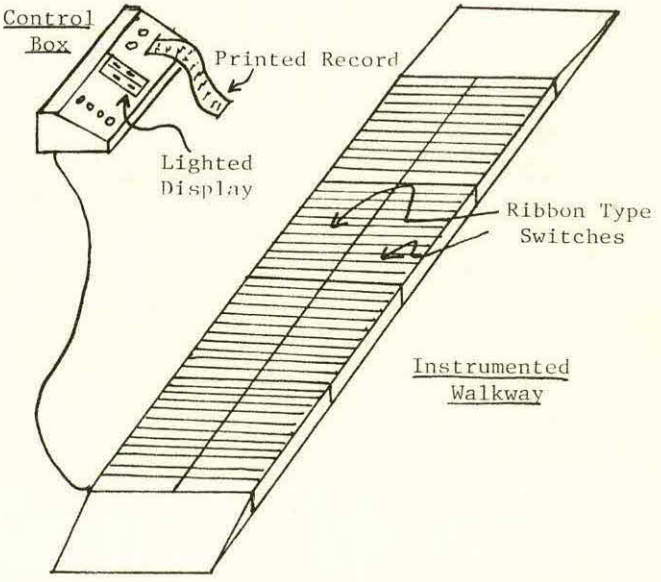
The most frequently used objective measures for gait evaluation are step length, cadence, velocity, and swing-stance times (1,2). The simplest method for obtaining step length and average velocity is to get an imprint of the subject's foot as he walks over a measured distance while observing with a stop watch (3,4,5). Recording methods have included inked shoes and paper, chalked shoes and a black rubber substrate, or oiled feet and paper. The subsequent measurements made from the record are awkward and time consuming. Cinematographic and time lapse photographic techniques also require significant data reduction time and similarly entail a delay in obtaining the desired measures. More recently developed instrumentation (6,7,8), including that described here, uses a walking surface instrumented with electrical contacts or (9) uses foot switches and photocells to obtain the data.

DESCRIPTION

The instrument described here employs a microprocessor which scans a series of switches embedded in the walkway substrate and provides a printed record of both the spatial and temporal events as they occurred during a subject's traverse over the mat. Right and left sides are measured separately and the data can be examined and edited before computation. Temporal and spatial resolution is 20 milliseconds and 15 millimeters with a force threshold of five pounds.

The instrument consists of four right and four left walkway sections, each about a meter long, which may be placed end to end to form an instrumented path of about four meters in length. The walking substrate consists of a series of contiguous pressure switches which are sequentially scanned by the electronics as the subject progresses down the walkway. A microprocessor, with associated memory circuits, records the time and location of each switch closure and release, and performs subsequent calculations on the recorded data. A lighted display shows the location of each right and each left footfall as it was observed by the instrument so that the data may be examined and edited before calculation.

The walkway with all four sections consists of 512 sections of twelve inch ribbon type switches, arranged with 256 on the right and 256 on the left side, oriented transversely to the direction of the subject's progression. The switches are sequentially scanned by a multiplexor in order to minimize the number of wires required to connect the mat with the control box. Sixteen decoders on each side handle 16 switches each so that only a fourteen conductor flat cable is required for interconnection (see Fig. 1).



The control box contains digital circuitry and a microprocessor to generate the multiplexor coding, to store switch closure and release times, to operate the lighted footfall display, and to perform the computations and printing of the results. Four control buttons allow the operator to begin the acquisition of data, to terminate it, to print the data, and to reset and clear the device in preparation for another pass by the subject.

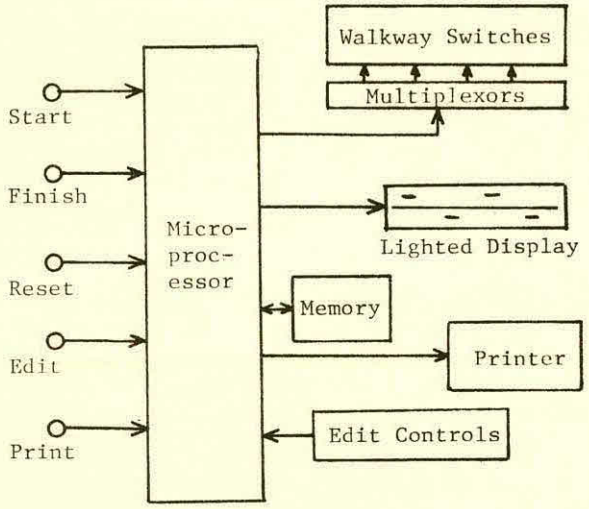
The start button begins the scanning operation and starts an internal clock. As each closure under a footfall is detected, the time of

closure is stored in memory at the location corresponding to the position of the switch on the mat itself. Another portion of memory stores the time of release for each of the 512 switches. The memory capacity required is 2048 bytes at 8 bits per byte for storage of the closure and release times as four decimal digits each.

The acquisition of data is terminated by the "stop" button, which renders the device insensitive to further switch activity, and the stored footfall pattern is displayed with right and left sides separated on two scanned linear LED arrays. Partial steps on crossovers may be deleted, and if the acquired data appears satisfactory, a push on the "Print" button initiates calculation of average step lengths, velocities, and swing/stance times. These are printed on a strip chart along with the raw data arranged on the paper by distance.

A Reset-Clear button erases all data and resets the device in preparation for a new traverse along the walkway.

A block diagram of the control electronics is shown in Fig. 2.



RESULTS

The completed apparatus has been in use by clinical personnel in order to obtain an initial evaluation. The device is easy to use and has encouraged the proposal of studies for evaluation of physical and pharmacological therapeutic treatment of patients with neuromuscular deficits. Preliminary studies to evaluate the effects of cortisone injections on arthritic patients and to evaluate the effect on gait of orthotic devices have indicated that the equipment is easily manageable by clinical personnel and that significant data can be obtained in only two to three minutes per patient.

DISCUSSION

For clinical personnel who have used the ink and paper or the cinematographic methods for step length and cadence measurements, the electronic device described here has meant an enormous economy in obtaining and reducing data for these gait parameters. Experience so far has indicated the utility of the editing feature, which allows deletion of crossovers and partial steps prior to calculation and printing of data, and the longitudinal division, right and left, for subjects with very short step lengths. Some redesign of the walking substrate is indicated in order to make it more adaptable to use in parallel bars of the type found in the usual physical therapy gym.

It is hoped that the ability to easily and quickly obtain these rather basic measures of locomotion performance will encourage the use of objective measures in the assessment of patient function and the effects of prescriptive treatment.

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GAIT ANALYSIS OF AN AMPUTEE WITH MINIATURE TRIAXIAL SHOE-BORNE LOAD CELLS

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ABSTRACT:

Triaxial shoe-borne load cells are being used to analyze the gait of a BK amputee. The gait is studied from first step in a temporary pylon to a final "normal" gait in a permanent prosthesis. The distribution of the loads under the various parts of the foot are presented. The amputee's gait is also compared to that of normal.

The load cells are an effective method of analyzing an amputee's gait. It is possible to identify misalignments from the output of the various load cells. This method may be useful for identifying and correcting problems in the painful BK stump.

INTRODUCTION:

Normal and abnormal human gait has been studied extensively over a considerable period. The vast majority of this work has been carried out on force platforms or walkways. Although these devices have led to a much better understanding of the gait cycle, yet they have made little impact on clinical evaluation of individual patient problems.

In order to overcome this problem Harrington et al (1) used an instrumented shoe. However, these devices used failed to breakdown the force distribution under the foot in sufficient detail. More recently Ranu et al (2) developed a system which is capable of meeting the needs of a below knee (BK) amputee.

It should be realized that the rehabilitation of a BK amputee is no longer seen as a surgical exercise plus the construction of a limb replacement. But more of a complete knowledge of human locomotion of an amputee, including rehabilitation and social re-integration within the society. In order to achieve these aims this paper discusses the development and use of miniature shoe-borne load cells in gait analysis for a BK amputee and comparing this with a normal subject.

DESCRIPTION OF LOAD CELL:

A miniature triaxial load cell 14 mm thick x 19 x 19 mm in overall size has been developed (Figure 1). It consists of 4 loops of load bearing elements placed symmetrically around a 12.5 mm square which is located at the center of the load cell. There is no contact between the top and bottom square plates to allow load transfer through these loops. A total of 16 small foil-type strain gages are used to achieve triaxial force output with minimal cross-talk.

Prior to installation in the shoe, each load cell was calibrated statically along each of its 3 sensitive axes. An orthogonal coordinate system was used in order to identify the direction of the applied force. During loading all three channels were monitored in order to check for cross-talk.

The load cells were attached to the shoe by means of a rigid backing plate. They were mounted directly on the heel and under the 1st and 5th metatarsals and the toe. The placement of the 2 load cells under the heel is not critical. The location of the 3 load cells under the rest of the foot require a subjective input from the test subject who was asked to stand and give directions as to how and where the transducers should be moved for maximum support and comfort. Figure 2 shows such a set up. For an amputee, it was not possible to carry out this subjective procedure and the load cells were placed approximately under the metatarsals and the toe.

DATA ACQUISITION:

The 15 channels of force data were recorded on an analog tape recorded via lightweight flat trailing cables which were suspended on a series of rollers in a low-friction track. A mini-computer was used to provide on the spot data digitization and processing for 10 of the 15 channels. Scaled plots of force versus time could be viewed immediately after a test to ensure that the critical channels have come through and to compare results with previous tests. The data could be displayed on a screen or copied onto a hard copy for a permanent record.

RESULTS:

Ranu et al (3) have shown that for a normal subject the data from the triaxial shoe-borne load

cells correlated very well with the walkway force plate.

Figure 3 shows the total vertical force-time history for two runs for an amputee. The time difference between the two runs is approximately one month. The figure represents a quantitative measure of adaptation of the amputee to the prosthesis. Although the general pattern of the force-time history is some what similar, the initial run shows that the amputee requires a much longer time to support the body weight through the stump leg. Also peak force for the inital run is less, when compared with the later run. The reason could be pain in the stump or simply a matter of confidence in the prothesis. Note that Figure 4 shows the complete stance and swing phase for the right (artifical leg) and left foot of an amputee.

Figures 5-7 show the force-time histories under different regions of foot of an amputee for an initial and later run. 'HL' denotes heel force at lateral position, and 'M1', is force under the first metatarsal and 'T' is the force at toe of foot. From these graphs it is recognized that there is a significant change in force-time history under different regions of foot within the two time periods.

Figure 8 shows the movement of the center of foot pressure in forward direction against time for both the amputee and a normal subject. For the amputee the center of pressure moved abruptly in the forward direction. This is a quantitative description of vaulting which is not present in the normal subject. Also note that the time taken between heel strike and toe off is longer for the amputee. However, the initial time taken for heel strike for both cases is the same.

Figure 9 shows the paths of center of foot pressure for a normal and an amputee subject. There was excessive pronation of the amputee's foot at heel strike as evidence by the proximity of the center of pressure to the medial aspect of foot. This can be contrasted with data from the normal subject who had a slight amount of tibial varus, resulting in the lateral location of the center of pressure. Center of pressure data for the normal subject compared favorably with those of Grundy et al (3).

CONCLUSIONS:

Conclusions drawn from this investigation are:

(1) The transducer is a viable tool to assist in the rehabilitation of an amputee and other forms of pathological gait.

(2) The transducer can detect subtler differences between normal and abnormal gait.

(3) The transducer can help speed up the training of an amputee in the use of a prosthesis. It is also possible to aid him in the development of a more normal gait.

(4) The path of the center of foot pressure can act as a guide in correcting prosthesis misalignments.

(5) Rapid data acquisition and processing are required for maximum effectiveness.

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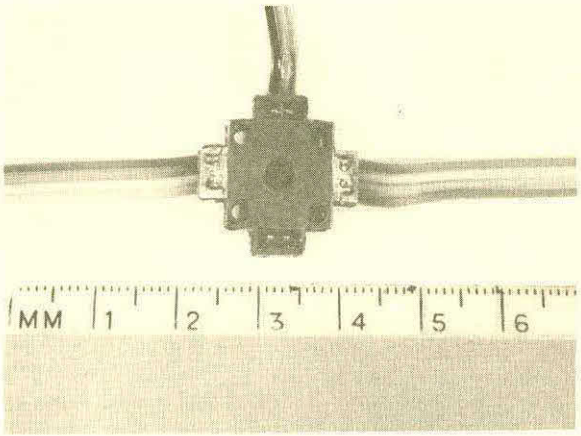

Figure 1: Top View of Miniature Triaxial Shoe Load Cell.

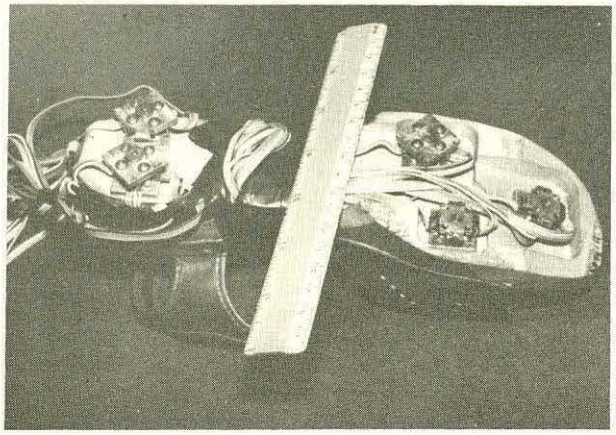

Figure 2: Five Load Cells Mounted Under a Shoe.

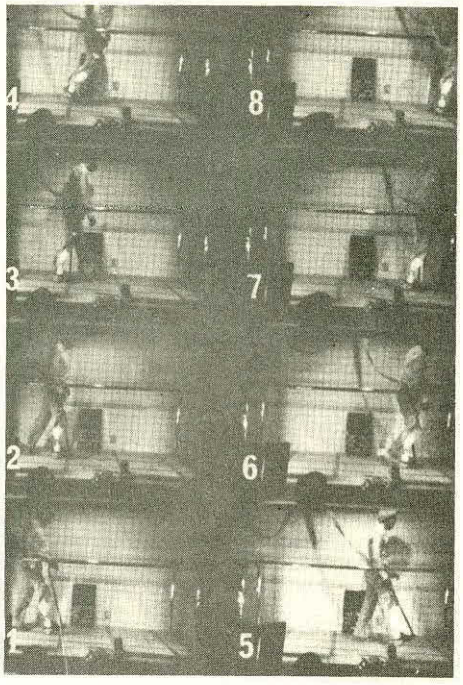

Figure 4: Stance and Swing Phases for an Amputee.

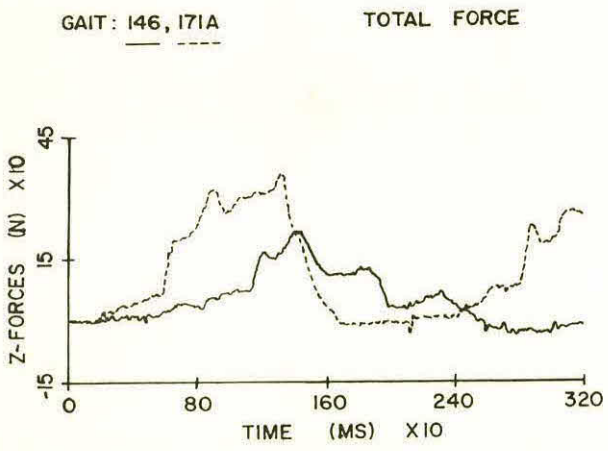

Figure 3: Total Force-Time Histories for an Amputee (Z-Axis). (——First Run) (---Second Run).

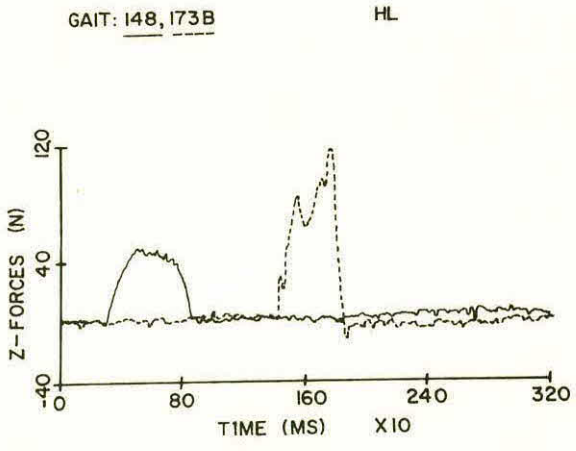

Figure 5: Force-Time Histories Under Heel for Load Cell at Lateral Position. (Z-Axis). (——First Run) (---Second Run).

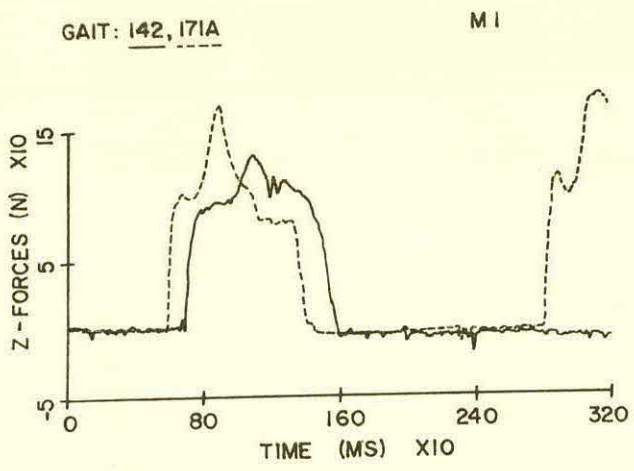

Figure 6: Force-Time Histories of Load Cell under the First Metatarsal. (Z-Axis). (——First Run) (---Second Run).

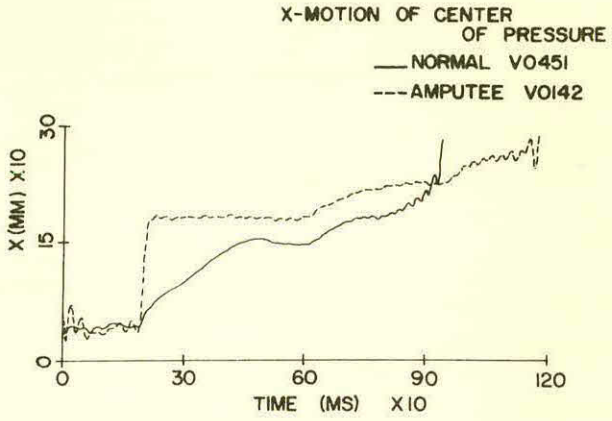

Figure 8: Path of Center of Foot Pressure-Time in Forward Direction for an Amputee and Normal Subject.

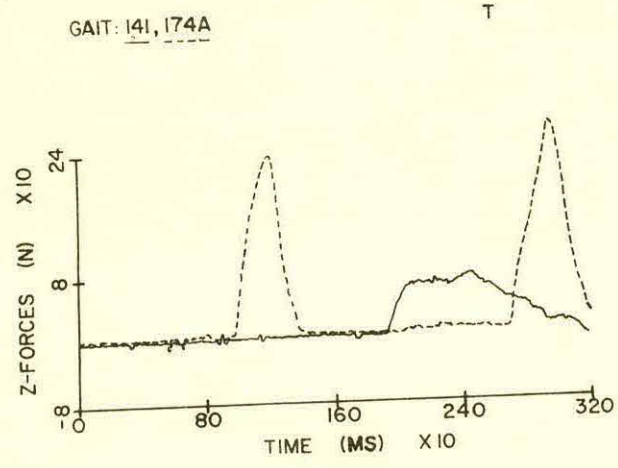

Figure 7: Force-Time Histories of Load Cell under the Big Toe. (Z-Axis). (——First Run) (---Second Run).

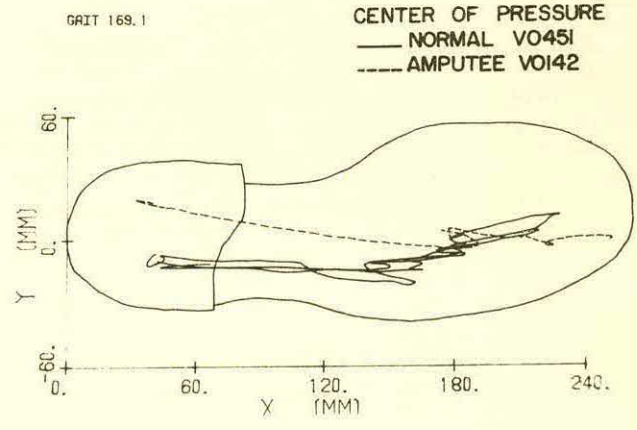

Figure 9: Path of Center of Foot Pressure in Y and X-Axes for an Amputee and Normal Subject.

GAIT ANALYSIS IN REHABILITATION MEDICINE

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ABSTRACT

This paper argues the need for gait analysis techniques which can be used to assess disability in the Rehabilitation Clinic. A technique which has been used in this context is described. The technique measures the temporal and distance variables of gait objectively. The paper also describes the ways in which these objective measurements of gait are used to assess disability in hemiplegics and in arthritic patients before and after total hip replacement. The importance of data presentation is emphasized.

INTRODUCTION

Walking is a complex series of periodic motions and as such is extremely difficult to analyze objectively especially if the analysis is to be regarded as complete. As with most complex movements, in order to analyze the events that are occuring the total movement has to be broken down into simpler patterns. For example one might look at gait in terms of the duration of contact of the feet with the ground. The measurement of these contact times allows at least part of the gait pattern to be expressed numerically and therefore objectively. If one were to attempt a more complete analysis one would need to measure the following:

1. Kinematic Parameters
   a. Temporal factors
   b. Distance factors
   c. Angular displacements
2. Kinetic Parameters
3. Muscular activity
4. Metabolic Cost

The measurement of all these parameters simultaneously would be ideal for research but would provide an unwieldly amount of data for clinical purposes. When gait analysis is undertaken therefore, it is usual for only one or two of the above variables to be measured. Although there are laboratories which are capable of undertaking much more thorough analyses, these laboratories, because of the complexity and cost of the equipment required are, by and large, research institutes. The need for some system for gait analysis in the Rehabilitation Clinic is however self-evident. If one is administrating some form of therapy to a patient in order to improve his/her ability to walk then it is important to monitor the progress made, if any. Most commonly this is achieved by means of subjective assessment techniques. The reasons for this might be that the eye of an experienced clinician most closely approaches the following criteria for an 'ideal' gait analysis system for clinical use:

1. Measures all gait variables
2. Is of no encumbrance to the patient
3. Simple to use
4. Fast
5. Accurate
6. Presents results
   a. immediately
   b. in a readily understood manner
   c. in a form which can be stored with the patient's file
7. Zero cost

However, although the clinician may get an impression of most of the gait parameters the problem arises when it comes to communicating

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those impressions. This is normally done by describing the gait in terms such as "walked with a pronounced limp" which mean different things to different individuals. Hence the need for objective measurement. To build a system that meets all the above criteria is an impossibility. How for instance can criteria 1 and 7 be met? There must obviously be a compromise. One compromise solution which appears to have the potential for use in Rehabilitation Clinic is the subject of this paper.

METHOD

Perhaps the most commonly measured gait parameters are those that relate to the temporal and distance aspects. Techniques to do this range from the use of a stopwatch to time a patient over a known distance to complex opto-electronic devices. The temporal and distance parameters of gait are extremely useful in assessing gait pathologies and since they are reasonably easy to measure they would appear to be ideal for the purposes of a Rehabilitation Clinic.

The gait analysis system used has been described in detail earlier[1]. The basis of the system is a walkway linked to transducers which measures the duration and position of each foot/floor contact. This is achieved with the minimum of encumbrance to the patient. In fact all that is required is a strip of self-adhesive, electrically conducting tape attached to the patient's own shoes. The walkway can also be used to determine the duration and position of contact of any walking aids used. The temporal/distance factors measured are both single and double support times, swing time, stride time and hence cadence, step length and stride length.

The results sheet was designed to accompany the patient's notes in the same manner as the results from other laboratory procedures, such as E.C.G. Consequently only the mean values of the temporal/distance factors are presented. The mean values are obtained from the data for each step of each traverse of the walkway. To make the results sheet more readily acceptable the data are presented in both numerical and graphical forms. The time factors graph simply indicates the time of contact of the feet and aids. The second graph shows the relative positions of heel strike of the two feet and contact points of any aids used. Between them the two graphs, which are drawn using the mean values, show diagrammatically the temporal and spatial symmetry of the gait. Double braking support time is defined as the interval between the initial contact of the foot under consideration and final contact of the contralateral foot. This is equivalent to the double thrust support time of the contralateral limb. Relative velocity is the ratio of velocity to stature. This ratio was introduced to take account of differences in height which are known to affect walking speed[2]. Temporal factors are expressed as percentages of stride time and likewise step length is given in absolute terms and as a percentage of stride length.

The walkway has been used to monitor the walking ability of arthritic patients pre and post total hip replacement and also a group of hemiplegic subjects.

DISCUSSION

In both pathologies considered one comes up against the problems of which parameters are best suited for assessment of disability and with what should the results be compared to.

In this study the patients performances have been compared with their own previous or subsequent performance, i.e. the subject is his/her own control. There are two other comparisons which should be made, firstly to a normal population and secondly to a population with a similar pathology.

However, it is difficult to select the measurements of gait which should be used as the basis for comparison. Even the relatively simple measurements of the temporal and distance factors of gait provide a burdensome amount of data. It is clearly best to use the minimum number of variables and to record only those which most accurately reflect the degree of disability. The variables chosen to reflect one pathology such as the osteo-arthritic hip may not be suitable for another type of disability such as hemiplegia.

From a consideration of the results obtained in this study, on the total hip replacement patients and on the hemiplegics, it would appear that there are two aspects of gait which should be considered. The first is the degree of symmetry of the walking pattern. In the normal condition the degree of asymmetry both temporal and spatial is only slight. When the degree of asymmetry increases the walking pattern becomes noticeably abnormal and the patient is perceived to have a limp. Secondly, consideration must be given to the absolute values of the gait such as stride time, stride length and velocity. A patient may walk with a perfectly symmetrical gait but this may be abnormal because it is so slow. Conversely a patient may be capable of walking at normal or fast speeds but with an abnormal degree of asymmetry.

Since the speed at which a person walks is one of the most commonly measured gait parameters it was chosen as one of the absolute values. An increase in velocity can be achieved by decreasing the stride time or increasing the stride length and is normally achieved by a combination of both. To determine the relative contributions of these parameters to any change of velocity both were included.

Step length as a percentage of stride length

was selected as the measure of spacial symmetry. In a normal pattern this value would be 50% since right and left step lengths are equal. To reflect temporal symmetry the values of the contact times expressed as percentages of stride time were selected.

The results from the study on hemiplegics show that in terms of the absolute values chosen there is a marked difference from normal[3]. There is also a lack of symmetry which is indicative of a favouring of the affected limb. This lack of gait symmetry and a tendency to reduce weight bearing on the affected side are two problems common to stroke patients. This supports the theory that therapy should be directed to improving such factors[4].

Group results for the patients who had unilateral hip replacements indicate that the step length of the affected side is slightly longer than normal before operation. Six months after total hip replacement the step lengths for both sides are more equal and by twelve months normal. The same pattern was seen for the support time of the affected limb. When the support time for the normal leg was considered it was seen that before operation the foot is in contact for an even longer percentage of the stride than the affected side. Near normal values were obtained by six months with little change in the ensuing six months. The temporal asymmetry before operation illustrates the tendency for the patient with a unilateral hip disorder to favour the arthritic hip by bearing weight through it for a shorter period of time. The shorter step length on the normal side probably results from this, since it is during the single support phase on the affected side that the normal leg is being swung through and placed ahead of the other foot. Reducing the swing time could result in a reduced step length and this may explain the spacial asymmetry which has been noted.

With regard to velocity by six months and even more noticeably at twelve months the males show a marked improvement. In both males and females the improved velocity is achieved by increasing stride length and decreasing stride time. Before operation the velocity of males and females was similar but at six months after total hip replacement the males displayed a higher velocity than the females. This trend is normal and reflects the longer stride-lengths used by males.

From the clinician's viewpoint this system of gait measurement produces a record of the patient's walking in a natural manner, unencumbered by apparatus and in their normal shoes. The data provides both a graphic record and an absolute measure of specific components of the gait cycle. It can be applied to any variety of pre and postoperative assessments of procedures affecting the lower limbs.

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WATERLOO BIOMECH - PART I, A COMPLETE COMPUTER PACKAGE FOR THE ANALYSIS OF PATHOLOGICAL MOVEMENT

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Abstract

The purpose of this paper is to describe a computer package, WATERLOO BIOMECH, which has been developed for the analysis of the biomechanics of human movement. These programs have wide flexibility and are applicable to the monitoring, analysis and diagnosis of pathological gait. The features of the package allows for a modelling of the complete body, and permits kinematic, kinetic, energy, power and momentum analyses. Example plots demonstrate these capabilities in the analysis of normal gait.

INTRODUCTION

In its assessment of human movement the Department of Kinesiology at the University of Waterloo has developed over the past 7 years a generalized package of computer programs for the assessment of human movement. The programs have equal applicability in the assessment of normal, athletic and pathological movement: all that differs in each case is source of the data, and the types of assessment questions that the diagnostician may wish to ask. The purpose of this paper is to document the flexibility of the models that can be analysed and the many options available for diagnostic assessment.

DATA BASE

All biomechanics programs require kinematic, kinetic and anthropometric data. For a complete analysis kinematic data must therefore be in an absolute spatial reference plane and the kinetic data must be synchronized in time and space. Therefore the kinematic programs require anatomical coordinates to be available from cine, TV or other optoelectric imaging systems. The essential characteristics of the data base are that the anatomical coordinates can be collected as follows:

1. The camera system can be fixed or mobile (on a tracking cart).
2. The patient can be fixed (on a treadmill) or mobile (on a walkway).
3. Any number of strides may be analysed.
4. Any number of limbs can be modelled (arms, legs, ipsilateral and/or contralateral).
5. Gait can be symetrical or assymetrical.
6. Patient can be walking in either direction (to the left or right).
7. Simultaneous force plate data for both feet can be utilized.

OVERALL ANALYSIS CAPABILITIES

The evolution of this flexible program enables the researcher and clinician to select from a wide variety of options depending on his data base and requirements. An overall block diagram of the system is seen in Figure 1.

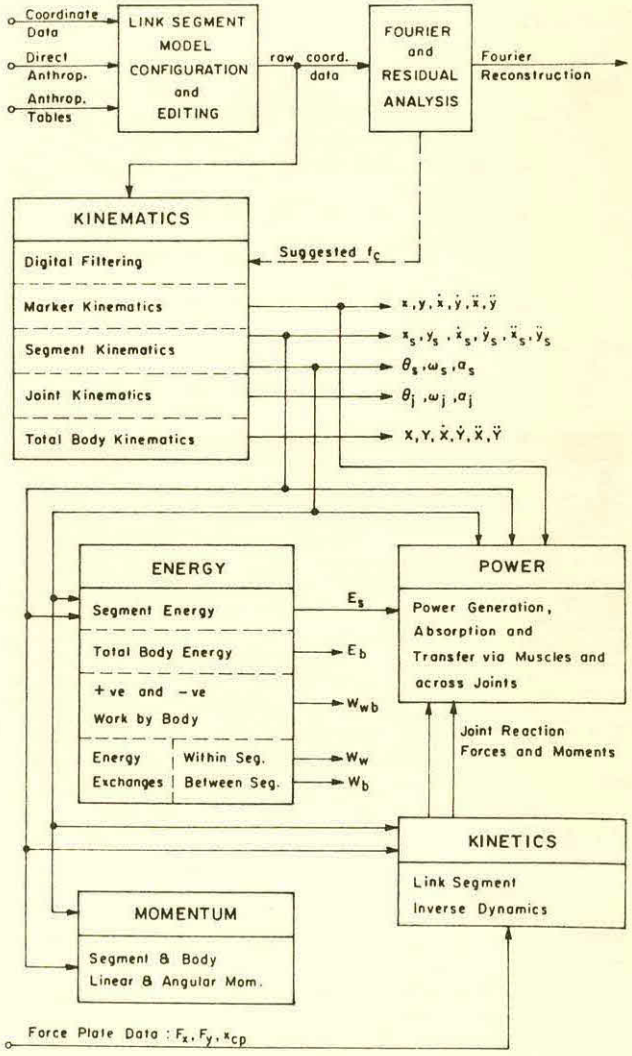

Figure 1

Model Configuration and Editing

The first stage of any biomechanical assessment requires a generation of the link segment model in terms of the anatomical position of the markers in conjunction with anthropometric measures. The researcher can use anthropometric tables (based on height and weight of patient) or may amend these values to account, for example, for direct measures taken from a prosthetic limb. The modelling can be done in advance of inputting the coordinate data from the body markers. Editing features alert the investigator of large errors in marker coordinates (due to human error in film digitization, excessing skin movement, momentary loss of a marker) and permit him to make corrections. Also, there is an option where the investigator may correct for image distortion due to the lense system or because the camera axis was not exactly at right angles to the plane of movement of the walkway. Such a correction requires a calibration matrix to be taken before or immediately after the trial. Figure 2 shows such a system that is used regularly in our gait assessments.

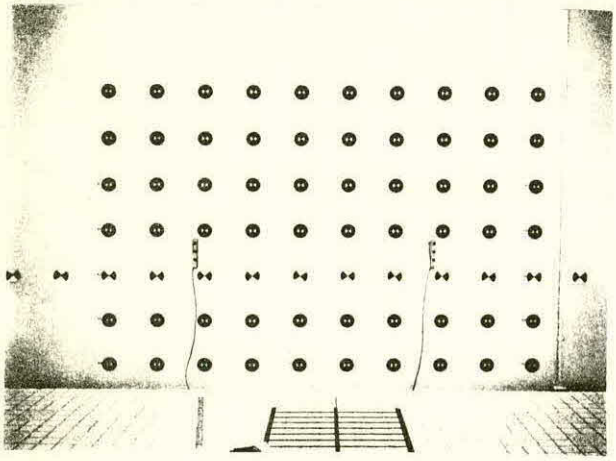

Figure 2

Kinematics

The raw edited coordinate data can now be fed to the main kinematics program which calculates the scores of kinematic variables that may be required to describe or further analyse the movement.

The first stage of processing the data is to filter the data prior to calculation of velocities and accelerations. The technique employed is digital filtering, a zero-lag low pass filter is employed (1) which has been validated (2) and found to be superior to other data smoothing techniques. The only decision the researcher must make is the cut-off frequency of the filter. Our experience over the past 7 years has shown that 4 Hz is adequate for a slow walk and 6 Hz for a fast walk. However, if there is some concern about the right cut-off frequency the investigator has the option of carrying out a Fourier analysis of the raw coordinate data, and by a method involving residual analysis a suggested cut-off frequency for each individual marker is suggested.

The filtered data now forms the basis for calculation of all kinematic variables. In the absolute reference plane the marker velocities and accelerations are calculated. Each segment has nine variables to define its movement in two dimensions: displacement, velocity and acceleration of its centre of gravity along with its angular position, velocity and acceleration. The joint kinematics (angle, angular velocity and acceleration) are now calculated. Finally, the weighted kinematics of each body segment yields the total body centre of mass position, velocity and acceleration. Any of these descriptive variables can be plotted using a flexible plotting program or be used as inputs to more advanced analysis programs, all of which are options.

Energy

Energy analyses have been found to be extremely valuable in the diagnosis of human movement. They are powerful in showing the total body energy changes over the gait cycle from which the net positive and negative work can be determined. Also, the magnitude of the mechanical energy exchanges within each segment and between adjacent segments can be calculated and used in the assessment of the efficiency of the movement.

The basis of the energy analysis are the individual segment energies,

$$E_s = mgh + \frac{1}{2}mv^2 + \frac{1}{2}Iw^2 \qquad \text{Eqtn.1}$$

A summation of all segment energies over the gait cycle yields the total body energy, E (3). Such a curve, because it is an algebraic summation across all energy components (kinetic and potential) and all body segments, accounts for all exchanges of energy within each segment and between each segment. Figure 3 is such a curve for normal gait.

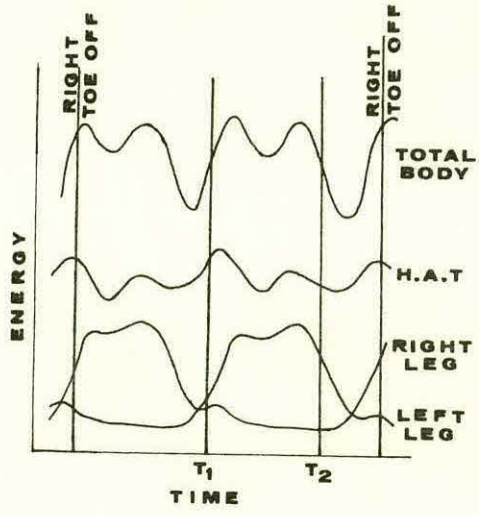

Figure 3

Increases in energy of this curve mean that muscle activity has resulted in a net positive work; conversely decreases in body energy reflect net negative work. A summation of all these positive and negative work changes over the gait stride yields the net internal work, W , done by the body (4). The mechanical energy "cost" of gait for normals has been found to be about 1.1 J/Kg.m. Thus a 60 Kg person walking 1000 meters does about 66,000 J work. The work calculations now permit a calculation of efficiency in level gait, and an assessment of the total energy exchanges between segments and within segments (5). Some new and surprising results arose from these analyses. Positive work efficiency was found to be in the 40-50% range compared with the 20-30% previously reported. This difference was attributed to the fact that previous techniques were not able to calculate the work done by the limbs themselves (internal work) nor did they account for any negative work being done. Also, the energy exchanges within and between segments was found to be 2/3 of the total observed energy changes. These last two findings now permit us to assess the efficiency of amputees or orthotic patients during level gait.

Kinetics

An assessment of the net effect of muscle activity or the influence of a prosthetic mechanism requires joint kinetic information: reaction forces and moments of force. Standard link segment equations (6) are employed which commence their analyses at the distal end of each limb segment and calculate successive forces and moments at the more distal joint until the trunk is reached. Simultaneous analysis of 2 arm and 2 leg segments is possible, with ground reaction forces introduced at the support foot. Properly synchronized ground reaction forces, F_X and F_Y, plus its centre of pressure, X_{cp}, are required. Figure 4 shows the joint moment histories for a normal adult. Such diagnostic curves in conjunction with EMG records has proved invaluable in the assessment of many types of pathological gait.

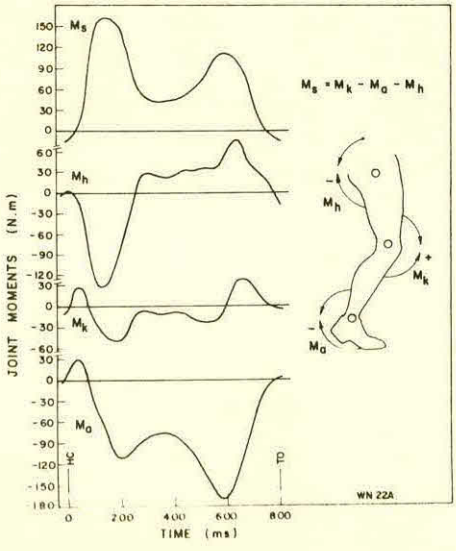

Figure 4

Power

A recent offshoot of the energy and kinetics programs is the assessment of mechanical power. Rates of energy changes in segments and at joints gives complete insight into the response of the muscles or prosthesis. The basic power equations calculate the muscle power,

$$P_m = M_j w_j \qquad \text{Eqtn. 2}$$

where M is the joint moment and w is the angular velocity. It has been shown that this power can be partitioned into 3 components: generation, absorption and transfer (7). Also, the passive rate of transfer of energy across the joint centre,

$$P_j = F_j . V_j \qquad \text{Eqtn. 3}$$

can also be calculated. F_j is the joint reaction force and V_j is the joint absolute velocity. These passive transfers are extremely important in the assessment of the dynamics of prosthetic and orthotic gait where muscle forces are minimal or absent. One major finding to date in slow and normal speed gait is the dominating influence of the ankle (plantarflexors) muscles in creating new energy once per step (8), and the relatively low contribution of the knee and hip muscles. The interpretation of these curves is somewhat complex and can best be seen by looking at anatomical plots such as Figure 5 which shows the rate of energy changes (power) taking place at

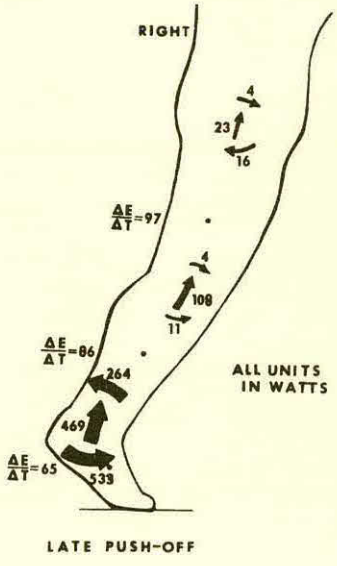

Figure 5

the ankle, knee and hip joints during late push-off. The arrows passing through a joint centre indicate energy transfer (see Eqtn. 3); the arrows as arcs around a joint show energy generation, absorption or transfer through the muscles (see Eqtn. 2). In Figure 5 we see the plantarflexors are generating energy at a rate of (533-264) = 269 watts, and are tranfering at 264 watts from the shank into the foot. For a detailed description of the total gait stride at all joints the reader is referred to reference 8.

Momentum

A final option available is the analysis of segment and total body linear and angular momentums. At present this has not been utilized for pathological gait; its primary purpose is in the assessment of rapid athletic movements with airborne phases.

CONCLUSIONS

A detailed summary of WATERLOO BIOMECH has been presented which describes a complete set of software programs capable of providing powerful diagnostic information regarding pathological gait. At present the programs have analysed over 200 normal and pathological trials in the saggital plane. Expansion to three dimensions could be readily accomplished by the addition of additional data and analysis arrays and by relatively simple additions to logic statements.

Acknowledgements

The authors wish to acknowledge the software contributions of many persons over the past 7 years: W. Kuryliak, G. Robertson, R. Wells and M. Pierrynowski. Also, the financial support of the Department of Kinesiology, the Medical Research Council of Canada and the Federal Department of Supply and Services is gratefully acknowledged.

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WATERLOO BIOMECH - PART II, APPLICATION OF BIOMECHANICS PACKAGE TO THE ASSESSMENT OF PROSTHETIC AND ORTHOTIC GAIT

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INTRODUCTION

Many types of biomechanical gait analyses can be conducted on amputees and orthotic patients. The assessments involve everything from a look at a simple plot of some data right up to a complete link segment analysis in three dimensions. The level of the assessment depends on many factors: type of questions being asked, availability of data and skill of the researcher in interpreting the results. The purpose of this paper is to illustrate the use of the package of computer programs, WATERLOO BIOMECH, as a flexible response to the problems of assessing individual patients or improved appliances. In all cases it is assumed that at least one stride of image data are available, and that it has been tested for validity and reliability.

LEVELS OF ASSESSMENT

Figure 1 depicts the various levels of assessment that can be conducted. The visual observer is faced with the tremendous task of measuring, describing and monitoring the patient's progress. In many cases, gross changes

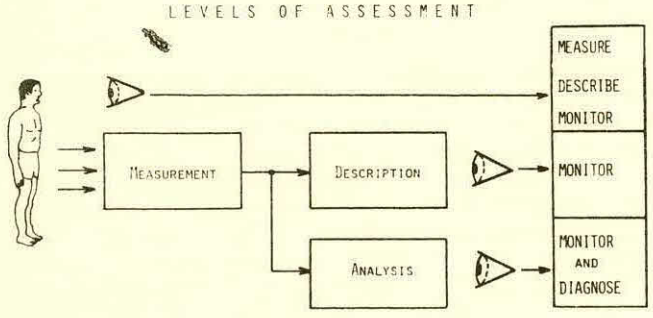


Figure 1

are readily noted. However, in complicated cases direct observation places a tremendous "overload" even on the most experienced observer. All measures are subjective and are almost impossible to compare with those obtained previously. The purpose of this paper is to look at higher levels of assessment that utilize objectively collected data.

Monitoring

Monitoring is the lowest level in terms of diagnostic power. It requires a description of one or more of the measured variables, such as knee angle or force plate time histories. The description can take many forms: pen recorder curves, plots of body coordinates, a stick diagram, etc. In Figure 2 we see a data system (cine camera) and two different descriptive

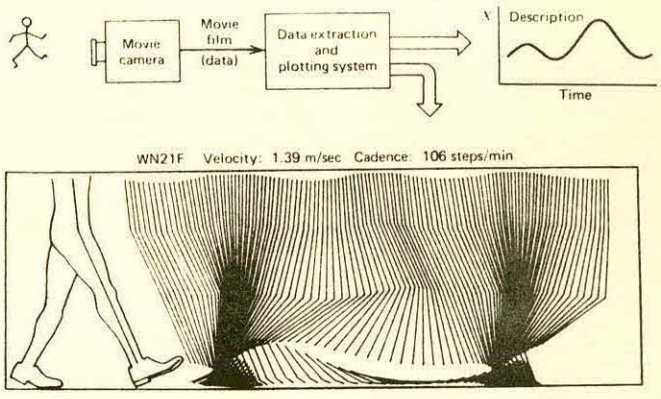


Figure 2

plots. The body coordinates can be plotted at regular intervals in time. Such plots are useful in monitoring detailed changes of a particular variable. A total description in the plane of movement is the stick diagram. Here, trajectories, velocities, and accelerations can be visualized. Note that a stick diagram summarizes a computer listing that fills 2 full pages.

Such descriptive plots only contain information regarding the return (or non-return) of the patient's gait towards a normal pattern. Thus, monitoring answers the question: Is the gait better now than last month? Is it more normal? Only in special circumstances does it answer the question...why? It is often inferred, circumstantially, that surgery, therapy or appliance modifications that have taken place between gait assessments are responsible for the monitored changes. This can be illustrated by two examples. Re-alignment of the prosthesis by the prosthetists can be monitored before and after, and because of the short time interval between measurements it is fairly safe to conclude that the observed changes were due to the prosthetists intervention. On the other hand, a therapy program can result in many changes that are more subtle, such that it may be impossible to conclude a cause-effect relationship. Also, major questions such as design and development are better answered using a higher level of assessment.

Analysis

Imaging measurement systems yield data that are suitable for a variety of anslyses. Analysis can be defined as any mathematical operation that is performed on a set of data to produce a variable that is not directly measureable. From the analysed data new information may be extracted to improve our assessments. Figure 3 is a schematic diagram to show the relationship between data

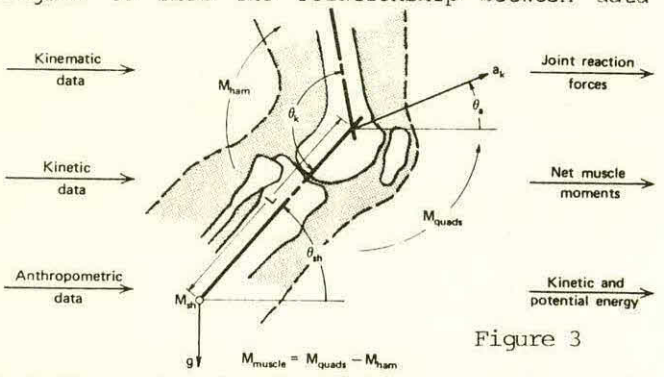


Figure 3

variables and analysed variables. Here kinematic data, kinetic (force plate) data and anthropometric data are inputs to link segment modelling, where computer analyses yields many valuable kinetic variables: reaction forces at each joint, net muscle moments at each joint, segment and total body mechanical energy. Visual inspection of these analysed variabled now yields the detailed diagnoses that were not possible from the original data descriptors.

Diagnoses

The entire purpose of a detailed assessment is to determine the cause of changes that we see, and thus give the clinician information to improve the gait or redesign the device. Thus, diagnoses should answer the question: what is right or wrong, and why?

EXAMPLE ASSESSMENTS

We would now present some typical assessments to illustrate the principles described above and to demonstrate the flexibility of WATERLOO BIOMECH to respond to the various levels of assessment.

COMPARISON OF ANKLE ORTHOSES

A total gait study assessed spiral and rigid polypropylone orthorses on a child with flail ankles. Each trial involved the collection of cine and force plate data in the Gait Laboratory in the Department of Kinesiology at the University of Waterloo. Coordinate data were converted using a Numonics Digitizer, and were scaled and corrected for parallax error and image (lens) distortion. Raw absolute coordinate data were then available to the package of programs described in Part I of this paper.

Monitoring

The descriptive variables that were monitored were the time histories of the joint angles at the ankle, knee and hip joints. Figure 4 shows these plots, the solid line is for the

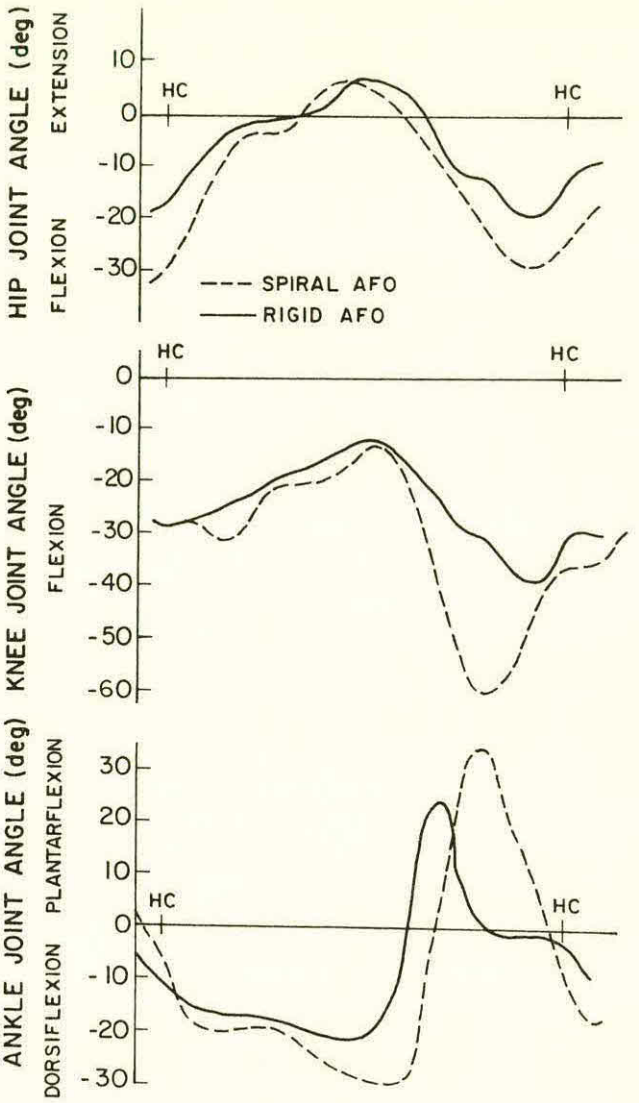


Figure 4

rigid AFO, the dotted line for the spiral AFO. At all joints the dynamic range of motion was greater for the spiral AFO than for the rigid AFO. Hip movement varied from 10^{o} extension to 30^{o} flexion for the spiral device, but only to 20^{o} flexion for the rigid orthoses. Knee for both devices was about 30^{o} at heel contact. Both orthoses resulted in a decrease in knee flexion to about 15^{o} at toe-off, but the spiral AFO resulted in far more normal swing phase flexion (60^{o} vs 40^{o}). Finally, the ankle movement was greater than that seen in normal children (1). Normal gait has 20^{o} dorsiflexion during late stance, shortly after heel-off. The rigid AFO had 20^{o}, but the spiral allowed 30^{o} of dorsiflexion. Finally, at toe-off the plantarflexor angle exceeded the normal value of about 15^{o}, the rigid AFO was 25^{o}, the spiral AFO was 35^{o}. The conclusion of monitoring was that the spiral AFO was more normal at the hip and knee, but less normal at the ankle. However, there is very little we can conclude regarding the cause of these differences. Also, we would have no information regarding the relative efficiencies of each device.

Analyses

Two types of analyses were carried out on these data: joint moment and mechanical energy analyses. The joint moment plots for both orthoses are presented in Figure 5 covering the period of stance and show drastic differences in their histories.

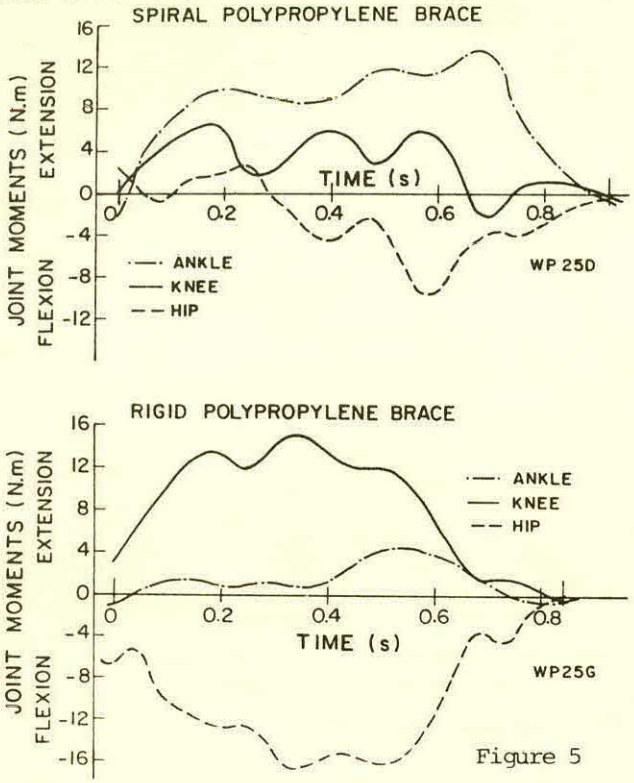


Figure 5

For both devices this child had an almost flat foot at initial weight bearing. Thus, as body weight was accepted the shank rotated over the foot causing a plantarflexion (extensor) moment. The moment created by the spiral AFO was quite high initially (10 N.m), rising to a peak of 14 N.m at push-off. For a child of this mass (21 Kg) the extensor moment is quite good, and not excessive compared with normals. It can be seen that spiral brace is many times better as a moment "generator". The other two joints have responded to these passive moments generated by the brace. Knee stability in the rigid brace is achieved almost entirely by the knee extensor muscles while the spiral brace gait has only a small knee extensor moment (because its large plantarflexor ankle moment prevents excessive rotation of the shank over the foot). Because of the extra large knee extensor activity for the rigid AFO the thigh is accelerated rapidly over the knee, and to keep the trunk from falling backward the child compensates with above normal hip flexor activity. To summarize, the moment curves show that the spiral AFO gives better ankle stability, thus relieving the knee extensors of major activity. The rigid AFO, on the other hand, had less than adequate support thus requiring excessive knee extensor activity plus a compensating excess hip flexion.

Figure 6 shows the total body energy for these two devices. The spiral brace has a more normal curve because of the increase in its

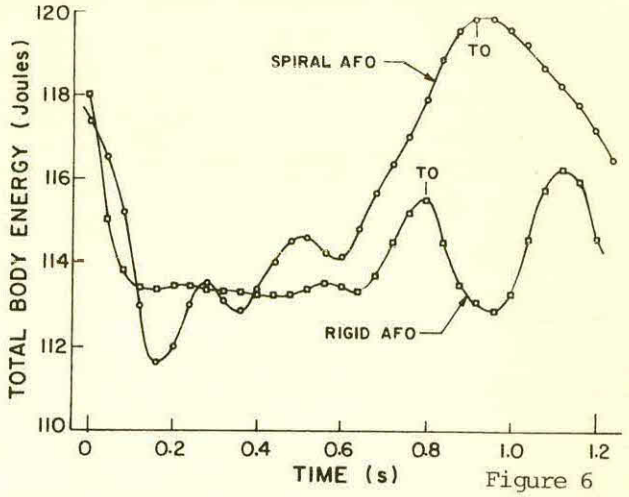


Figure 6

energy during swing which reflects the more energetic swing through. The rigid AFO resulted in a jerky and hesitant swing through, as indicated by the sudden drop in energy during mid swing. If we sum all the energy increases (positive work by body muscles) plus all the energy decreases (negative work by muscles) we calculate the total internal work done during the stride. Normal values have not yet been established for children, however, for adults the mechanical energy cost of walking was determined to be 1.1 J/Kg.m (2). Both of these devices resulted in a higher cost: 1.52 J/Kg.m for the rigid AFO, 1.66 J/Kg.m for the spiral orthoses. The difference between these two is only 9%, probably not significant, indicating that there is little difference between their mechanical energy requirements. Had there been a significant difference the inefficiency would be easily pinpointed by identifying the points in time when there were major increases and decreases in body energy.

ASSESSMENT OF HIP DISARTICULATION AMPUTEE

The following example shows how these computer programs can be used to assess the function of a high level prosthetic device. This assessment has been reported before (3) and is summarized here. The amputee was an active 20 year old congenital hip disarticulation patient wearing a prosthesis developed over 20 years ago (4). The aim of the assessment was to measure kinematic and mechanical energy variables to yield information that may improve the redesign of the prosthesis.

Kinematic Descriptors

Basic temporal data showed the stance/swing ratio to be .86 for the prosthetic limb versus 1.86 for the normal limb. This compares with 1.5 for normal gait, indicating that he is strongly favouring his good leg.

Figure 7 plots the forward velocity of the heel of each foot against time. Because of the short duration of swing of the good leg (.43 sec) compared with the prosthesis (.73 sec) the maximum forward velocity for the normal side was considerably higher (3.9 m/s) than for the prosthetic leg (2.5 m/s).

Figure 8 shows the vertical displacement of

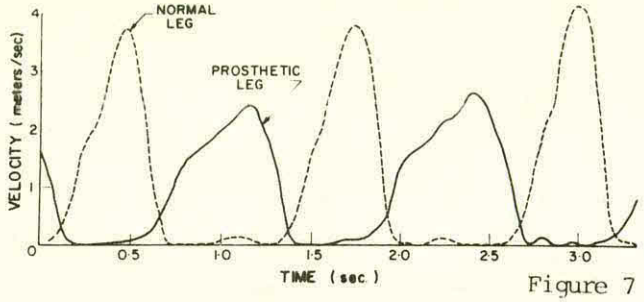


Figure 7

VELOCITY OF HEEL IN FORWARD DIRECTION (HIP DISARTICULATION AMPUTEE)

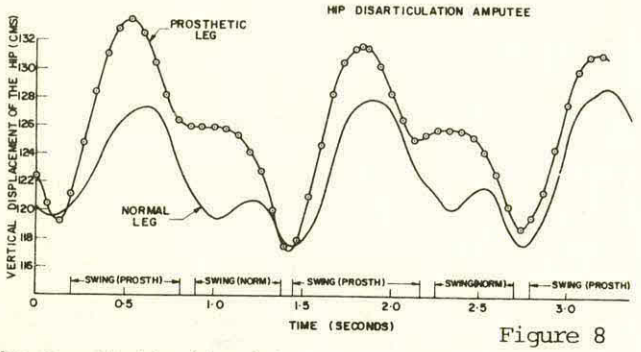


Figure 8

the prosthetic hip joint compared with the normal side. The average change was 13.5 cms on the prosthetic side compared with 10.0 cms on the normal side. Both of these displacements were considerably greater than the 5.5 cms recorded for normal adults (5).

Both of these kinematic curves shows interesting differences, but the real impact that they have on the movement are not available until a proper energy analysis was done.

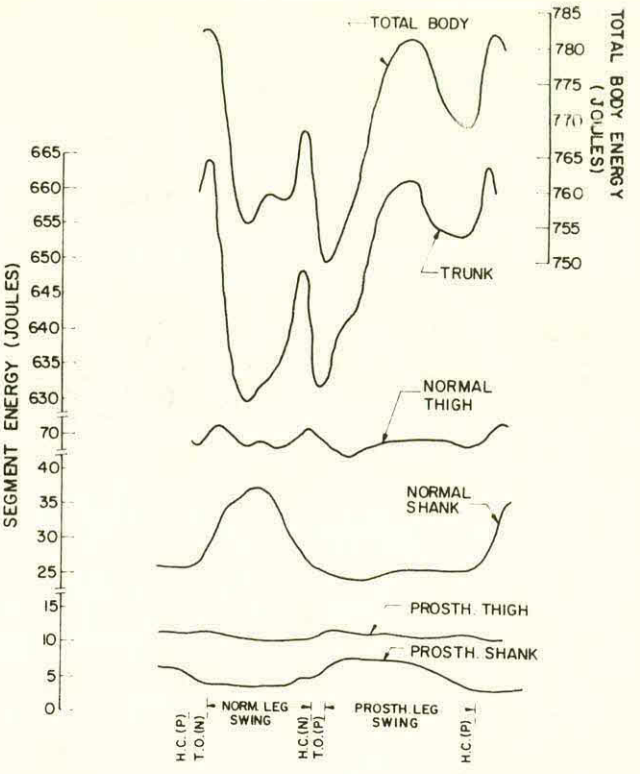
Mechanical Energy and Inernal Work Done

The results of the energy analysis are summarized in Figure 9. Each segment's energy was calculated to be the sum of its potential and kinetic energy components and the total body energy is the sum of the energies of all body segments (6). Note that the energy scale is biased to reflect the potential energy of each segment above a ground datum of zero. The following can be deduced from these curves:

(a) The contribution of the prosthesis to the total body energy is insignificant: this was traced to its low swing velocity (Figure 7) and low mass (less than half of the normal limb).

(b) The dominant energy changes are due to the trunk and these are a result of potential energy changes. The large vertical displacement of the trunk results from hiking of the prosthetic hip during swing, and because of a rise of the normal hip as he vaults over that leg.

The total internal work done by this amputee was calculated by integrating the energy changes in the total body curve (2). These positive and negative work components were found to be 116 joules for the stride. The amputees stride length was 2.4 meters and his mass was 76.6 Kg. Thus the mechanical energy cost was .63 J/Kg.m. This is quite good compared with normals (2); however, when we realize that all this energy is being generated and absorbed by his good limb it is apparent that he may fatigue fairly early.



BODY ENERGY (HIP DISARTICULATION AMPUTEE)

Figure 9

Also, a major redesign is indicated to decrease the energy associated with hiking.

CONCLUSIONS

This paper has illustrated the flexibility of the computer package, WATERLOO BIOMECH, in the assessment of prosthetic and orthotic gait. The case studies presented show the various levels of assessment, from descriptive monitoring of kinematics up to diagnostically powerful mechanical energy and joint moment assessments.

Acknowledgements

The author acknowledges the support of the Medical Research Council of Canada.

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A GENERAL EMG-POWER MODEL

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ABSTRACT - A general EMG-power model is developed in the form of a non-linear differential equation. The model takes into account muscle parameters which are length-dependent. Experiments were performed to verify the model over a large range of elbow flexion. The experimental results support the power model indicating that the root mean square value of the myoelectric signal is a direct measure of muscle power.

INTRODUCTION

Since the beginning of modern electromyograpy models have been proposed in an attempt to relate the electrical and mechanical activity associated with the contraction of skeletal muscle. It is now well established that muscle tension is directly related to the processed myoelectric activity during isometric contractions, [1]. There is still some dispute, however, as to whether this relationship has a linear or non-linear characteristic, [2]. This confusion is lessened by more recent investigations, [3], indicating the need for a standardization of experimental procedures. Muscle velocity has also been shown to be directly related to the processed myoelectric activity under the constraints of constant velocity contractions, [4]. From these observations it seems there may be a more fundamental relationship which relates muscle velocity, muscle tension, and the processed myoelectric activity. This is the approach taken by Patla et al, [5], and from which the model given by Equation 1 was developed.

$$K\sigma(t) * h(t) = bMg + (Mg+a)V(t) + b\dot{M}V(t) + \dot{M}V(t)V(t) \quad .....1.$$

where M is the mass of the load on the muscle, g is the acceleration due to gravity, V(t) is the velocity of shortening of the muscle, $\dot{V}(t)$ is the derivative with respect to time of V(t), a and b are constants determined by the muscle physiology, h(t) is a second order linear system convolved with the root mean square value of the myoelectric signal, $\sigma(t)$, [6], and K is a constant of proportionality.

This variation on Hill's basic equation relates the root mean square value of the myoelectric signal to the mechanical power output of a muscle. In this model certain intrinsic length-varying parameters were taken as constant which meant that verification of the model could only be done at one muscle length, (joint angle).

In the present work Patla's power model is modified to include these length varying parameters. The general EMG-power model is then verified over a large range of muscle lengths. The model illustrates that the processed myoelectric activity is a measure of muscle power, and therefore can only be used to find muscle force directly under isometric conditions.

THE GENERAL EMG-POWER MODEL

For Equation 1 to be valid over a range of joint angles it can be rewritten as

$$K(\theta)\sigma(t) * h(t) = bMg + (a(\theta)+Mg)V(t) + b\dot{M}V(t) + \dot{M}V(t)V(t) \quad .....2.$$

where $K(\theta)$ and $a(\theta)$ denotes the variation of these parameters with joint angle, θ.

Under isometric constant force conditions Equation 2 reduces to

$$K(\theta)\sigma(t) = bMg \quad .....3.$$

The dependence of $K(\theta)$ on joint angle was investigated in a previous study on the elbow flexor group, [7]. This study considered $K(\theta)$ to be the efficiency of the muscle and the relation of $K(\theta)$ to the

muscle length , ℓ, was found to resemble the length-tension relationship as given by Pertuzon, [8], for this flexor group.

In Equation 2 the constant b is usually given in terms of V_o, the maximum velocity of shortening of the muscle under zero load and with maximum excitation. Although V_o cannot be measured on intact muscles, its value may be estimated as

$$V_o = n\ell_o$$

where ℓ_o is the rest length of the muscle and n is an integer given values ranging from 2 to 10 in the literature, [9,10].

In the present work a value of 6 was assumed for n. The biceps brachii has a rest length of approximately 150 mm, thus

$$V_o = 900 \text{ mm/s}$$

and from the work of Hill, [9], and Petrofsky and Phillips, [11],

$$b = .25V_o$$

or $b = 225$ mm/s

For a muscle of a given rest length, V_o is constant, and therefore b may be assumed constant over the entire range of elbow flexion.

The coefficient $a(\theta)$ in Equation 2 is usually defined in terms of P_o, the maximum isometric contraction. It is well known that P_o varies with muscle length and therefore it is reasonable to assume that $a(\theta)$ varies in a similiar manner.

The general EMG-power model given by Equation 2 indicates that once $K(\theta)$, b, and $a(\theta)$ are determined an estimate of the muscle velocity can be obtained from the myoelectric signal root mean square value via a non-linear system. The following experiment was carried out to verify this result.

EXPERIMENT

The apparatus, shown in Figure 1, consisted of a metal frame which housed a large central pulley. An aluminum disk was attached to the axis of the pulley and the subject was seated with his right upper arm resting on an adjustable horizontal platform. The axis of rotation of the elbow was matched to the axis of the pulley disk system while a padded cuff held the subject's right forearm firm against the disk. In this position any mass attached to the pulley would exert a force perpendicular to the wrist for all angles of flexion. A potentiometer was coupled to the pulley shaft for measuring joint angular position as a function of time. Angular velocity and acceleration were obtained by differentiation.

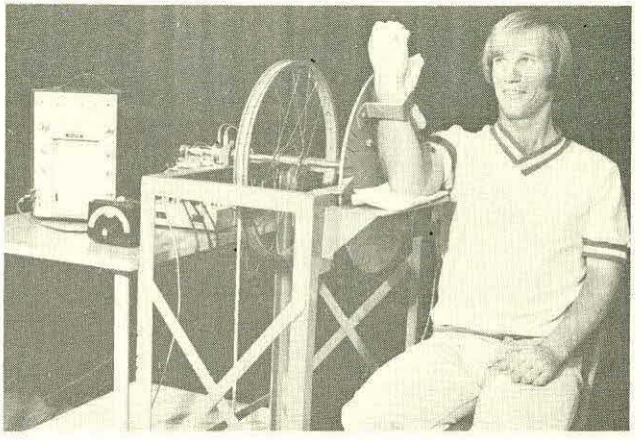


Figure 1.

The myoelectric signals were obtained using Beckman surface electrodes in a bipolar configuration. These were placed over the biceps with a 20 mm spacing. A transverse arrangement was chosen in order to reduce the effect of electrode displacement relative to the muscle during flexion.

The experiment was performed in two parts. Part one was concerned with determining $K(\theta)$ for the range of elbow flexion of 160° to 40°, (180° is full extension). From Equation 3 it is seen that this can be done with isometric constant force contractions. For a given mass, M, and joint angle, θ, the subject was required to maintain an isometric contraction for a period of 5 seconds while $\sigma(t)$ was measured. This procedure was repeated for elbow angles from 160° to 40° in steps of 5° and for various values of mass. From the data collected the coefficient $K(\theta)$ could be calculated for the entire elbow flexion range.

Part two was concerned with determining $a(\theta)$ for the same range of elbow flexion, and with collecting velocity data to verify the model. The subject was required to perform a repetitive flexion and extension of the elbow with a given load, M. During the flexion phase of the motion angular position, velocity and acceleration, along with the biceps and triceps myoelectric activity, were sampled simultaneously using a PDP 11/34 minicomputer.

Due to the nonstationary nature of

the random myoelectric signal an estimate, $\hat{\sigma}(t)$, of $\sigma(t)$ must be obtained by ensemble averaging. For this purpose the subject ideally should produce a number of identical elbow flexions. To avoid fatigue the subject was permitted to stop and continue the flexion routine at any time during the run while the computer accepted only those runs which had the same angle and velocity time courses, (within a preset error). The peak velocity of flexion was controlled by having the subject match the beat rate of a metronome. The second order system, $h(t)$, in Equation 2 was approximated by a pure delay of 100 ms. This procedure was repeated with four subjects and for various loads and peak velocities. From the data collected an estimate, $\hat{a}(\theta)$, of $a(\theta)$ is determined. Thus knowing $\hat{a}(\theta)$ an estimate of V(t) may be found from $\hat{\sigma}(t)$. The results are given in the following section.

RESULTS

Plots of $\sigma(t)$ versus isometric force for a given joint angle showed a linear characteristic with a high degree of correlation, ($0.85 < r < 0.99$). It is seen from Equation 3 that the coefficient $K(\theta)$, may be found from the slope of the regression line relating $\sigma(t)$ and isometric force at a given joint angle. Figure 2 shows the variation in $K(\theta)$ over the range of muscle length studied. In this figure joint angle has been expressed in terms of the muscle length, ℓ, at that angle. This figure illustrates that the amount of muscle force produced for a given level of excitation is dependent upon muscle length. In this manner $K(\theta)$ is analogous to the length-tension relationship of the muscle and the classical form of this relationship is clearly shown by Figure 2.

The coefficient $a(\theta)$ was calculated using $K(\theta)$ and anisometric data. It is a characteristic parameter of the muscle, and therefore its value for a given subject and muscle length should not vary from day to day. Figure 3 is a plot of $a(\theta)$ versus muscle length for one subject over three separate days of experiments. As suggested these curves show that the calculated value of $a(\theta)$ remains relatively constant from day to day.

Figure 4 shows measured and estimated velocity versus time curves for one subject. The normalized root mean square error, ε_n, between the measured velocity and the velocity estimated from $\sigma(t)$ was consistently below 15% for all subjects. A large portion of this error may be attributed to the error involved in estimating $\sigma(t)$. For the system used the theoretical limit of the estimation error is given by Bendat and Piersol, [12], as

$$\varepsilon_n = 1/(8B_n \tau N)$$

where B_n is the statistical equivalent bandwidth of the myoelectric signal; (100 Hz), τ is the time constant of the measuring system; (0.01 s) and N is the number of runs in the ensemble average; (10).

This gives a value of 11% for ε_n and indicates that the anisometric data supports the model of Equation 2.

Further evidence to support the power model is given in Figure 5. This is a plot of the input power to the muscle, estimated by the left hand side of Equation 2, versus the mechanical output power given by the right hand side of the equation and includes data from all four subjects. The slope of the regression line between input and output power is approximately unity, and there is a very high degree of correlation. This implies that $\sigma(t)$ is a direct measure of muscle power.

CONCLUSIONS

The EMG-power model developed by Patla et al, [5], has been modified to make it valid over a large range of muscle lengths. The model indicates that the root mean square value of the myoelectric signal is directly related to muscle power which is in turn related to muscle velocity through a nonlinear differential equation. The model implies that muscle force cannot be estimated directly from the myoelectric signal during anisometric contractions. The experimental results support the general EMG-power model.

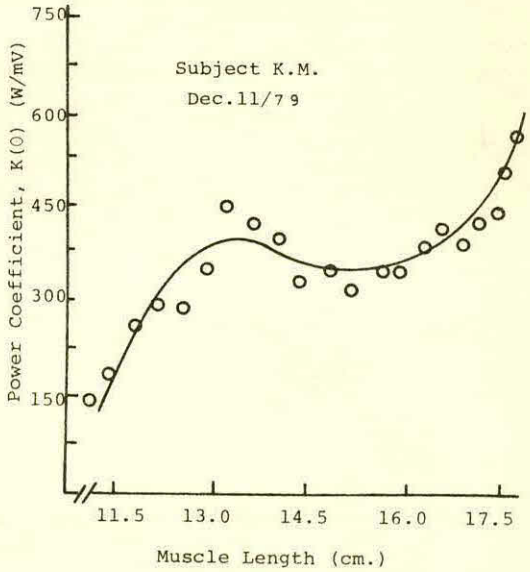
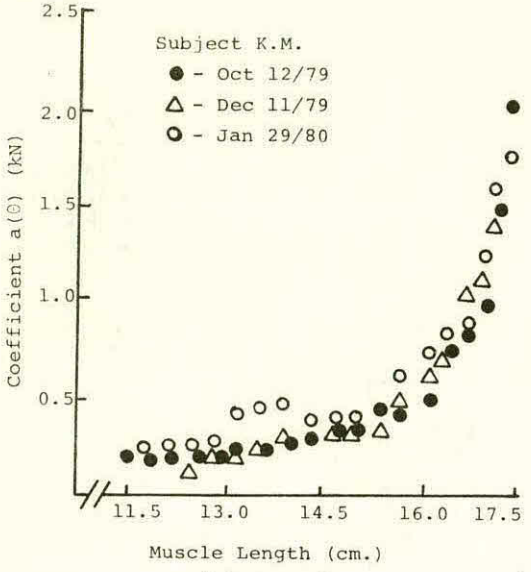


Figure 2.



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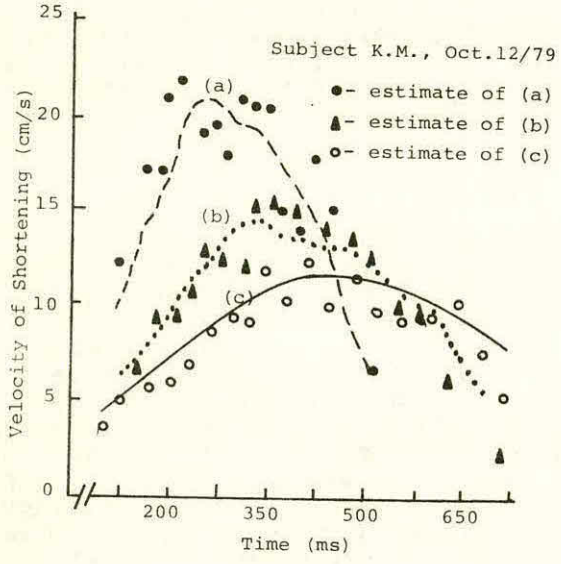

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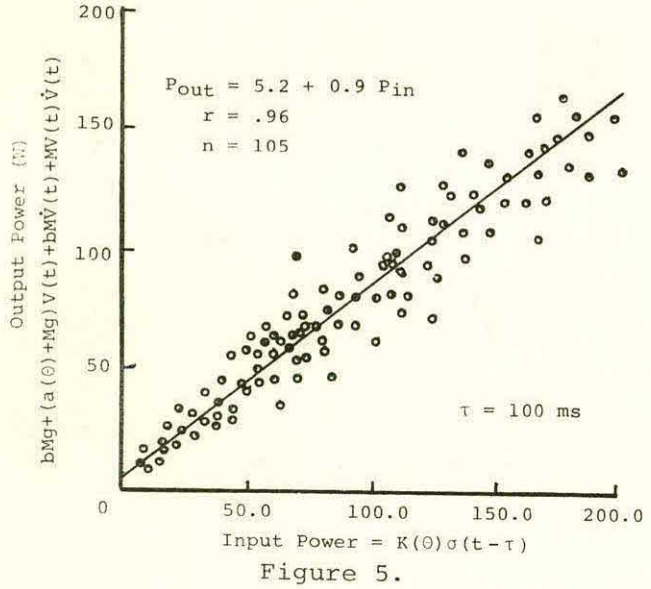

Figure 5.

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WORK CAPACITY AND POWER OUTPUT TESTING: A NECESSARY ADJUNCT TO REHABILITATION EVALUATION

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ABSTRACT

The knee maximum torque, total work capacity and power output of sixteen post-menisectomy clients was examined using an isokinetic dynamometer during the course of their rehabilitation exercise programs. Initial testing indicated pathological ipsilateral and contralateral imbalances on all parameters. Final tests revealed that while knee extension maximum torque increased substantially (91%), work capacity (150%) and power output (163%) increased considerably more. Measurement of work capacity and power output should be integral of the rehabilitation process to properly assess progress and modify program content accordingly.

INTRODUCTION

The goal of all rehabilitation is to progress towards the restoration of normal function in the affected area of the body. Rehabilitation is, by its very nature, a dynamic process. As the patient changes in functional ability, the therapist must accommodate these changes with revised program activities designed to enhance the rate of change towards normal. These accommodations may be made based on a purely subjective analysis of progress, or program changes may evolve from objective assessment of the patient's functional status. Objective assessment should include the examination of all those aspects of performance which comprise full functional ability.

Isokinetic dynamometry is well known to those in the rehabilitation field, however, generally it is little used and insufficiently understood. Conventional therapy employs static dynamometer readings of force output but in a functional sense it is the parameters of work capacity and power that are important in the assessment of functional capability. The isokinetic dynamometer not only allows for the examination of range of motion and force output but also permits the measurement of work and power development throughout the range. The present paper describes the integration of isokinetic dynamometry with the rehabilitation process, and illustrates the usefulness of isokinetic information in updating therapeutic program activities.

METHODS

Sixteen post-menisectomy clients of the Exercise Research Clinic acted as subjects for this study. These clients were medically referred for evaluation of and rehabilitative programs for knee range of motion and strength. Each client was given a pre-test of knee capability on a "Cybex II" Isokinetic Dynamometer (Lumex, Inc., Bay Shore, N.Y.) previously calibrated according to the method of Moffroid et al[4]. The instrument allows for the measurement and recording of turning force applied to the lever arm. Since the velocity of the lever arm is held constant, variations in turning force applied are met with accommodating resistance and are recorded directly. Thus, the turning force capabilities of the joint may be examined throughout the entire range of motion.

Each client had the extension and flexion capability of both knees tested in a seated position. The principles of parallel and rotational alignment, and stabilization were strictly adhered to.[1] The client sat at the edge of a padded table with arms crossed in front. The trunk was supported vertically and the test leg thigh was "locked" into a padded channel which prohibited lateral movement. Knee extension began at 90° of flexion (or maximum flexion) and knee flexion began at full extension.

During each testing session the client performed three to five extension-flexion pairs of efforts at a machine speed of 30°/sec. Maximum torque (T-max), total work and average power for extension and flexion were calculated from the largest isokinetic curve obtained.

Eight subjects participated in the program for two weeks or less, while eight were tested over a three to sixteen week period. The average program duration was seven weeks. Knee extension, knee flexion and the ratio of extension/flexion data were obtained for T max, total work and average power during each test session. These parameters were compared with normative data in the literature to assess the degree of impairment. In addition, the changes in these data were examined over the course of the rehabilitation process in both the affected and normal limbs.

TABLE I

Mean Isokinetic Responses during Initial Test

	Tmax (N.m)		Work (Joules)		Power (Watts)	
	AFFECTED	NORMAL	AFFECTED	NORMAL	AFFECTED	NORMAL
KNEE EXTENSION						
FEMALE	83	122	62	87	16	35
MALE	110	214	83	171	32	63
KNEE FLEXION						
FEMALE	54	75	56	62	22	24
MALE	103	133	90	115	36	42
EXTENSION TO FLEXION RATIO						
FEMALE	1.56	1.64	1.11	1.45	.85	1.49
MALE	1.04	1.61	0.96	1.53	.93	1.56

TABLE II

Mean Isokinetic Responses during Final Test

	Tmax (N.m)		Work (Joules)		Power (Watts)	
	AFFECTED	NORMAL	AFFECTED	NORMAL	AFFECTED	NORMAL
KNEE EXTENSION						
FEMALE	140	161	118	136	38	44
MALE	157	212	148	181	52	63
KNEE FLEXION						
FEMALE	81	91	89	56	28	29
MALE	118	121	149	125	50	43
EXTENSION TO FLEXION RATIO						
FEMALE	1.67	1.77	1.21	1.57	1.29	1.55
MALE	1.42	1.77	1.17	1.50	1.24	1.57

RESULTS

When the data presented in Table I are compared with normative data[1], it is apparent that the initial torque, work and power responses of the clients' normal limb are well within the normal range. (Standard score 75). However, their responses with the menisectomy affected limb are well below these values (Standard score 50). Their extensor to flexor ratios are well down into the pathological region (below 1.25:1) largely as a result of extensor weakness. When contra-lateral extensor dominance is examined this pathology becomes more apparent. Normally, the extensor capability of the weaker limb is 83% ± 13% of that of the stronger limb. The same comparison amongst the menisectomy clients evidences a 65% relationship of affected to normal limb.

A comparison of Table I and Table II pinpoints the effects of the rehabilitation process. While all isokinetic parameters for the affected limb have drawn much closer to or exceeded normal limb values by the final test, the changes from initial test to final test reveal important differences between parameters. Knee extension maximum torque of the affected limb increased 91% from initial to final test. During the same time period, however, total work increased 150% and average power increased 163%. The same trend is apparent for knee flexion, although the magnitude differs; maximum torque increased 36% while total work and average power increased 79% and 52% respectively.

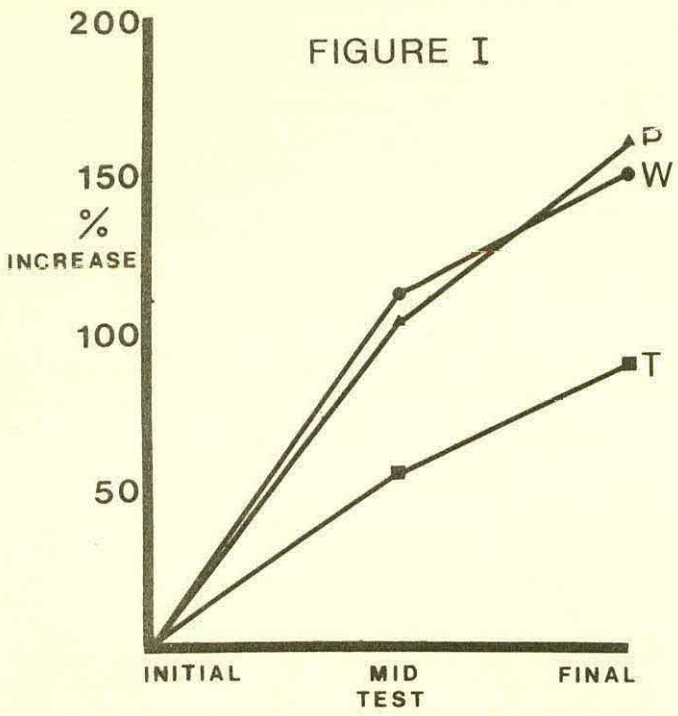

FIGURE I Knee extension isokinetic responses: per cent increase during rehabilitation. (T-torque max, W-total work, P-average power)

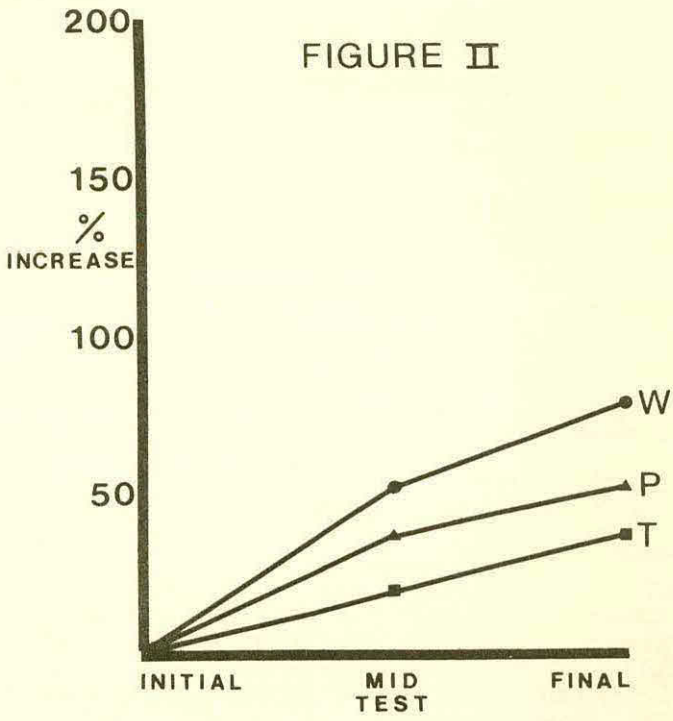

FIGURE II Knee flexion isokinetic responses: per cent increase during rehabilitation. (T-torque max, W-total work, P-average power)

DISCUSSION

Had a standard isometric dynamometer been used to assess progress, it would have been impossible to assess work and power changes over time. For this group of menisectomy clients, the extra information provided special insight into the rehabilitation process.

Those in rehabilitation fields are well aware of the necessity to improve the power output of musculature.[5] As Hellebrandt[2] has said, "Power is the variable on which extension of the limits of performance depends." Thus, as power is a very important part of the return to functional ability, many of the exercises used in our Clinic were accomplished at progressively higher velocities as rehabilitation proceeded. The rate of change in the velocity of movement during these exercises was mediated, to a large degree, by the isokinetic evaluations performed.

Many of the strength and endurance building exercises used in our program were isotonic in nature. Consequently, muscle potential was developed relatively more in the inner and outer portions of the range of motion than was developed in the middle of the range.[3] Examination of the changes in work capacity and, more specifically, of the changes in the shape of the isokinetic work output curve provided invaluable information for the modification of client programs.

In summary, it can be said that while objective analysis of static dynamometer output will provide an indication of change in rehabilitation progress, it presents an incomplete picture. The isokinetic dynamometer provides not only force output information but also the work and power output data so necessary to successfully modify the therapeutic regimen.

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PROVIDING AIDS FOR THE HANDICAPPED - THE SWEDISH WAY FROM IDEAS TO AIDS

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Depending on the definition, between 0.5 and 1.5 million people in Sweden are considered to be handicapped. Most of them live a comparatively normal life. For the daily living, many of them depend upon technical aids like wheelchairs, tape recorders, stoma aids, prostheses or hearing aids.

In principle, all aids necessary for daily life or the job or school situation are provided free of charge for the individual user. During 1979 the county councils paid more than 350 MSkr (70 MUS$) for such aids. The estimated level during 1980 is about 500 MSkr (100 MUS$). The government and the county councils play important roles in the process. Research and development (R & D) are financed by the government through The National Swedish Board for Technical Development (STU) to an amount of about 7 MSkr/year (1.5 MUS$). STU's principal task is to contribute to the strength of the economy and to social development in general by supporting and initiating the development of new products, methods and systems. This is done for example by promoting development of products and systems satisfying needs of the society among which are aids for the handicapped. The government and the county councils in common pay about 12 MSkr/year (2.5 MUS$) for research administration, testing and information, the executive body of which is the Swedish Institute for the Handicapped. The Institute and the Swedish Board for Technical Development cooperate in long term planning concerning investigations, R & D and production.

In order to supply an aid free of charge, several conditions should be fulfilled. One is, that the aid should be available at a reasonable price (which does not necessarily mean a low price) and with a responsible marketing organization. Another is, that the aid should fulfill demands for testing set by the Institute. A third is, that the prescribing and training personnel is properly educated and well informed.

The first set of conditions normally don't cause problems as far as goods already in production are concerned (e.g. tape recorders, binoculars, wheelchairs and hearing aids). However, as soon as aids under development are concerned problems related to risk capital for the R & D, production process and marketing appear. Concerning the R & D phase, STU can provide the individual inventor as well as industrial companies with risk capital. Also, since more than two years, special money devoted for the start of serial production and risk taking for the first serial production can be provided by the Institute. The experience is very positive. A range of products from very simple to rather complicated ones have been developed and put into production.

The demand for testing has put rather big pressure upon the Institute to put up with specifications on various kinds of aids as well as a testing capacity corresponding to the demands from the handicapped population and the producers. A large portion of the testing resources of the Institute is now tied to periodic testing in conjunction with economical agreements made between the central purchasing body of the county councils (SUB) and the suppliers.

It is very important that the society has an effective organization for the application of aids at it's disposal. As mentioned, the county councils in Sweden have the responsibility of providing most aids to the individual user. The prescribing and training personnel is very much depending on comprehensive product information, information about new methods and ideas as well as further training and education. It is the task of the Institute to take steps and measures to provide the personnel with requested support of that kind.

Sweden is a small country with a comparatively small number of disabled people. In order to provide a comprehensive set of aids at reasonable prices both the National Swedish Board for Technical Development and the Swedish Institute for the Handicapped are striving at international cooperation, ending up in import as well as export of aids. Research and development, agreement on specifications and testing methods as well as exchange of information in general are important tasks for this cooperation.

THE JAPANESE VOCATIONAL REHABILITATION SYSTEM AND POLICY

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ABSTRACT

This paper describes the Japanese system and policy for employment and training of the physically handicapped. The data concerning the handicapped population and their employment situations are presented and the employment systems for the handicapped are briefly described. One of the most significant aspects of the Japanese system is the Physically Handicapped Persons Employment Promotion Law, which specifically sets the minimum quotas for the physically handicapped in mainstream private and public sector employment. This paper discusses and evaluates its significance and impact in employment of the physically handicapped from the resource allocation standpoint.

INTRODUCTION

Japan is one of a few countries that have developed a comprehensive rehabilitation program for their handicapped population. Japan's growing economy has enabled her to allocate increasing resources for the benefit of the handicapped. And, being the only sizable minority of the Japanese society, the handicapped population has received the undivided attention of society and government. One characteristic of the Japanese rehabilitation program is the strong vocational orientation encouraged by public policies to enhance the allocative efficiency in resource use. The objective of this paper is to describe and analyze the Japanese vocational rehabilitation system with a particular emphasis on the Physically Handicapped Persons Employment Promotion Law. In view of the growing needs of gainful employment for the handicapped accompanied by the recent handicap movement, a review of the Japanese system would be useful in providing a useful insight to training and employment of the handicapped in the U.S.

HANDICAPPED POPULATION AND THEIR EMPLOYMENT (2,4)

The total number of physically handicapped persons was estimated to be 1,410,000 according to a survey conducted in 1970 and that of mentally handicapped persons was estimated to be 310,000 according to a survey conducted in 1971. Among the physically handicapped, the number of adults aged 18 or over was 1,314,000. This amounted to approximately 1.8% of the total Japanese adult population. This paper primarily concentrates on the vocational rehabilitation systems and policies toward this handicapped population because (1) they account for the large portion of the entire handicapped population, (2) the major effort of vocational rehabilitation has been directed toward the physically handicapped, and (3) information on vocational rehabilitation for the mentally handicapped is limited.

In Japan, physically handicapped persons are evaluated and certified by the Prefecture (state) as handicapped. They are classified into six categories according to their severity of disabling conditions. This classification is an important aspect of the Japanese system because the types of the services and the amount of the pensions available to handicapped individuals depend on their grades.

The disabled population classified according to the disabling condition was: 19% visual impairment, 18% speech and hearing impairment, 58% orthopedic disorders (including cerebral palsy), and 5% circulatory and respiratory disorders. Persons with multiple handicaps were estimated to be 121,000.

In 1970, an estimated 579,000 physically handicapped persons, or 44.1% of all physically handicapped adults were employed. The employment ratio has increased by 4.8% from 1965. The employment rate according to the disabling conditions was: 34% of those with visually impairment, 43.4% of those with speech and hearing impairment, 45.1% of those with orthopedic disorders, and 3.8% of those with circulatory or respiratory disorders.

The manufacturing sector employs 30% of the working handicapped persons. This is followed by the agriculture, fishery and forestry sector which hires 27%. The industrial/occupational distributions by disability type are substantially different from one disability type to another.

EMPLOYMENT SYSTEM AND POLICY FOR THE HANDICAPPED

Japan has pursued employment and training of the handicapped in both sheltered and nonsheltered arrangements. Employment is sheltered[2] when it is accomplished within a noncompetitive and protective environment. Employment is nonsheltered when it is accomplished in mainstream private businesses or the public sector. While the major objective of mainstream employment is economic gain and security, sheltered employment has a dual goal of economic gain and rehabilitation. Under the Japanese system, sheltered employment is

regulated by the Ministry of Health and Welfare while mainstream employment is regulated by the Ministry of Labor. Sheltered employment is provided by nonprofit organizations[3] while mainstream employment is provided by private enterprises for profit or by public and government agencies.

The sheltered employment system in Japan (4,5) is essentially similar to the sheltered workshops system in the United States (1,8). It is the Japanese mainstream employment system and policy that is most unique.

Model Factory

The model factory provides a special form of mainstream employment. A model factory is a privately run company specializing in the employment of the physically and mentally handicapped. It is different from sheltered employment in the following aspects: First, both disabled and able-bodied workers are employed in production, whereas all production workers are handicapped persons in sheltered workshops. Second, it is a profit organization which has no restrictions in distributing its profits, whereas the sheltered workshops are nonprofit organizations. Third, being a business organization, it emphasizes the improvement of the productivity of handicapped workers. Frequently, it utilizes newer and more productive or mechanized equipment to compensate for the low productivity of handicapped workers.

The Ministry of Labor requires model factories to maintain at least a 50% ratio between handicapped and nonhandicapped employees, exclusive of executives. In this aspect, model factories also differ from regular businesses. Thus, it may be considered as an intermediate step between the workshop and complete mainstream employment. We may consider it mainstream employment because (1) it is regulated by the Ministry of Labor and (2) workers at model factories can be considered completely vocationally rehabilitated.

In 1979, 60 model factories were in operation primarily in light manufacturing areas. The government plans to expand them by 25 each year for the next five years. A sample survey (7) of ten model factories, including six factories primarily for the physically handicapped, two for the aurally handicapped and two for the mentally retarded, showed that the average number of employees was 67, of which 55.6% were handicapped workers. The average annual earnings were ¥1,035,078[4] for the physically handicapped and ¥701,871 for the mentally retarded. The average earnings for the able-bodied workers were ¥1,307,027.

Model factories offer many services to the handicapped workers. These services are related to both production and nonproduction activities. Because of these extra services, model factories have a significant cost disadvantage. The survey of the ten model factories reported that the model factories incurred an average of ¥760,000 of additional annual costs per handicapped worker. Compared to similar sized regular companies, model factories' investment in factory and equipment was 1.7 times greater. Thus, model factories could not compete with other companies in the open market if they did not receive government subsidies. The federal government provides funds covering four-fifths of the cost of constructing factories and housing for handicapped workers and purchasing of machines. The capital subsidies are generally limited to one hundred million yen, but can exceed that under special circumstances. These government subsidies play an important role in improving the productivity of handicapped workers.

Model factories are sometimes a business outgrowth of a workshop organization or a subsidiary of a big business. Big businesses have some incentives to establish a model factory because handicapped workers in excess of the 50% minimum requirement can be counted towards their statutorial employment quotas for handicapped individuals. Big corporations may contract out certain operations suitable for handicapped workers to their model factories. This arrangement would increase the model factories' economic efficiency and also enhance their economic stability as they have built-in contractors. Thus, the economic potential of model factories, particularly when they are associated with big companies, seems to be promising.

Employment in Regular Private and Public Sectors

Model factories, though an interesting concept representing an unique relationship between business and government for promotion of employment of the handicapped, are small in number and still in the formative stage. Quantitatively most important mainstream employment has been employment in regular businesses and public agencies and governments. This mainstreaming effort began seriously with the passage of the Physically Handicapped Persons Employment Promotion Law in 1960 (6). The law established voluntary quotas to private and public sectors in the employment of the physically handicapped. The 1976 amendment of the law revised upward the quotas and made the quotas mandatory. The current quotas for public agencies specify that 1.8% to 1.9% of the total employees should be handicapped individuals. The quota for private business is 1.5% and for publicly owned business is 1.8%. These quotas reflect the proportion of physically disabled persons in the labor force to the total labor force in Japan. The quotas also take into account the differences in types of work and work environments among industries. Thus, the actual quotas are different from one industry to another. In order to encourage employment of the heavily disabled, the amendment specifies that severely handicapped workers are counted twice in fulfilling the quotas. The quotas are applied only to the physically handicapped and they have not been extended to the mentally disabled.

The law stipulates that large employers with more than 300 workers are required to contribute ¥30,000 per month per handicapped worker short of their quota. Small businesses are exempt from this requirement because they have shown better records in employment of handicapped persons (see Table I). The law established the Employment Promotion Projects Corporation to administer the funds thus collected. The funds are used as cash grants to employers hiring handicapped workers in excess of their quotas and capital grants to

demonstration projects (e.g., model factories).

ECONOMIC EVALUATION OF THE PHYSICALLY HANDICAPPED PERSONS EMPLOYMENT PROMOTION LAW

Table I shows the number of handicapped workers and their proportions to able-bodied employees in private businesses, public businesses, and public agencies in 1978.[5] The employment situation of private companies were further broken down by the size of companies in terms of the number of employees. In general, small companies hire a higher percent of handicapped workers than large companies do. The firms having 67 to 99 workers showed that 1.68% of their workers are handicapped. The ratio steadily declines as the size of firm increases. For the firms employing more than 1,000 workers, only .83% of their employees are handicapped. The noncompliance rates are also higher for large firms. The differences in the employment records between small and large businesses may be explained by the following hypotheses: (1) handicapped workers are frequently considered a secondary work force; large firms with more resources recruit from the primary workforce and leave the secondary workforce to small firms; (2) employment of the handicapped requires a great deal of commitment by management and requires managerial flexibility and discretion to meet the special needs of handicapped workers. It may be easier to find such a quality in small businesses; and small business may be in a better position to implement rehabilitation technology and to cooperate with government in solving the problems of employing the handicapped; (3) large firms may be concentrated in industries where the quotas are set low.

The differences in the employment records between small and big businesses may decline as time passes. The data presented are those of 1978 which was only two years after the amendment of the law. Thus, the 1978 records may not fully reflect the impact of the law. However, it seems impractical to expect that the differences will significantly diminish. Under inflationary circumstances, the nominal cost of employing handicapped individuals will increase, but the real cost of the levy will decline since the levy is specified in nominal money. Thus many big firms may find it cheaper in real money to pay the levy rather than to acquire resources to hire handicapped workers. Also, to the extent the second hypothesis stated above to explain the employment difference is valid, the levy alone cannot achieve the quotas.

Reducing the difference in employment ratios between small and large firms itself, however, is neither important nor desirable. The levy provides a flexibility in the otherwise rigid quota system. Companies will compare the levy with the cost of employing the handicapped and will decide whether or not to use the levy. In some cases, companies may have to rely on the levy payment because of mismatches between employers and handicapped workers over geographic locations and occupations. It is the average proportion of handicapped workers across industries [6] and the size of business that is important. As long as the average employment proportion becomes 1.5% for private businesses, the distributions of handicapped workers over individual industries and companies are not relevant. In fact, the levy and grant system of the law encourages an efficient allocation of resources by transferring resources from inefficient employers of the handicapped (e.g., large firms) to more efficient employers (e.g., small firms).

Will the law effectively increase the average employment ratio of handicapped workers? It primarily depends on the amount of the levy and grant. The levy and grant should approximate the actual cost of employing handicapped workers. The annual levy per worker is ¥360,000, while the survey of ten model factories shows that the additional cost of employing a handicapped worker was ¥760,000, of which ¥486,000 was production related costs and ¥274,000 was nonproduction related costs. Thus, the amount of the levy seems not to be enough to cover even the production related costs. Also, the actual average employment ratio of 1.11% for private businesses was substantially short of the target ratio of 1.5%. This may be another indication that the amount of the levy is not enough to increase the average ratio over time.

In the United States, mainstream employment of the handicapped is encouraged through the affirmative action programs authorized by the Civil Rights Act of 1964 and the Rehabilitation Act of 1973. The affirmative action programs basically advocate equal opportunities and attempt to reduce anti-discriminatory practices in employment. Compared to the affirmative programs, the Japanese law is a more specific and more powerful way of achieving the social goal in employment of the handicapped.

Finally, what is the prospect for a similar quota system in the United States? Most observers would agree that the prospect is not promising. There are many economically disadvantaged minority groups in the U.S. who seek their own economic gain. The quota system for the handicapped may invite similar claims by other minorities, unless the differences between the handicapped and other minorities in employment are clearly pointed out and accepted by the society. The minimum wage laws prevent the firms in the U.S. from hiring low productive severely handicapped workers who cannot earn the minimum wage. Thus, the quota cannot be applied to these workers, and any quota system excluding them would not be meaningful.

TABLE I

EMPLOYMENT OF THE PHYSICALLY HANDICAPPED IN FIRMS BY SIZE AND NONPROFIT ORGANIZATIONS, 1978

Organization	Number of Organizations	Total Employees	Total Handicapped Workers	Employment Ratio of Handicapped (in percent)	% Not Achieved Quota
Private Businesses (1.5% quota)					
67[a]-99 employees	8,514	688,147	11,571	1.68	43.8
100-299	18,720	2,824,296	42,193	1.49	43.5
300-499	3,467	1,158,938	13,819	1.19	56.4
500-999	2,310	1,379,698	14,357	1.04	65.9
1000 and over	1,728	5,385,823	44,553	.83	79.5
SUBTOTAL	34,739	11,436,902	126,496	1.11	47.9
Nonprofit Organizations (1.8%)	98	72,216	868	1.20	66.3
Public Agencies[b] (1.9%)	3,790	2,203,306	37,229	1.84	NA
Public Agencies[c] (1.8%)	251	746,464	13,840	1.85	NA
GRAND TOTAL	38,878	14,458,893	178,433	1.23	

Source: Japan Ministry of Labor, A Survey of Employment Status of the Physically Handicapped, Tokyo, 1978.

[a]Firms smaller than 67 workers are not covered by the law.
[b]Nonphysical production organizations, e.g., diet, courts, government ministries.
[c]Physical production organizations, e.g., post offices, government mint, printing office, forestry.

FOOTNOTES

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[2]Sheltered employment in the U.S. would imply employment in a sheltered workshop which is exempt from the minimum wage law (the Fair Labor Standards Act). Sheltered employment here implies employment outside the competitive labor market.

[3]Nonprofit status is concerned with the disposal of profits, not with the amount of profits or the methods of obtaining profits. Thus these nonprofit organizations may not be different from regular businesses in their pursuit for profits.

[4]The average exchange rate between the Japanese yen and the U.S. dollar was ¥216.53 per one U.S. dollar in the first three-quarters of 1979. International Monetary Funds, International Financial Statistics, Vol. 32, No. 12 (December 1979).

[5]The actual head count of handicapped workers should be less than the number shown in the table because of the double counting of the severely handicapped.

[6]Note that the law already allows for the differences in types of work in setting up the employment quotas for industries.

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ECONOMIC DISINCENTIVES TO REHABILITATION

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ABSTRACT

The economic incentives and disincentives for the employment of the handicapped are viewed from the standpoint of the handicapped individual, the government, the employer, and the service facility. Balance sheets (income versus expenses) are developed for the client and the government for various income levels. These balance sheets are subsequently utilized to develop continuous curves of net disposable income for the client, and economic incentives and disincentives for initial employment and advancement. A short explanation of the public assistance programs that affect the employment of the handicapped is also given.

INTRODUCTION

The approach towards the handicapped population in the United States is shifting from one of dependent care to independent lifestyle. One of the principal means for a handicapped individual to achieve this independence is through successful and meaningful employment (1,2). This paper will examine the effects of the employment of the handicapped in terms of economic incentives and disincentives for the four primary parties affected: 1) the handicapped individual, 2) the government, 3) the employer, and 4) the service facility. The current public assistance programs will be explained and the effects these have on the various affected parties will be examined for different levels of employment. All the data presented will be relative to a single adult living in section 8 type housing in Kansas during 1979. Also all cost figures will be expressed in annual amounts.

Several previous studies have examined disincentives to employment from various viewpoints (6). Walls et al (5) utilized preference curves to explain why an individual receiving a guaranteed income may not be motivated to work. A study performed by the University of Alabama Research and Training Center concluded that nonbeneficiaries of public assistance were more likely to become rehabilitated than beneficiaries. Treitel (4) showed that the number of dependent children, predisability earnings, and level of benefit were significant variables in recovery of disability insurance recipients.

PUBLIC ASSISTANCE PROGRAMS

Handicapped individuals may be eligible for a variety of public assistance programs. Some of the programs are specific for handicapped individuals while others provide assistance due to their poverty level income. A brief description of the major programs are described below.

- SSI(Supplemental Security Income) pays a maximum amount of $2498.40 per year for an individual with an annual income less than $1020. For every dollar earned above $1020, the individual receives 50 cents less of SSI up to a maximum income of $3360.00 (the substantial gainful activity level), at which time the individual loses eligibility. Eligibility for a handicapped client requires a disabling handicap and a low income.
- Section 8 Housing pays for housing and utility expenses in excess of 25% of the individual's income. Qualification for a handicapped client is an income less than $9300.
- Medicaid provides medical coverage for those that are eligible for SSI.
- Food stamps assist qualified individuals with food costs. The maximum annual benefit is $732 for an individual with no income (including SSI). This amount decreases to $120 per year when the income reaches about $600 and remains there until the individual's income exceeds $5012 at which time food stamp benefits are discontinued.
- Title XIX provides aide and attendant care for handicapped individuals requiring this care. Income qualifications are the same as SSI eligibility.
- Title XX provides training, counseling, physical therapy, and transportation services. Qualifications require being handicapped with income less than $6700.

BALANCE SHEETS

In trying to assess the economic incentives and disincentives associated with the employment of the handicapped, it is useful to develop "balance sheets" of income and expenses of the individual worker and the federal government for various earned income levels. Tables 1, 2, and 3 present balance sheet data for $0, $3300, and $6300 annual income levels respectively. A handicapped individual eligible for the programs described in the previous section who has no earned income would have $128.28 "net disposable income" after fixed expenses. The individual would receive $2498.40 from SSI and $234.48 for food stamps.

Assuming they lived in a "Timbers" type housing (a special housing project in Wichita, Kansas, for handicapped individuals), then the total annual housing expense is $3852 of which the individual pays 25% of the SSI payment. Also at The Timbers if food preparation is necessary, the cost is $150 per month or $1800 annually. Finally, the cost of a telephone which is connected to the communication system at Timbers is $180 annually. Thus the income equals $2732.88 and the expenses are $2604.60. The government expense for the SSI, section 8 housing, medicaid, and food stamps is $7160.27.

TABLE 1
$0 Salary Balance Sheet

Client			
Income		Expenses	
SSI	$2498.40	Housing	$ 624.60
Food Stamps	234.48	Food	1800.00
	$2732.88	Other Fixed	180.00
			$2604.60

Net Disposable Income = $128.28

Federal Government			
Income		Expenses	
		SSI	$2498.40
		Housing	3227.40
		Medicaid	1200.00
		Food Stamps	234.48
			$7160.27

TABLE 2
$3300/yr. Balance Sheet

Client			
Income		Expenses	
SSI	$1358.40	Housing	$1164.60
Salary	3300.00	Federal Taxes	14.00
Food Stamps	120.00	State Taxes	16.72
	$4778.40	Social Security	199.65
		Food	1800.00
		Other Fixed	180.00
			$3374.97

Net Disposable Income = $1403.43

Federal Government			
Income		Expenses	
Taxes	$ 14.00	SSI	$1358.40
Social Security	399.30	Housing	2687.40
	$413.30	Medicaid	1200.00
		Food Stamps	120.00
			$5365.80

Net Expenses = $4952.50

TABLE 3
$6300/yr. Balance Sheet

Client			
Income		Expenses	
Salary	$6300.00	Housing	$1575.00
	$6300.00	Federal Taxes	484.50
		State Taxes	89.62
		Social Security	381.15
		Food	1800.00
		Other Fixed	180.00
			$4510.25

Net Disposable Income = $1789.75

Federal Government			
Income		Expenses	
Taxes	$ 484.50	Housing	$2277.00
Social Security	762.30		
	$1246.80		$2277.00

Net Expenses = $1030.20

When the handicapped individual assumes a job at $3300/year, the SSI payment is reduced, the housing payment is increased, and now federal and state taxes and social security must be paid. The net disposable income has increased to $1403.43. The federal government associated net expense is reduced to $4952.50 because the worker is now paying taxes and social security (the employer matches the social security) and there are reduced SSI and housing payment.

If the handicapped worker is employed full-time at $3.03/hour, slightly above the minimum wage ($2.90/hour in 1979), then the annual salary is $6300 which eliminates SSI and consequently medicaid and Title XIX. The net disposable income is $1789.75 and the net federal government expense is $1030.20. This worker was assumed to be employed by a company that pays for the worker's health insurance. If employed by a company without this benefit, then personally financed coverage is necessary which in turn will reduce the net disposable income.

Figure 1 shows net disposable income as a function of yearly salary. The points at which a major loss of eligibility occur are also depicted on the figure. While in general there appears to be incentives for employment, even though slight, there are also income increases that would result in less net disposable income which are definitely disincentives to employment. For example, a client with an opportunity for a job with an annual income between $3400 and $5300 will receive less than for one at $3300.

Some of the more severely handicapped individuals require assistance in dressing, toileting, feeding, and housekeeping, Costs of these services for a typical client are shown in Table 4. If the client qualifies for Title XIX benefits, then these expenses are covered. However if they do not qualify, their annual income would need to be in excess of $15,000 in order to meet the expenses associated with the disability. This level of income is unreasonable for many severely handicapped individuals, which may result in them electing not to be employed at a level above $3360.

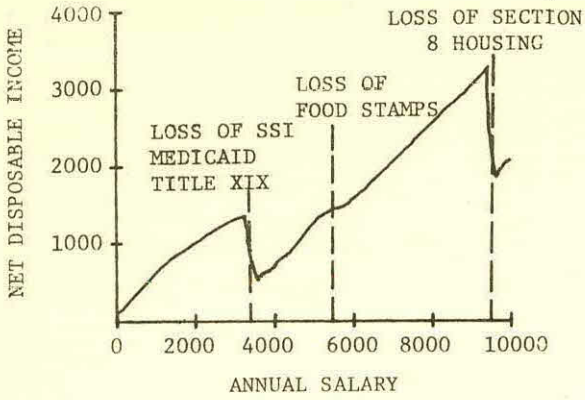

FIGURE 1. NET DISPOSABLE INCOME

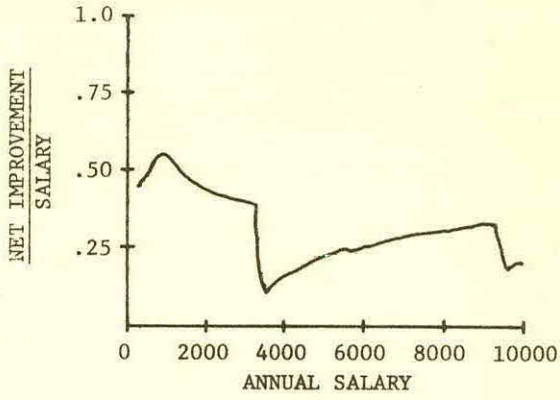

FIGURE 2. EMPLOYEE INCENTIVE FOR INITIAL EMPLOYMENT

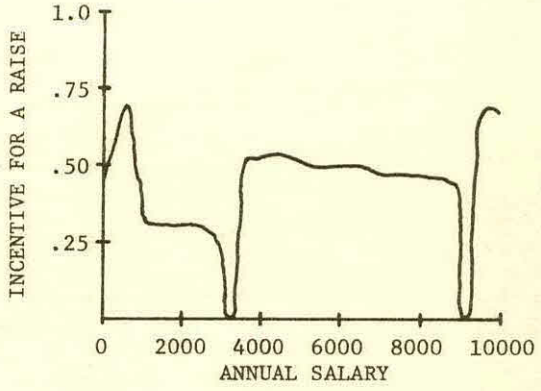

FIGURE 3. EMPLOYEE INCENTIVE FOR A $25/MONTH RAISE

TABLE 4
Expenses Associated With Disability

Housekeeping Aid	$ 720.00
Aides and Attendants	4776.00
	$5496.00

ECONOMIC INCENTIVES FOR EMPLOYMENT

There are several ways of expressing incentives for employment for the handicapped (3). One consideration is the incentive to accept initial employment where another is the incentive associated with advancement after employment. Since the handicapped individual has an income without employment, it is useful to calculate the proportion of salary that represents an increase in his net disposable income. Figure 2 shows the incentive calculated in this manner. For example, the individual who becomes employed at a salary of $3300 increases his net disposable income by $1275.15 or 38.6% of the $3300 salary. The ratio decreases to approximately 12% at the $3600 annual salary level; i.e., for every dollar earned, there is only an increase in net disposable income of $.12. The incentive a client has to increase their salary by $25 a month ($300 a year) is depicted in Figure 3. There are two points on the curve that represent negative incentives for a raise, at $3300 and $9300. Between $1200 and $3000 salary, the incentive is approximately 30% which can be interpreted as a $7.50 increase in net disposable income. The able-bodied worker's associated incentive is approximately 87%, nearly three times the incentive level of the handicapped.

GOVERNMENT ECONOMIC INCENTIVES FOR EMPLOYMENT OF THE HANDICAPPED

While there are only nominal incentives, and sometimes disincentives, for the handicapped individuals to employment, there are substantial incentives to the federal government as evidenced by Figures 4 and 5. Figure 4 shows the expense (income) to the government when the client is employed at various levels. The "out of pocket" costs to the federal government is approximately $7200 if the client is not employed and decreases as employment level increases. The "break even" point is approximately $8000 annual salary; that is where the expense to the government is balanced by income (taxes and social security contributions). The incentive of the government for the initial employment of a handicapped individual is shown in Figure 5 in terms of the ratio of decrease in government expense to salary level of the client. At $3600, the expense to the government is $2578.90 or $4581.37 less than the expense at the $0 salary level. Thus the government savings is greater than the entire salary that the client makes, this results in a 127% incentive to the government versus the 12% incentive that the client has.

ECONOMIC INCENTIVES FOR EMPLOYERS

Currently the only federal government program that economically encourages employers to hire the handicapped is "Targeted Jobs Tax Credit". This program allows tax credits in the first two years of employment of a certified client. Since the program provides only short term incentives to the employer, it is doubtful that it will successfully influence employers to place handicapped workers.

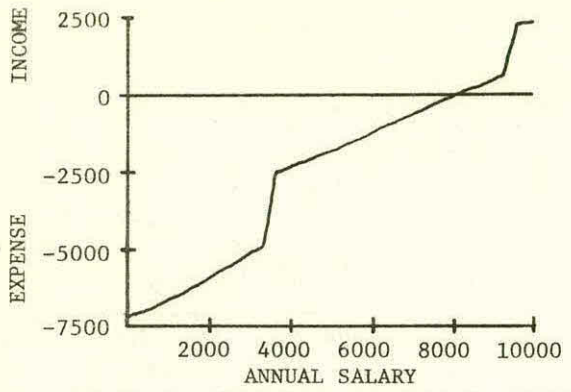

FIGURE 4. GOVERNMENT EXPENSE AND INCOME

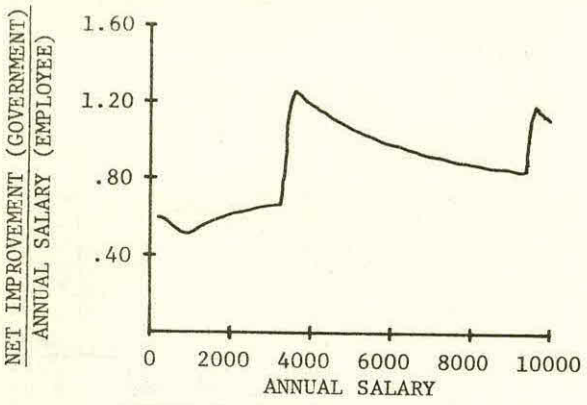

FIGURE 5. GOVERNMENT INCENTIVE FOR EMPLOYEE INITIAL EMPLOYMENT

SERVICE FACILITY DISINCENTIVES

One of the main objectives of Cerebral Palsy of Kansas is to provide an independent lifestyle for its clients. To realize this goal, Center Industries Corporation was established as an employment center and Timbers as a housing project for the handicapped. There exists a conflicting situation for clients who are employed and require aide and attendant care. In order for the clients to become independent, employment is necessary, but if the employment results in loss of Title XIX eligibility, then the service facility has no feasible way of collecting for the services provided.

SUMMARY

The present assistance programs provide nominal incentives and sometime disincentives for the handicapped to become employed, while the federal government is provided with very good incentives. With the dollars currently being expended, equitable programs could be devised that would provide adequate incentives to all participating parties. There are several legislative programs before the U.S. Congress that will modify some of the existing programs, but there is not a comprehensive package that is being considered. For handicapped individuals to move toward independence, productive employment is necessary; this means there will need to be positive incentives provided.

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K STREET MALL PROJECT
(EMERGENCY EGRESS SYSTEM)

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The California State Department of Rehabilitation, with its major central office located in a renovated and remodeled building at 830 "K" Street Mall in Sacramento, experienced a desperate need for and sponsored the development and implementation of the Project described herein. It is a unique, integrated fire protection system which provides for adequate emergency egress for all members of the Department, not only those in the wheelchairs, but those required to remain and help others in the wheelchairs and those who might have been hampered from evacuating the building in the event of a fire. The heart of the total building system is an automatically deploying flexible fire barrier which protects the elevator shaft and creates a refuge area on each involved floor.

BACKGROUND

Sections 501 through 504 of the 1973 and 1978 revised Rehabilitation Act mandate non-discriminatory action by an employer on the basis of disability of an individual. In addition, it mandated the removal of all access barriers for the disabled so that, in effect, there would be no handicapping as far as gaining entrance to a job. In reality, this "sword" is double edged, that is, once access is gained to the building for employment or any other purpose, and that access is to floors other than grade level, egress becomes a problem of greater magnitude than access for employers.

Inability to provide egress from or find refuge in a building for a person during an emergency, carries great impact from the standpoint of an employer's personal liability. Considerable concern is being expressed by employers naming this liability as a principal reason for not hiring handicapped people.

APPROACH

As experience was gained in dealing with the various constraints and aspects of the problem in meeting the needs of disabled people, it became very clear that many needs of the population as a whole have been neglected or ignored in the general body of data available for building design, fire supression system design, and emergency planning in general. Once into the investigation and definition of the problem the need for a much more global view for the Project became clear.

Evaluation of the specific needs of a limited population (i.e., the disabled) was broadened to correctly view the needs of the entire population. This is entirely appropriate since it is within the entire population that the Rehabilitation Act mandates the disabled be involved. And it is with the needs of the entire population that some of the largest undefined system constraints exist in trying to define the emergency egress problem for the disabled as described below under Rationale for Systems Objectives.

METHODOLOGY

Through the mandate and support of the Department's Director, Edward V. Roberts, himself a severely disabled quadriplegic, William H. Webster, and Clarence L. Nicodemus conducted limited investigations involving possible alternatives to emergency egress system mechanisms for the "K" Street Mall Building. In addition, an interagency agreement contract was initiated with the Departments of Mechanical Engineering, College of Engineering, and Physical Medical and Rehabilitation, School of Medicine, University of California at Davis in order to establish a "brain trust" group to more thoroughly evaluate problem constraints and alternative solutions. The focus for these investigations was constantly maintained on the specific retrofit needs for the "K" Street Mall Building in order that an immediate solution be found.

SYSTEM OBJECTIVES

The majority of the Project was conducted then in this manner: the work statement presented to the UC Davis group was in the form of system performance objectives for whatever type of emergency egress system that was to be involved. These objectives are as follows:

* be useable by all human beings in the building at any given time,

* be fool-proof, simple to understand and to operate,
* be storable for long periods of time without affecting its operational function,
* not obstruct normal and emergency pathways as specified by codes,
* provide immediate (four minutes or less) safe refuge for all persons, including those with limited mobility,
* facilitate or enhance existing emergency systems (i.e., alarms, elevators, stairwells, etc.)
* handle wheelchair and all attached wheelchair equipment as well as the wheelchair occupant,
* be operable by a paraplegic alone, but may permit an attendant for the needs of a quadriplegic,
* be operable by the blind, the deaf, and the mentally disabled without assistance,
* accommodate seeing-eye dogs and other assistive devices required for mobility,
* have minimum dollar impact on implementation in existing buildings,
* require no adaptation or attachments to wheelchairs (A visitor in the building must be accommodated equally as well as a regular employee in a wheelchair.),
* not require the ambulatory to remain in the building to assist the nonambulatory (except in the case of the quadriplegic),
* not rely on any external power sources for its junction.

These performance objectives are considered by the Project to be minimal, but certainly not exhaustive. As work continues, these will be and have been to some extent modified, mitigated, or otherwise met.

RATIONALE FOR SYSTEM PERFORMANCE OBJECTIVES

The justification for establishing these performance objectives comes from an analysis of the make up of the total population for any given day in the normal business of the Department of Rehabilitation. Theoretically, this population should reflect generally the same make up in all public agencies (or private business for that matter) exercising the proper hiring practices.

FUNCTIONAL LIMITATIONS OF THE GENERAL POPULATION

The nation's population is made up of a multitude of people all with different capabilities and disabilities, handicapped only where there are barriers placed in pursuit of needs. Barriers can be of many variety and description, and need not be hardware or architectural. They can be social (i.e., language), economic (excess costs), perceptual (non-understood symbology), sensorial (not able to be seen, heard or felt), attitudinal (prejudice), informational (misrepresentation), etc.

All of these barriers exist in one form or other and must be considered in the planning effort for a total emergency egress system. These barriers are not limited to the disabled population, and are handicapping to able bodied individuals. An excellent example of a nondiscriminatory handicapping architectural barrier is the spherically shaped door knob. Obviously, this shape of door knob handicaps a double upper extremity amputee because there are no functional hands and fingers available to squeeze and rotate the knob. But the same can be said for a fireman with burned hands, a housewife carrying a child, an arthritic elderly person, a stroke victim, a child with grease or water covered hands, etc. Thus, the need to consider the total population as a spectrum of needs, shared by all members, either on a temporary or permanent basis, is a much more realistic design approach of any system, but especially an emergency egress system.

The following chart lists the major (but not all) functional considerations to be taken into account in designing an emergency egress system. These functional considerations are related to specific disabilities and then to project constraints which will hopefully mitigate any handicaps that might evolve for those using the emergency egress system.

Functional Considerations	Lack Of Ability To:	System Constraints
Mobility	Move or propel oneself horizontally or vertically.	Safe area refuge must be close by or quickly transported in order to meet a four minute "rule of thumb" smoke environment survival time requirements.
Manipulation	Physically operate controls of various kinds.	All controls must be extremely simple and operable by gross motor activities only; all controls within 40 inches above the floor.
Hearing	Detect alarms and follow audible instructions.	Provide visual and tactile alarm systems.
Sight	Detect and follow visual alarms and instructions.	Provide audible and tactile alarms and instructions.
Perception	Mentally understand controls and instructions (through differing languages, lack of education, presence of learning disabilities,	Provide nonlanguage specific instruction and control directions, use clearly understandable symbolic instructions, use

Perception (cont.)	presence of developmental disability or loss of other mental integrative capability).	psychologically appropriate con-instructions (in an emergency a person wants to push on a door to get out, not pull on it).

It would be well here to review a list of "handicapped" individuals which the reader may find surprising when thought of in this context, but which must be considered in planning:

Children and Visitors

Children and visitors fall into the category of having perceptual difficulties in an emergency in that they are unfamiliar with surroundings. In the case with the former, they may not understand many of the more obvious instructions.

Pregnant Women, the Arthritic and the Aged

Pregnant women, the arthritic, and the aged all fall generally into the category of mobility considerations and that in an emergency situation, emotions and psychological effects may render them unable to adequately egress either horizontally or vertically.

"Totally Able Bodied" Persons

During an emergency of any sort totally able bodied persons may become victims of injury which will render them functionally disabled in any one of these five categories.

MAJOR RESULTS OF THE PROJECT

Besides the generally beneficial result for the Department of having explored and becoming involved with the need for emergency egress planning, the Project has developed several very specific results:

Problem Definition

System constraint analysis. As explored briefly above, the Project has brought into focus many of the primary concerns in developing a system which must adequately function for all members of a given building population. These are the kind of considerations that need to be assessed for each building site in terms of retrofit for existing buildings and in terms of new design criteria for future buildings. The use of these analyses are the first step in a realistic planning of emergency egress needs.

In the conduct of the analysis and expecially in comparison of objectives with available data, a great number of deficits were found: inadequate code specifications to meet the needs of the total population; inconsistant code requirements; total lack of data for human performance in an emergency or fire environment; contradictory data in codes as compared to research (for example, sprinkler secondary effects on life safety, smoke removal capabilities, toxic gas generation from new materials) and many others.

Model

The bottom line of this effort is the first modeling of a standardized design approach for self analysis and evaluation by other public agencies for emergency egress and life safety planning.

Solution

After considerable exploration and brainstorming of alternatives for a total emergency egress system, it became clear the problem naturally breaks separately into two uniquely identifiable, yet fully integrated parts:

Safe Refuge Area on Each Floor Level. This meets the requirement for rapid physical removal from the immediate threat to life safety. For the "K" Street Mall Building, a system of automatic operable, flexible fire rated smoke and gas barriers were utilized to form safe areas on each floor and to provide an integrated building-wide vertical egress protection system. Installation was completed on October 1, 1979.

Vertical Egress Enhancement. With or without a safe refuge area, eventual evacuation of a building involved in fire emergency will be necessary; the refuge area merely "buys" time. The realistic alternative avenues for vertical egress are the elevators and stairwells, both under the "protection" of the refuge area. Both require enhancement to become available to the total population before the system will be complete.

SYSTEM DESCRIPTION

The system consists of a safe area constructed on each floor to take maximum advantage of existing egress facilities (stairways and elevators). The location of these safe areas were carefully selected in close cooperation with the State Fire Marshal's Office only after an analysis of the building on a floor by floor basis was made to avoid areas of likely fire potential and to include most direct and secure avenues of egress.

Dual Purpose Placement

The placement of these safe areas was not only for the immediate refuge they provide on each floor, but also the overall protection of the elevators for use by personnel for vertical egress out of the building or by fire supression and rescue people entering the building during the fire.

Egress/Rescue. If power is available and the elevator can be used as an evacuation means, then it is protected from contamination by smoke, fire and toxic gases. Therefore, not only has safe refuge been provided, but a major step toward providing a protected entry way for fire supression personnel has also been accomplished since the largest stairwell in the building also lies in the safe area.

Property Protection. This accomplishes another beneficial aspect of this design, namely that it prevents the transfer of fire and its precursor, hot ignition gases, from leaping from floor to floor vertically through the elevator shaft. A major step has been taken in protecting the property damage of the building through blocking a common avenue of fire transmittal from lower to upper floors.

Flexible Fire Barrier

The key elements restricting the construction of a solid wall refuge, 40 foot long, were the critical need for useable floor space and the interruption of the HVAC air circulation for the lower level (below grade) and second floor open areas.

Unobstructive. Because of these key elements, the deployable flexible fire barrier system was selected. Since no "wall" exists until the barrier is activated by a signal from a smoke detector or other device, both floor space and air flow were not obstructed.

Accessible/Multipass. It's touch activated opening mechanism has been specifically modified into a "crash" panel for use by a person in a wheelchair, thus making the flexible barrier a multipass, accessible port into the refuge area.

Vertical Egress System

Recognizing that the safe area does not meet the need to evacuate a building for a given fire situation, alternatives for vertical egress and evacuation of the building were considered. Two alternatives centering on the existing stairwells and existing elevator system were selected for possible enhancement in this specific building. The possible enhancement of the elevator system included the addition of auxiliary electrical power provided by an automatic controlled generator system.

Stairwell Enhancement Concept. In place of the elevator, an alternative system was proposed to make use of the stairwell already in place. Considerable engineering design effort was devoted to developing the feasibility of a gravity powered vertical egress assistance system installation.

The concept is currently in the prototype stage to test feasibility of its operation in the existing stairwell.

Rail System. Very basically, the proposed system consists of a continuous rail beginning at the upper floor landing area and terminating at grade level. The rail serves as a track on which transport modules will be attached.

Transport Modules. These modules are of lightweight frame design that are to be folded and stored in a convenient, nonobstructive manner in each landing. They are designed to be rapidly attached for use by an individual with limited mobility to decend the stairs in a controlled, safe, continuous fashion.

Controlled Decent. The system is designed to transport the entire wheelchair and person in a level attitude under controlled automatic braking with damping and partial manual override for additional speed control.

"Self" Operated. The system is designed so that the individual being transported can control decent himself, or in the case of a quadriplegic, whose upper limbs may be involved to the extent of not being able to manipulate this simple speed control (actually a brake), it can be operated by the attendant or anyone else willing to accompany the individual down the stairs.

Prototype. While the system undoubtedly has a great many difficulties yet to be worked out, its proposed design does meet all of that of a total emergency egress system. Its design and prototype model formed the subject of a M.S. Thesia in Mechanical Engineering by Mr. Kevin Allen, under the supervision of Professor J.M. Henderson, University of California at Davis and the authors.

CONCLUSION

Three major identifiable results have been achieved within the "K" Street Mall Project:

Standardized Approach

A first model for a standardized design for general building structures incorporating emergency refuge areas and life safety evacuation procedures has been developed.

Inplace System

Actual implementation of a fully integrated and automatic, retrofitted emergency egress system in a multistoried, leased public agency building brought about through cooperation of six public agencies and private building ownership has been accomplished.

Prototype Stairwell System

Prototype design of a retrofitted, stairwell in-dwelling, gravity powered, vertical evacuation assistance system as an augmentation or alternative to auxiliary powered elevators.

NEED FOR FURTHER WORK

Much additional effort needs to be conducted as a result of the work described above and as a matter of supplying an enormous amount of realistic design data for use by those who may be (or should be) involved in this process in the immediate future. This effort should be channeled along the following general pathways:

Improvement

Addition of alarms and strobed lights for the deaf; exit and safe area access delineation for low visibility; 2-way communications between safe areas and command center outside building; pressurization and/or fresh air flow into safe areas.

Testing

This project must confirm the functional viability of the implaced flexible fire barrier system. This will be done on site to the extent possible, but also in burn tests conducted in cooperation with the California State Fire Marshal's Office, Phil Favro, Director.

Investigations

Many areas of investigation must be pursued to improve future designs and supply data: elevator pressurization, realistic human performance/characteristics in a fire emergency, smoke removal equipment, requirements for highrise building or large open areas, effects on breathable environment of sprinkler interaction on fire and combustion by-products, etc.

Education

Many professional groups of people must be advised and made aware of the total building life safety needs of the population and to whom they can turn for information. These groups include public agencies, building designers (architects, engineers, etc.), fire marshals and chiefs, elevator and other equipment designers and manufacturers, the consumer (handicapped and able bodied) and those who develop and promulgate codes and standards.

The Project is dedicated to continue work in all of these areas as time and funds permit.

ACKNOWLEDGEMENTS

Many people and organizations have assisted in many different ways in order to allow this unique development to happen, among the major contributors are: Department of Rehabilitation, State of California - financial support and total commitment; Office of the State Architect - collaboration on space utilization; Won-Door Corporation, Salt Lake City, Utah - willingness to modify flexible fire barriers to meet project needs; U.S. Fire Administration - encouragement, interest and information; National Task Force on Life Safety and the Handicapped - encouragement to persist.

A PROPOSAL FOR A GRADUATE LEVEL PROGRAM FOR REHABILITATION SCIENTISTS

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ABSTRACT

It is intended through this paper to evolve a basic framework of a Master's level program for Rehabilitation Scientists to be implemented at a University. This paper is based on observation of the working of five clinics and four engineering units. These clinics and rehabilitation units are listed in Appendix I. The first part of this paper addresses itself to rehabilitation in general. It focuses on the technological input in rehabilitation to understand the role of the Rehabilitation Scientist. The second part brings out the salient features of the proposed program for Rehabilitation Scientists. A proposal for possible implementation of the program at a University is presented.

INTRODUCTION

A standard definition of rehabilitation is 'the restoration of maximum physical, mental and vocational capacities to disabled persons'. Along with this restoration process the patient learns to adapt to an alternate lifestyle that suits his functional capabilities. This training aspect makes rehabilitation unique. The application of scientific medicine is tempered with humanistic health care. Besides treatment and training, education forms an integral part of rehabilitation. Relatives and friends of the patient, the public and the patient need to be informed.

Rehabilitation draws from three major resources: medical, technological and counselling. The medical services include the physiological and the therapeutic aspects of medicine. Physicians with various specializations concentrate on the physiological aspects, whereas the nursing staff and the physiotherapists concentrate on therapy. Technological services rely on a staff of engineers, prosthetists, orthotists and occupational therapists. Social, psychological and vocational counsellors provide a link to the patient's family, friends and in some cases the employers. Each one of these professionals contributes to the overall process of rehabilitation of the patient. The very nature of rehabilitation requires cooperation and a complementary effort from the patient. It is a team effort.

All professionals involved in rehabilitation receive related education and training. The only exception is the engineer. He is, generally, trained in the traditional fields of engineering such as electrical or mechanical and is called upon to translate his skills to suit the rehabilitation needs. The nature of his job and the environment he has to work in, suggests that engineering skills alone might not be enough.

To better understand the role and the skills required of an engineer in a rehabilitation setting, the technological input to rehabilitation is examined in some detail.

TECHNOLOGICAL INPUT

It is agreed that all the efforts in a rehabilitation setting are directed towards helping the patient. Acceptance by the patient is the final barometer of success of any solution that materialises from these efforts. Some

other important considerations that affect the technological input to rehabilitation are discussed below.

Environmental Constraints

The engineer in a rehabilitation setting is exposed to a unique, often unfamiliar, people oriented environment. He has to understand the patient's needs and concerns not only to provide better solutions but to identify where the problem sources might be.

His approach to a problem differs, generally, from that of his colleagues. The time frame and the economic considerations are different; the working language is unfamiliar. For a good overall integration of efforts of the team members there is a need for familiarisation and appreciation of each others roles.

Present Status of Knowledge of the Disability

The effect of this factor on the technological solution to a rehabilitation problem is obvious. To identify and solve the problem, some knowledge about the physiological nature of the disability is essential; however, the human body is a set of interdependant subsystems. Isolated treatment of a problem is the exception rather than the rule.

This requires, for example, consideration of the patient's cardiovascular condition before prescribing a prosthesis. Also damage in a particular disability may be such that the treatment may not yield any benefits. Consider for example two cases of motor disfunction that require some form of orthosis. First in the case of instability during rotation of the knee joint in a young person, it is possible to provide a orthosis that will be helpful. But in the case of a cerebral palsy patient requiring a leg brace, damage to other faculties may be severe enough to overshadow any benefit that may be gained by the brace.

Physical & Functional Capacity of the Patient

The rehabilitation goals vary with the patient's physical and functional capacities. The aim in terms of rehabilitation will be different for a young person as compared to an old person, a working person as compared to a retired person. Not only do goals differ but the emphasis is different too. For example cosmesis is more important to a young person with upper limb amputation than say for a worker where the function is more important.

Further, any solution has to be based not only on the past history and the present needs and condition of the patient, but also on some sensible prediction of the future turn of events. For example in the case of an amputee due to vascular disorder one has to consider the possibility of the other limb being amputated at a later stage.

Present Status of Technology

It is not only the technical expertise that affects the solution but also how cheaply and readily that technology is made available. New products are introduced in the market everyday, but it takes time before it can be used routinely in a clinical setting. An adapted minicomputer system for the use of quadriplegics is a case in point. Also, different aspects of technology progress at a different pace. This asynchrony affects the rehabilitation product. The Swedish Arm where the electronics are far superior to the mechanical and control aspects of the limb, is a good example.

All of this assumes that the available technology will be used innovatively.

ROLE DEFINITION

The preceding discussion clearly indicates that though traditional engineering training may be an asset, these skills need to be augmented with other engineering and non-engineering skills. This calls for post graduate training and hence a well defined program. It is hoped that the proposed program will help meet these needs.

The analysis presented also suggests that not all the engineering skills are directly applicable to rehabilitation. It should be possible for students in other sciences like physics, kinesiology and from technical institutes to acquire additional appropriate skills to fulfill the requirements of this position. This

diversity of disciplines introduces the problem of terminology. It is therefore recommended that these people be called 'Rehabilitation Scientists'. The term rehabilitation describes their area of activity whereas the term scientists defines their role appropriately.

A rehabilitation scientist is basically a problem solver in a clinical environment. His work includes the following:

Servicing. This includes meeting the immediate needs of the patient and his colleagues on a day to day basis such as modifying existing equipment or apparatus or consultation.

Engineering. This may entail assessing and translating the needs of the patient into a rehabilitation apparatus, assessing and translating his own or his colleagues needs into say a information gathering device.

Research & Development. This may call for assessing the need for a new development and developing a new product.

Education. This may include demonstrating to his colleagues on how best to utilise his resources and educating the patient on the capability and limitation of the apparatus in use.

A framework for a graduate level program for training of Rehabilitation Scientists is presented in the following sections.

PROPOSAL FOR A GRADUATE LEVEL PROGRAM FOR REHABILITATION SCIENTISTS

The main objective of this program will be to train graduates from engineering and other sciences to fulfill their roles in a rehabilitation setting. Besides training these graduates, this program would offer a unique opportunity to other professionals in the technical resources pool to upgrade their skills. With increasing tendencies towards automation as in the field of prosthesis fitting, this retraining of professionals might become imperative.

This program should begin with a few interested students. Once these trained students are introduced to the rehabilitation system, their potential will be realised and any subsequent need for more graduates will be created. The program will have the flexibility to adjust to the needs of the system.

Entrance Requirements

Any student with a degree in engineering or sciences or in some cases with a technical diploma together with the motivation to enter this field should be eligible.

Structure & Timing

The program is proposed to consist of course work and 'hands on' experience in a rehabilitation setting. The course work is divided into four modules: medical, technological, social and psychological and internship. As the program will be open to graduates from various fields the timing will be set by the individuals needs and requirements. Each student will be required to have a fair grasp of all the course work in the first three modules before proceeding to the internship program. For students already working in the rehabilitation area the internship program will be modified to suit their needs.

Course Modules

The main objectives and basic considerations of the required coursework is presented. Since this program began with training of engineers in mind, a partial listing of courses (available in the B.C. lower mainland area) relevant to the medical course work is presented in Appendix II.

Medical Course Module. The main objectives of this course module will be to give a comprehensive view of how medical professionals treat various disabilities, to understand the medical aspects of the problem he has to deal with such as skin interface problem, physiology and the biomechanics and generally to acquaint himself with the terminologies he will encounter. The emphasis in the various courses will be different. For instance, the physiology course will deal with the interdependance of various subsystems and the constraints it puts on the solution to a rehabilitation problem.

Technological Course Module. This module will have courses from electrical, mechanical and chemical engineering. The electrical engineering courses will be on circuit design and applications, instrumentation, and computer programming. The mechanical engineering courses will include courses on strength of materials, design, mechanics and workshop training. The chemical engineering course will focus on biocompatibility of materials.

Social & Psychological Course Module. This will have psychology and sociology courses related to disabled people. The student should not only be made aware of the social and psychological needs of the disabled people, but also of the public attitudes towards disabled people and how it affects them.

Internship Module. Through the internship program the candidate should be exposed to as many rehabilitation facilities as possible. At the end of this program he should have completed a chosen project. The objectives of this 'hands on' experience is to demonstrate the practical applications of what the student has learned so far, to get acquainted with the environment, to understand how rehabilitation teams operate and realise how he fits in the scheme of things. He should be encouraged to correspond with other centers across the country and build up a comprehensive list of existing aids and solutions and pertinent literature on the subject.

Resources Required for the Implementation of the Program

The three major components required for effective implementation of this program are; financial resources, skill or faculty resources and facilities for the internship program.

Any geographical area where there is a medical and a engineering school should be able to offer the courses that follow the general outline proposed in this paper. This would greatly minimise the demand for new faculty members. Mainly personnel to coordinate the program will be required. It is assumed that any area that has a medical and a engineering school will have some reasonable rehabilitation facilities. If the students are encouraged to choose projects that meet their existing needs, they should be happy to provide their facilities for the students.

If the existing facilities are used optimally, the financial requirements to set up and run the program should not be hard to meet. Financial support for the students can be obtained in the form of research and teaching assistantships through their respective departments.

Conclusions

A rehabilitation scientist trained through this program will have the following skills:

a) Medical knowledge of various disabilities.
b) Knowledge of human body mechanics.
c) Familiarity with the existing aids and solutions available for various disabilities.
d) Knowledge of engineering principles applicable to rehabilitation.
e) Knowledge of materials and technology used currently in rehabilitation.
f) Familiarity with the roles of his colleagues.
g) Knowledge of social and psychological problems of the disabled population.
h) Administrative, managerial and communicative skills.

That there is a need for rehabilitation scientists was apparent from the clinics and rehabilitation units attended in the preparation of this paper. In Canada there is no formal program, even for rehabilitation engineering. Many centers provide on the job training on an adhoc basis. The recommendations presented in this program are in concurrence with the suggestions made in the 1976 report on rehabilitation engineering education by the United States Department of Health and Welfare and the Veterans Administration.

APPENDIX I

The following clinics and rehabilitation engineering units were visited during the course of preparation of this report.

a) Canadian Arthritic and Rheumatism Society Clinic, Vancouver, B.C.
b) Vancouver General Amputee Clinic, Vancouver, B.C.

c) Shaughnessy Hospital Clinic, Vancouver, B.C.
d) G.F. Strong Rehabilitation Center Clinic, Vancouver, B.C.
e) The Lady Hoare Rehabilitation Engineering and Therapy Unit, Chailey Heritage, Sussex, U.K.
f) Oxford Orthopaedic Engineering Centre, Oxford, England.
g) Dundee Limb Fitting Center, Dundee, Scotland.
h) Biomechanical Research and Development Unit, London, U.K.

APPENDIX II

The following is a partial listing of courses available in B.C. and relevant to the medical course work outlined in this paper.

HCE 508 - Seminar on Rehabilitation (Philosophy of care in rehabilitation medicine in hospital and community setting)
RM 202 - Practical and Applied Anatomy
RM 201 - Medicine & Surgery (Comprehensive view of diseases encountered in the clinics)
RM 429 - Rehabilitation Seminar (Occupational and Physio therapy overview)
OS 720 - Orthopaedic Basic Science Course.

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THE UTILIZATION OF UNDERGRADUATE RESEARCH TO ASSIST THE HANDICAPPED

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ABSTRACT

There exists in many colleges across the country a large untapped resource for research to assist the handicapped. This resource is the senior student enrolled in one of the many Engineering Technology programs. With modern technology, many students are quite capable, with proper faculty guidance and adequate funding support, of developing sophisticated devices to assist the handicapped. For the past four years at this campus of Purdue University, selected students have been engaged in such undergraduate research. Some past, present and future projects are described and the methods of funding explored. The development of the Purdue program for the utilization of undergraduate research to assist the handicapped is detailed to aid other institutions that are considering similar programs.

INTRODUCTION

The words "Julie" and "Hat" are simple, seemingly insignificant words. Yet for a special girl named Julie who recently spelled these words, and for others like her, they are a positive step forward, perhaps the forerunner of greater things to come in helping in the handicapped person communicate.

Julie, the victim of severe cerebral palsy, has no speech and her spastic movements have made her attempts to write or to use a typewriter impossible. Julie is seated facing a display board where there are letters of the alphabet, numbers and punctuation marks enclosed in marked squares. In each square is a light-emitting diode that glows red as the display scans through a complete sequence. Every so often, Julie tips her head slightly to the left and a teletype machine next to her types out a letter character. Soon Julie leans back and laughs. We look at the teletype paper and see that she has written "JULIE HAT." This is the first intelligible message that Julie has been able to communicate to anyone in her seventeen years of life. Julie has had cerebral palsy since birth. She cannot speak and has no predictable control over any part of her body except that she can tip her head slightly when she desires. She is confined to a wheel chair but attends special education classes in the public schools. Just how much she has learned is difficult to assess, since she can communicate with her teachers only with smiles, noises and by tipping her head. Now, for the first time, Julie has "spoken" to the outside world.

John, a senior electrical engineering technology student at Indiana University-Purdue University at Fort Wayne, has spent a year on his Senior Design Project designing this device for Julie. John's device scans through the ASCII code in the order of the most frequently used letters in a time-variable sequence. When Julie sees the letter she wants to type, she tips her head while wearing a special hat with a simple mercury switch. The result is a print-out on the teletype. The faculty advisor for this Senior Design Project had made John aware of the need for the device and also obtained a modest grant from a local philanthropic foundation. The teletype machine was renovated and donated by the General Telephone Company. The rest was pretty much up to John and Julie.

AN UNTAPPED RESOURCE FOR RESEARCH

The concept of Engineering Technology as differentiated from the traditional approach, is relatively new. Twenty years ago a few universities originated the technology curriculum which required less higher mathematics and concentrated more on the laboratory or "hands-on" approach to education. Since that time, hundreds of universities and colleges across the country have expanded the technology concept to such areas as Electrical Engineering Technology, Mechanical Engineering Technology, Civil Engineering Technology, Computer Technology, Architectural Technology, etc. Since there are only a few graduate schools for

the technologies, most universities require some form of Senior Design Project in the four-year program. Most of these technology students, having a practical orientation, opt to design and build a piece of equipment rather than perform the theoretical study which is typical of many graduate dissertations.

At this campus, which is one of the five Purdue state-wide system, the senior students have been free to select their own projects as long as they received faculty approval. The students are then responsible for funding their own projects Starting in 1976, the author, who also serves as an engineering consultant to the local Society for Crippled Children and Adults, encouraged selected senior students to design and build electronic and electromechanical devices that would assist crippled persons to relate to their environment and was able to obtain limited funding from local sources. By 1978, this program had grown to the point where it became expedient to obtain broader and longer-range funding, rather than to have to seek funding for each individual project. A proposal was submitted to the Easter Seal Research Foundation, which now funds a modest three-year grant for undergraduate research to assist handicapped persons.

Since the students are working on a required course for university credit, the student and faculty time involved on the projects are not charged to the grant. This means that all of the funds from the various grants can be expended for parts and supplies. Under this arrangement, a few thousand dollars can adequately cover five to ten projects at a time. Many suppliers are also generous in donating parts, which helps keep the cost of research down. And with today's technology, at least in the electronics field, a few inexpensive integrated circuits and microprocessor chips can perform many complex operations which would have been prohibitive in cost and time only ten years ago.

TYPES OF PROJECTS

In addition to the communications project described in the introduction, last year a senior in Mechanical Engineering Technology designed, built and demonstrated a rather unique wheel chair carrier. This carrier, which was not much larger than a folded wheel chair, was designed to be attached to a standard trailer-hitch on a passenger car. One side folded down upon opening and formed a ramp for rolling the chair in and out instead of having to lift the chair. The entire carrier was so pivoted on the hitch that it could easily be swung out of the way of the trunk of the car so that storage space was not blocked. With this clever device, even a woman could easily load and unload the chair with no lifting. Some handicapped persons, who drive their own special cars could even load and unload their own chairs.

Several years ago, an orthopedic surgeon suggested to the author the idea of designing an electronic device that would replace the customary weights and pulleys used in hospitals for orthopedic purposes. Most handicapped persons are all too familiar with this therapy. The author, at that time, was frankly not enthused over the need to replace something as simple in concept and seemingly as inexpensive as weights hanging on a rope, with a complex electromechanical device. This lack of enthusiasm was based upon the ignorance of the many problems associated with this type of treatment. The weights and pulleys are awkward, unreliable and potentially dangerous. Most hospitals employ teams of men that do little else than put on and take off weights as directed by the nurse or physician. Thus what appears to be an inexpensive device, when attended around the clock, is in reality expensive to operate. The system is also far from being portable.

After consulting with health care professionals about such things as patient movement while in traction, moving the patient while bedfast to other areas in the hospital, visitor manipulation of the equipment, sanitation and safety regulations, it was found that there were more than a few problems associated with the traditional weights and pulley method. This system is too bulky to allow for accessibility to elevators, necessitating the partial dismantlement of the traction system when a bedfast patient is transported between floors. Visitors, amazingly enough, often remove weights from the hanger in an effort to alleviate the patient's pain. Visitors, doctors and other hospital personnel inadvertently bump into the weights, causing the patient pain because of the impulse transmitted to the bones under traction. The weights must be placed on or removed from the hanger in a very gentle manner. And the repeated set-up and dismantling of the weight system requires considerable effort requiring personnel employed for this sole purpose.

As a result of these preliminary investigations, we now have at Purdue, two Electrical Engineering Technology students and one Mechanical Engineering Technology student designing and building an electronic servo-controlled device to replace the weights. This new device will feature a single knob setting and a digital read-out of the desired number of pounds of pull. It will operate from a

self-contained batter, with charger, for portability and emergency use. It is designed to automatically compensate for any movements on the part of the patient. Nurses will be able to operate the device, since there are no weights to lift, and a locking switch will be incorporated to limit operation to authorized personnel.

Another area in which research by undergraduates is being performed is that of assistance to the handicapped in the classroom. In just the past year or so, the Federal government has mandated the public school systems across the country to assume the responsibility for the education of handicapped children. Although the schools have accepted this responsibility, it is the considered opinion of the author that the schools, through no fault of their own, are presently ill equipped to accomplish this task. Traditional methods of instruction are hardly viable for the severely handicapped. There is no way that a handicapped child that does not have control over his or her arms and hands can learn to perform many of the tasks taken for granted in the educational process traditionally followed. Yet most schools are presently trying to follow the same methods for all students, handicapped or not. In response to this challenge, several Purdue senior students are presently designing equipment that will enable a handicapped child to operate a microcomputer. The microcomputer, which is relatively inexpensive, can be operated in several modes. In one mode it can be utilized in a manner similar to the equipment described in the introduction, that is, as an electric typewriter with a display and/or a printed output. In the second mode, the microcomputer can be used as a calculator to allow the handicapped student to perform mathematics lessons from simple arithmetic to algebra and trigonometry. In the third mode the microcomputer can be programmed as a teaching/testing machine to drill and test the student in a wide range of mathematics problems, multiple-choice and true-false questions.

A microprocessor based scanning device is presently being designed by several senior students that will allow a handicapped child to control the keyboard input to the microcomputer through whatever means are available to a particular individual, eg., a blowtube switch or any controllable single movement. Yet another student is designing a single-finger control that will allow a handicapped person that has full control over one finger to operate a microcomputer. Still in the planning stage is a system to give blind persons a synthetic voice readout of what they are typing on the keyboard of a microcomputer. Much of the software for this type of device has already been developed on a large federal grant at the University of Illinois. This type of device would be particularly useful to blind or visually impaired college students majoring in computer science or mathematics.

Still another Purdue student is designing, for his Senior Design Project, a device utilizing digitally controlled synthetic speech to call out floor numbers and other useful information for the benefit of the blind using the elevator at the local Historical Museum. This project is being funded locally by the Historical Society.

DEVELOPMENT OF A RESEARCH PROGRAM TO ASSIST THE HANDICAPPED

Involving senior students in research programs for the assistance of the handicapped benefits not only the handicapped but also serves to raise the social conscienceness of the university students involved. Perhaps the largest single problem to be faced in intiating a research program to utilize senior students to assist the handicapped through technology is the lack of communication among those persons, such as faculty members, that are experts in the technologies, and those persons closest to the problems of the handicapped, such as physicians, teachers, therapists, parents and the handicapped themselves. Since there is little in the literature that bridges both areas, about the best the technical person can do is to relate to as many of the aforementioned as possible. Here at the Fort Wayne campus of Purdue we formed a Faculty Research Committee for Equipment for the Handicapped. This committee assists in the selection of projects, initiates new projects, serves as a professional resource for the students, initiates new grant proposals concerning the handicapped acts as a liaison between the university and the community, and promotes the overall university expertise.

Most communities have one or more local agencies that are concerned with assisting the handicapped. The directors of these agencies can be useful in outlining some of the more obvious needs of the handicapped. A word of caution is appropriate here. Many times the people who work directly with the handicapped have little or no training in engineering or technology and so are not always aware of the difference between the practical and the "wished-for." Sometimes these people are not fully aware of devices that are commercially available. Often they may be aware of a commercial device but hope that through a student project that the device can be obtained at a lower cost.

This is false economy and the university should not be drawn into areas that are covered by established commercial enterprise.

There are a number of larger universities throughout the country that have sizeable government contracts to conduct rehabilitation engineering. A visit to one or more of these centers would be quite helpful. The author is fortunate in having a research grant from Purdue University to visit a number of such centers and to attend several national conferences on rehabilitation engineering. However, such trips are no substitute for working closely with local people such as the public school teachers who are concerned with the every day problems of working with the handicapped. These teachers are in great need of working closely with technical people who can not only make their jobs a little easier but can advise them on devices that may be commercially available.

At times it may appear that all of the significant research on devices to assist the handicapped is being done at large institutions with large federal grants. Despite the tremendous impact of these programs there is still plenty of room for undergraduate research at the smaller universities relating to the needs of the local community. Operating at this level, only small grants are needed to provide continuity of research. Local foundations and national agencies with local connections can often be a source of the funds necessary to sponsor such research programs.

A REGISTRY OF THE SEVERELY HANDICAPPED IN BRITISH COLUMBIA, CANADA

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ABSTRACT

In order to plan an integrated programme of rehabilitation in British Columbia, Canada an estimate of the disabled population is required. Data to January 1979 from a total of 7140 cases in 39 categories taken from the B.C. Government's Health Surveillance Registry were analysed. The survey is described and a table of the 11 most important categories is included together with three representative examples of the graphical presentation of the results. The reliability of the findings is discussed.

INTRODUCTION

Rehabilitation engineering should be no different from any other engineering discipline in its approach to a problem. First, the problem must be defined and data on the details and extent of the problem must be gathered and analyzed. Secondly, the available expertise must be reviewed and the technological tools required to solve the problem determined. Thirdly, proposed solutions to the problem must be tested and finally, the successful solution implemented.

British Columbia, Canada is a reasonably representative area of North America. It has a population of 2½ million of which approximately 1 million is in the metropolitan Vancouver area, ¼ million in its metropolitan Victoria area and the remainder spread throughout the province in smaller centres and in rural districts. This paper describes the Health Surveillance Registry in British Columbia and how data on the extent of the rehabilitation problem was extracted from this registry. Representative data is presented and its validity discussed.

THE HEALTH SURVEILLANCE REGISTRY

In 1952, the B.C. Government's Ministry of Health established a Crippled Children's Registry. It was later renamed the Health Surveillance Registry. (1), (2), (3). It includes registers of chronic handicapping disabilities, genetic defects, congenital abnormalities, and specific at-risk populations,as well as the provincial cancer registry. The purpose is to provide a reliable basis for estimating the prevalence, incidence, and distribution of specific conditions. It is a cumulative registry from 1952, and is updated every year. Registration is categorized not just by disease, but by handicapped condition, and information is obtained from ten separate sources:

1. Provincial and Metropolitan Health Units.
2. Special Treatment Centres: Registrations, case summaries, and other documents such as monthly separation data sheets are received from the Children's Diagnostic Centre, Woodlands, Glendale, and Tranquille (facilities for the mentally retarded), G.F. Strong Rehabilitation Centre, the Health Centre for Children, Pearson Hospital, genetic counselling clinics, and others.
3. Voluntary Health Agencies: The Canadian National Institute for the Blind and the Canadian Arthritis and Rheumatism Society register cases known to them. Branches of the Cerebral Palsy Association, the Hemophilia Society, and the Cystic Fibrosis Foundation also provide the information about the cases of the disabilities with which they are concerned. Traditionally these associations operate in close cooperation with Public Health Units, and registrations of many of their members have been routed through the Health Units.
4. Federal Indian Health Agencies.
5. Private Physicians.
6. The Physician's Notice of Birth: congenital malformations are noted and described on this form.
7. Death registration (of children under six years) and stillbirth registration: a pregnancy which terminates after 20 weeks' gestation requires a physician to complete a certificate, and in both stillbirth and death certificates there is a question concerning necropsy. The staff of the Division of Vital Statistics screen these certificates and if further information is required, contact the physician concerned. A monthly list of deaths occurring in the province is also scanned for names of registered persons, in order to add the notification of death to the records.
8. Obstetrical Discharge Summary Forms: Three of the major obstetric units in Vancouver; Grace,

St. Paul's and Vancouver General Hospital, whose total births comprise about one quarter of the annual provincial total, completed this special form at the time of the mother's discharge from hospital. It contains information on the infant, including the presence of malformations.

9. Hospital Admission/Separation Forms: for all children under seven years of age, hospitalized in the province with a Congenital anomaly: This source includes all hospitals in British Columbia, as all Admission/Separation forms are submitted to the Hospitals Programs branch of the Provincial Ministry of Health. It is of special value in detecting disabilities with onset after the first 48 hours of life, and in improving the accuracy of registration of previous cases. If ascertainment were to be extended to Admission/Separation forms of all age groups, it could be particularly valuable in providing information on adult disabilities.

10. The Ministry of Human Resources: in 1975 a program was initiated conjointly by the Health Surveillance Registry and the Ministry of Human Resources, with the purpose of improving ascertainment and statistical data concerning disabled adults, so as to obtain a better idea of the location and nature of special rehabilitative facilities which might be required in the province.

The information received from the above agencies is coded according to the International Classification of Diseases. (4), (5). This classification is currently being updated to ICD9 and all the records are in the process of being converted to this new code. Up to four codes may be utilized for each subject enabling multiple disabilities to be registered.

For example a person with multiple sclerosis, who is blind, a quadriplegic and has a speech defect would have the codes 3400, 3790, 3443, and 7815.

DATA ON THE HANDICAPPED IN B.C.

The computer printout of subjects by the ICDA 8 code up to January 1979 was studied in depth. A total of 7354 subjects was reviewed, composed of 4041 males and 3313 females in 41 categories. The search was limited to subjects whose disabilities made them likely candidates for patient operated technical aids and therefore of most direct interest to the rehabilitation engineer. The twelve major categories making up these totals are shown in Table I. In addition to these categories there are 16,773 cases listed under mental disorders and 4738 persons who are blind in both eyes. Some of these also fall into the categories shown in Table I. so that direct addition of the figures is not valid.

In the space available here it is of course not possible to publish the findings in detail. Examples are shown in figures 1-3 and the complete details are published in a report which may be obtained by contacting the Health Surveillance Registry, 828 W 10th Ave, Vancouver, B.C., Canada, V5Z 1L8.

TABLE I

ICDA 8 CODE	DISEASE CATEGORY	M	F	T
2730	CYSTIC FIBROSIS	108	106	214
3303	PROGRESSIVE MUSCULAR DYSTROPHY	130	40	170
3400	MULTIPLE SCLEROSIS	258	513	771
3420	PARALYSIS AGITANS	94	96	190
3430	INFANTILE SPASTIC HEMIPLEGIA	204	152	356
3432	INFANTILE ATHETOID SPASTIC PARALYSIS	102	78	180
3439	OTHER CEREBRAL SPASTIC INFANTILE PARALYSIS	1132	1022	2154
3441	ADULT HEMIPLEGIA	777	596	1373
3442	ADULT PARAPLEGIA	144	88	232
3443	ADULT QUADRIPLEGIA	169	73	242
3449	OTHER ADULT CEREBRAL PARALYSIS	58	45	103
8060	FRACTURE/DISLOCATION WITH SPINAL CORD LESION	476	114	590
		3652	2923	6575
	29 OTHER CATEGORIES	389	390	779
		4041	3313	7354

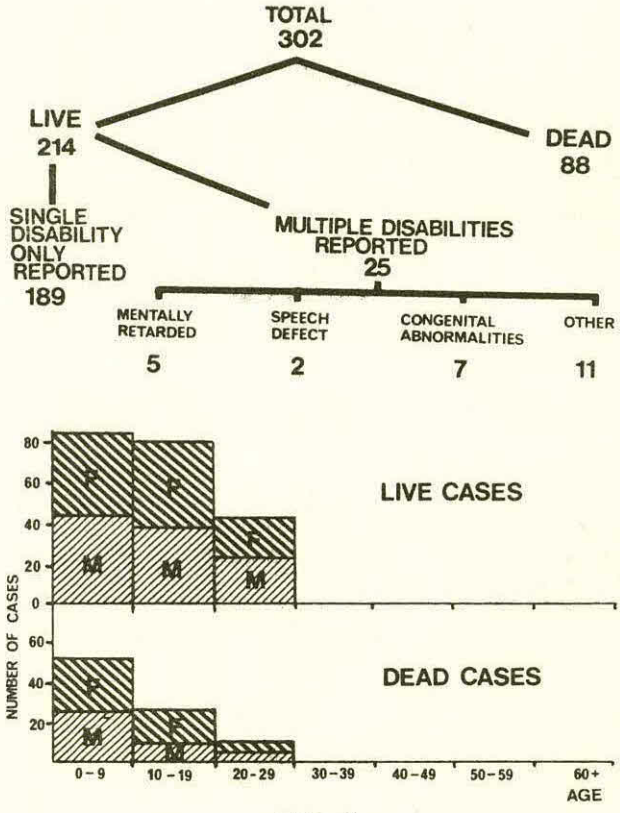

FIG. 1
CYSTIC FIBROSIS (ICDA 8 - 2730)

In figure 1, cystic fibrosis is chosen to illustrate a condition producing disabilities in a juvenile population with a short life span. These persons have a very low reported incidence of other disabilities and represent a low demand for technical aids. They do however require such devices as tents, pumps and other respiratory aids.

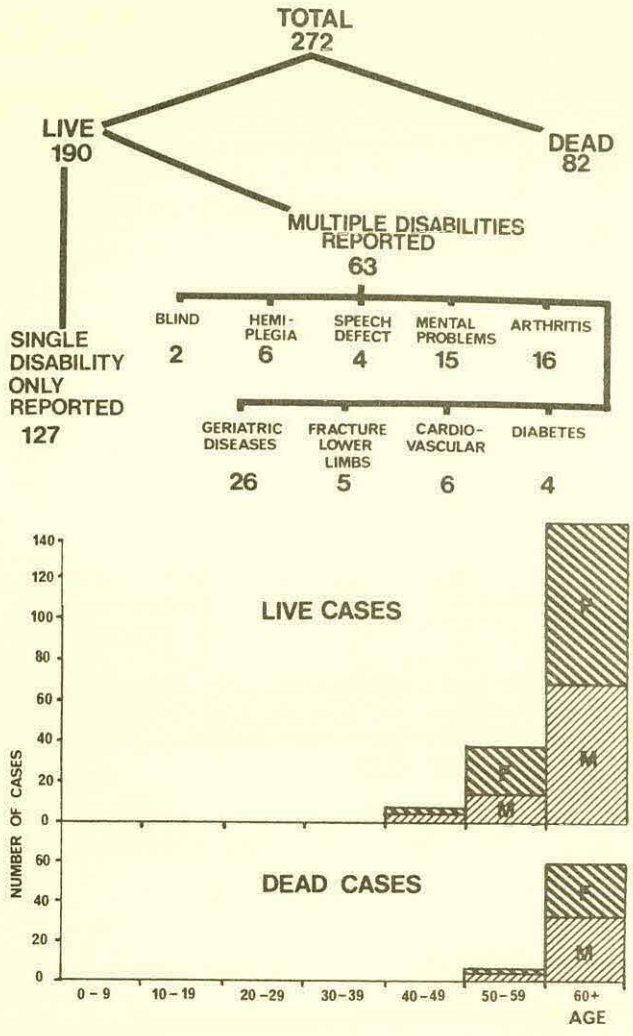

FIG. 2

PARALYSIS AGITANS (ICDA 8-3420)

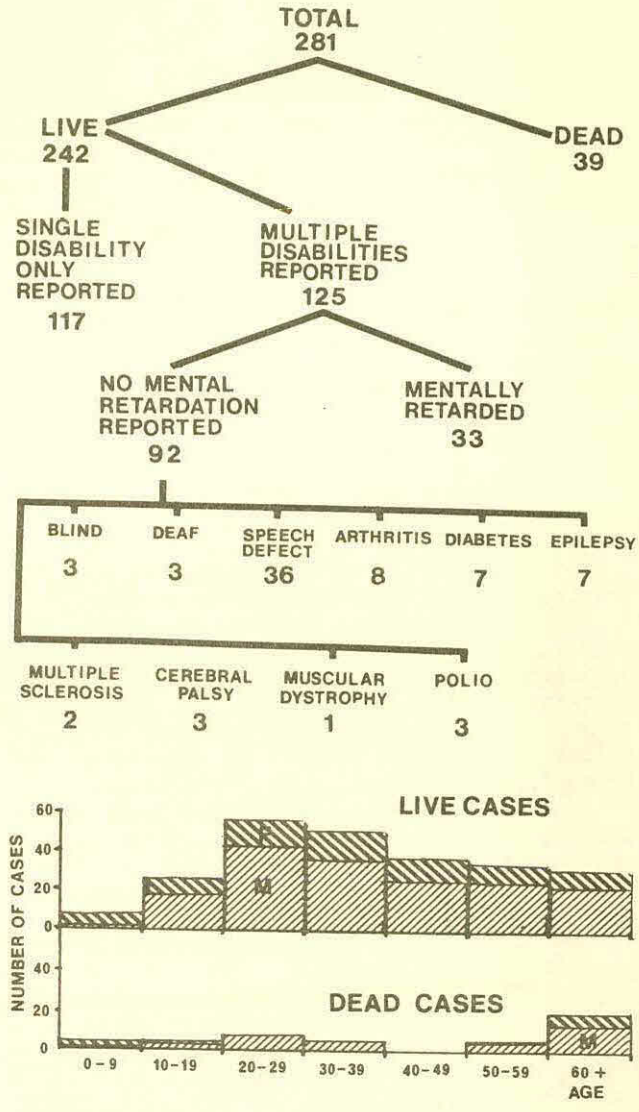

FIG. 3

QUADRI PLEGIA (ICDA 8 - 3443)

In figure 2, paralysis agitans is chosen to illustrate a condition with disabilities in a senior population. These, as might be expected from their age distribution, show an increased incidence of multiple disabilities mainly with diseases of old age. These persons also represent a relatively low demand for technical aids but have normal or slightly higher demand for patient aids such as wheelchairs, than the general geriatric population. They are of course of interest to those rehabilitation engineers working on tremor related problems.

In figure 3, quadriplegia was selected because it contributes the greatest demand for technical aids. It includes the requirement for environmental controls and communication aids and is probably the group whose life style is improved most dramatically by the application of rehabilitation engineering. Because this group includes subjects whose disability was produced by trauma (accident or disease) it has a very high incidence of multiple disabilities. This group shows a more even age distribution with its peak in the 20-29 year group and with a low death rate. Unlike the other two examples where the sex ratio is close to 1 this group has a male/female ratio of over 3/1 in the 20- 29 year group and over 2/1 in the overall group.

DISCUSSION

It must be emphasized that the above figures are based on the reported data. This data is only as good as its source and its reliability is very variable. For instance it is expected that the recording of cases of cerebral spastic infantile paralysis is relatively complete because of the number of reporting agencies involved. At th other end of the scale deafness is not reported reliably and the figures are completely misleading showing a peak in the 20-29 year age group four times as large as in the over 60 age group. In general the registry's coverage of children is very good but the adult figures are expected to be lower than the true incidence. This situation is being corrected and a large number of adult registrations have been added by the Ministry of Human Resources in the last few years.

At present the cases from the B.C. Workers' Compensation Board (WCB) (industrial accident) and the Insurance Corporation of B.C. (ICBC) (traffic accident) are not made available to the Registry. It is hoped this can be changed in the future. This data is not lost to the Registry however because most of these persons pass through the G.F. Strong Rehabilitation Centre in Vancouver and this organization does report to the Registry. The WCB has approximately 200 severely disabled persons on its files. A sample comparison with the Registry files indicated an 80% coverage by the Registry. A similar comparison of the ICBC figures is scheduled to be performed and a check will be made by the Kinsmen Foundation of B.C. who have 156 severely disabled persons in their technical aids programme. These comparisons are difficult to arange because of the confidentiality of data.

One problem with the data is the ICDA code which is a classification by disease not by disablility. ICDA 8 had no classification for quadriplegia for infantile diseases. 3443 being reserved for adults. This has changed in ICD 9 and it will now be possible to determine the number of quadriplegics under the classification for cerebral spastic infantile paralysis. ICD 9 does not solve many problems however: there is still no separate category for complete loss of speech function. Fortunately with the cooperation of the Registry staff, during the conversion a special code will be created for this category. The category for deaf in both ears which was available under ICDA 8 will no longer be available under ICD 9 and it will no longer be possible to code this as a disability.

In spite of the above problems the results obtained from this survey provide a good basis for planning of rehabilitation services and in particular the need for rehabilitation engineering.

B.C.'s populations of 2.5 million is approximately 1/10 that of Canada and 1/100 that of the USA so that approximate extrapolation of the expected frequency in Canada and the USA can be made. With the ongoing improvement to the Registry and improved reporting of multiple disabilities and the inclusion of data from WCB and ICBC, the figures will give a reliable indication of the handicapped population of British Columbia and provide to rehabilitation workers and industry an estimate of the requirements for their services and products.

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ACKNOWLEDGEMENTS

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THE DEVELOPMENT OF A REHABILITATION ENGINEERING SERVICE

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ABSTRACT

The development of a program to provide the unique services of an engineer to rehabilitation problems is described. Both inpatient and external referral systems are outlined. The role of the engineer as a team member is emphasized. Data relating to the diagnoses, problems, and solutions are given.

BACKGROUND

With the advent of support for engineering as it relates to the field of rehabilitation, the possibility of improving the quality of life for the moderately and severely physically disabled patient has certainly enlarged. The Rehabilitation Engineering Center at Moss Rehabilitation Hospital embarked upon an engineering program in July, 1977 with the designation of a rehabilitation engineer who would provide direct patient services. Our initial approach was to provide such services to inpatients who were admitted for a special inpatient unit known as the Research Utilization Service (RUS) under the supervision of the physiatrist who was also the medical director of the center. We found that the rehabilitation engineer was capable of providing a wide variety of services to these inpatients in conjunction with therapists in areas such as activities of daily living, communication, mobility, and home evaluation.

In November, 1977, a program emphasizing vocational rehabilitation engineering applications was developed along with the Pennsylvania Bureau of Vocational Rehabilitation. In order to accommodate this type of external referral of clients predominately post treatment, a team format was utilized. At this time the basic unit consists of the rehabilitation engineer and a therapist. The engineer devotes 100% of his time to engineering service application. His background is in mechanical engineering. The therapy position was defined as that of an occupational therapist. However, at the time no occupational therapist was on staff at the center and the position was temporarily filled by a physical therapist with extensive clinical background and some combined PT/OT education. Initially the therapist spent approximately 20% time on this project. As the service expanded, more and more therapy consultation time was required, and, finally, in March, 1980 a full time occupational therapist was hired for the project with the assistance of Moss Rehabilitation Hospital and a grant from the Pennsylvania Bureau of Vocational Rehabilitation. Other team members include the physiatrist, a rehabilitation nursing specialist, and other engineering support as required.

ACTIVITIES

The rehabilitation engineer participates in clinical activities and attends the twice weekly clinical inpatient rounds on the research utilization service with the physiatrist and other hospital personnel. He is also available by referral to all other medical staff and therapists for consultation, the installation of environmental control units, and as a controller of the evaluation equipment supplied by the Veterans Administration as part of their Rehabilitation Engineering Demonstration Unit (REDU) program. With the addition of a full time occupational therapist, it is anticipated that the inpatient service will be expanded and representatives from the team will attend all other hospital service rounds to insure early referral of problems to the team. As of this time, the engineer is the director of the Rehabilitation Engineering Service.

Outpatient referrals come from the Bureau of Vocational Rehabilitation or from any other external sources. For example, a physician with no relationship to the hospital may refer a patient for Engineering Clinic consultation. Private companies have requested consultations relating to work site evaluations for employees and equipment estimates have been made for third party payers. Such external referrals have involved the Defense Supply Depot, Blue Cross, Firestone Tire Company, International Rehabilitation Associates, Elwyn Sheltered Workshop, Conrail, and Sheltered Employment Service. In addition, a contract has been negotiated for a monthly consulting visit by both the engineer and therapist, with the Kent County School for the Orthopedically Handicapped in Delaware.

The following description summarizes the outpatient or BVR clinic format. The clinic is held one afternoon every second week and two clients are scheduled. To date this clinic has accommodated BVR clients, although the area has recently expanded to other referrals. BVR administration has appointed a liaison counsellor as a permanent representative. When an individual counsellor has a case requiring engineering consultation, he reviews the case with the liaison officer who

determines the applicability for the clinic and then prepares a case summary and lists the particular reasons for referral to the clinic, for example, job site evaluation, or home accessibility. The engineer and the liaison officer determine what action is necessary prior to a clinic visit. Such decisions may include a driver evaluation and a home visit. Home visits are done by the BVR liaison officer, the engineer, and the therapist. A clinic appointment is then made and the full team meets with the liaison officer and client to discuss the problems and present the results of prior evaluations. In all cases, whether BVR related or private referral, the final report of recommendations is sent to the referring agency and not directly to the client. We are now in the process of instituting a follow-up program to determine the action actually taken by the referring agencies or individuals based on the clinic recommendations.

ANALYSIS OF CLIENT'S PROBLEMS

As the service has developed over a period of two years, records have been kept on 77 clients (reporting date November, 1979). Table 1 outlines the sources of clients. Of these referral sources, 26% came from the inpatient RUS, and 27% came through the outpatient engineering clinic. These clients had a total of 113 problems (Table 2) for which solutions were recommended.

Table 1. Client Sources

Client Sources	# of Clients	%
In-Patient RUS	20	26
Clinic BVR	21	27
Consultation		
Temple Hospital	6	8
Moss Hospital	4	5
Einstein Hospital	2	3
Kent County School	12	15.5
Other	12	15.5
TOTAL	77	

Table 2. Types of Problems and Number of Clients with Each Type of Problem

Problems	# of Clients/ Problem	%
Activities of Daily Living	39	35
Mobility	18	16
Home	27	24
Communication	16	14
Transportation	6	5
Work Site	7	6
TOTAL	113	

Table 3 indicates the types of solutions that were most frequently required for each category of problems. For example, under activities of daily living (ADL), adaptive devices were required much more frequently than environmental control units or orthotic solutions. Problems of access were quite frequent and this included access within the home as well as getting into and out of the home.

Table 3. Types and Numbers of Solutions for Problem Categories

Solutions to Problems	# of Problems	%
ADL		
Adaptive Devices	31	
Orthotics	5	31
Environmental Control	5	
Mobility		
Locomotion Aids	6	
Wheelchair	10	15
Seating	3	
Home Evaluation		
Accessibility	27	23
Equipment Modification	3	
Communication		
Non-vocal Devices	8	
Vocal Devices	1	
Call Systems	4	12
Electronic Comm.	3	
Transportation		
Personal Licensed Vehicles		
Operation	4	
Access	4	9
Modification	4	
Work Site Evaluation		
Access	2	7
Equipment Modification	7	
No Solution	4	3

The vast majority of cases referred for engineering evaluation had disease of the central nervous system. The four most common disease categories were craniocerebral trauma, cerebral palsy, spinal cord injury and multiple sclerosis. Categories of problems as related to diagnoses are given in Table 4.

An area of major interest in this project is to establish the domain of rehabilitation engineering problems with particular regard to the type of

Table 4. Type of Problems Related to Diagnosis

Diagnosis	#	Problem					
		ADL	Mob	Home	Comm	Trans	Work Site
Spinal Cord Injury	12	6	3	6	3	1	1
Stroke	3	1	3	0	1	0	0
Peripheral Neuropathy	9	6	2	3	2	0	0
Cranial Trauma	12	4	6	4	3	0	0
Cerebral Palsy	17	8	2	2	6	1	3
Non-traumatic Brain Injury	4	3	1	1	1	0	0
Arthritis/ Musc. skel.	8	4	0	1	0	2	3
Multiple Sclerosis	10	6	1	8	0	1	0
Amputation	2	1	0	2	0	1	0
TOTAL	77	39	18	27	16	6	7

personnel necessary for providing such services. We have taken a team approach towards problem solving because it was anticipated that the engineering service delivery model for handicapped patients would be similar to a medical rehabilitation model which also revolves around a team approach. The primary engineering team consists of a rehabilitation engineer, a therapist, and, when concerned with inpatient and ongoing treatment, a physician. Additional resources available to the engineering team are a transportation specialist (a rehabilitation nurse), a prosthetist/orthotist, and a number of engineering center personnel with skills in mechanical and electrical engineering. In reviewing the case histories, problems and solutions for the past two years, it was apparent that the recommended solutions for patients referred to the Rehabilitation Engineering Service required team problem solving in the vast majority of cases. Contributions from health professionals as well as engineering staff were required to generate an appropriate solution. For the solutions generated, 67% required team problem solving, and 33% of the solutions required specific engineering problem solving over and above the contribution of the engineer to the team problem solving. That is, after the rehabilitation engineering team had defined a solution for a particular client, the engineer still had to develop a specific plan and/or design in order to implement the solution.

SUMMARY

In summary, the development and operation of a Rehabilitation Engineering Service has been described. Our model utilizes a team concept with the involvement of the client's existing treatment personnel wherever possible. Experience over a period of two years demonstrates that the model can be well adapted to accommodate external referral agencies as well as inpatient referrals.

Selected case studies will be presented if time permits.

CENTER FOR ADAPTIVE ENGINEERING: A PROFILE OF COMMUNICATION SERVICES

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UNITED CEREBRAL PALSY AND HANDICAPPED CHILDREN'S
ASSOCIATION OF SYRACUSE, INC.

The U.C.P. & Handicapped Children's Association offers a variety of therapeutic, educational, psychological, social, community and medical services to physically handicapped individuals, ranging in age from infancy through adulthood. In the fall of 1979, C.A.R.E. (Center for Adaptive Rehabilitation Engineering) was established at this facility to provide adaptive equipment services including assessment, design, construction and review. This federally funded program includes a rehabilitation engineer, a technician and associated personnel. The grant also provided for the purchase of materials and tools. Speech, occupational and physical therapists from the Center staff, team up with the rehabilitation engineer and his associates to improve the overall quality and effectiveness of existing services by dealing with individual mobility, seating, and communication challenges. Communication skills analysis and procedural strategies are presented in the following discussion.

INTRODUCTION

The primary objective of the Center for Adaptive Rehabilitation Engineering (C.A.R.E.) is to provide a comprehensive adaptive equipment program for those in-house individuals requiring such services and referred to the C.A.R.E. team or to one of its members. It is expected that these modified aids, devices and systems will directly and positively affect the well-being and life-style of the client. Education and activities of daily living should be significantly facilitated.

A secondary objective of C.A.R.E. is to become a resource center and referral agency for other facilities in Central New York treating individuals requiring adaptive equipment services. C.A.R.E. will also train personnel from outside organizations in the art of adaptive equipment development.

In addition, C.A.R.E. utilizing its unique attributes is to establish a data retrievable service and information clearinghouse for individuals, organizations, schools, and agencies seeking rehabilitation related literature and product technicalities.

There are other programs such as IMPART[1] which were created to perform similar functions, that is, to "make the prescriber-segment and the user-segment of the rehabilitation community aware of available equipment and devices and technical information related to rehabilitation and everyday living."[2] IMPART is, as CARE is, a public service; however, CARE is innovative in that its focus is on physically disabled children aged 0 - 7, not vocationally-oriented adults. CARE deals with the special design needs of young people and therefore must develop strategies and approaches which allow for size and weight gain, changing medical needs, variations in daily environments, and educational placements and milieus, in addition to more typical project concerns. The CARE engineer and staff must create plans to accomodate a dynamic, growing, human being, ever-changing and challenging. Habituated patterns and postures have not been formed and perhaps with versatile design may be precluded.

Adaptive equipment which is flexible for the child must also be flexible for the parent or caretaker, for this individual represents the "secondary user." The engineering effort should readily adapt to change and facilitate "primary user's" normal activities, thereby simplifying "secondary user's" daily responsibilities. This is the total concept of adaptive equipment design.

In the case of youngsters aged 3 - 7, it is expected that significant gains in the areas of postural tone, socialization, independence, self-concept, feeding, education, and communication

should be made.

ACTIVITIES OF CARE

An enumeration of general procedures employed by CARE program is as follows:

1. Each client referred to the program will have received a complete medical, psychological, therapeutic, and educational evaluation at the Cerebral Palsy Center.

2. An individualized program will be developed for the client by the rehabilitation team consisting of the client, engineer, physical therapist, occupational therapist, communication specialist, educator (when appropriate), and parent.

3. The engineer will meet with client and appropriate team members to examine physical status, determine specific equipment needs, and to formulate design conceptualizations.

4. The technician will construct the engineer's design.

5. Initial fitting will occur in the presence of involved team participants. Several fittings are anticipated.

6 Client and caretakers (secondary users) will be instructed by engineer and most appropriate team member as to the specifics re application of aid.

7. The equipment will be adjusted as needed.

8. Secondary users will be requested to report in writing at regular intervals on effectiveness of equipment.

9. As youngsters change educational environments, the team will discuss rationale and inception of adaptive equipment so as to establish a strong foundation of support.

10. The CARE will develop costs for each piece of equipment which will be available for distribution.

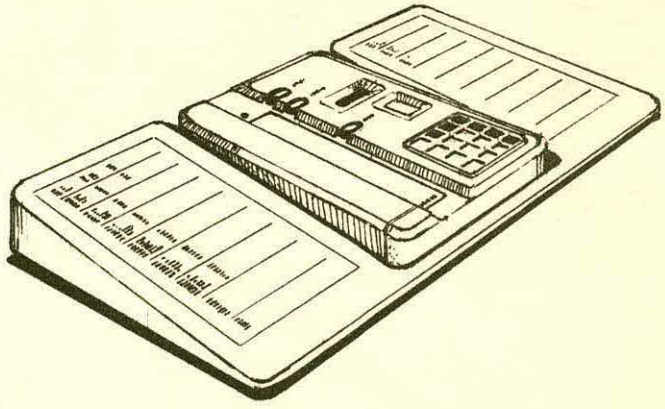
COMMUNICATION EVALUATION

The application of rehabilitation engineering technology to the design and development of communication systems and devices for non-speaking youngsters is a challenging and pioneering endeavor. CARE combines the non-speech expertise of a speech pathologist and the creative talents of the rehabilitation engineer to formulate an innovative program.

Prior to the client's referral to CARE for a speech prosthesis, a comprehensive "Election" and "System Selection" has been completed. Election[3] may be defined as the decision making process for determining an individual's candidacy for a nonspeech communication system. This decision results from a review of ten areas of clinical concern. One of three decisions may be generated by this procedure, "elect" (adopt a nonspeech system), "reject" the introduction of an augmentative aid, or "delay" the implementation of a speech supplement. Assuming the speech pathologist determines appropriate candidacy, that is, an "elect" decision is reached, a system "selection" process follows. Enlisting the aid of certain criteria, a "first-level" decision is made for the client; whether to adopt a manual "sign" mode, or a communication prosthesis (aid, device, language board, etc.). The"second-level" decision involves the identification of specific hardware (augmentative display) and software (augmentative content or symbol system) should a prosthesis be selected. The speech pathologist may examine the client's "demonstrated potential" on commercially available systems, samples of which have been purchased by the United Cerebral Palsy Center for assessment and training purposes. A device recommendation is made if appropriate and feasible for the client.

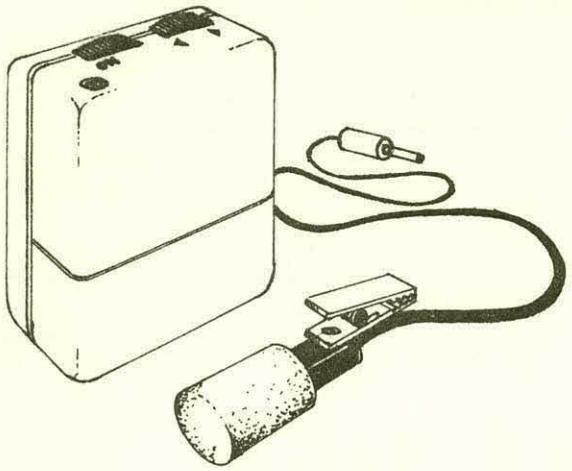
Available Options Provided By CARE

The majority of non-speaking individuals are candidates for augmentative aids. Very often though, the systems and devices accessible to the consumer require personalized adaptations to enable the user to derive the most benefit. CARE's services include modifications of existing instrumentation as is described below.



VOCABULARY TRAY DESIGN FOR PHONIC MIRROR HANDIVOICE HC120.

Some severely physically limited individuals require switch adaptations to enable them to activate a commercially available product.

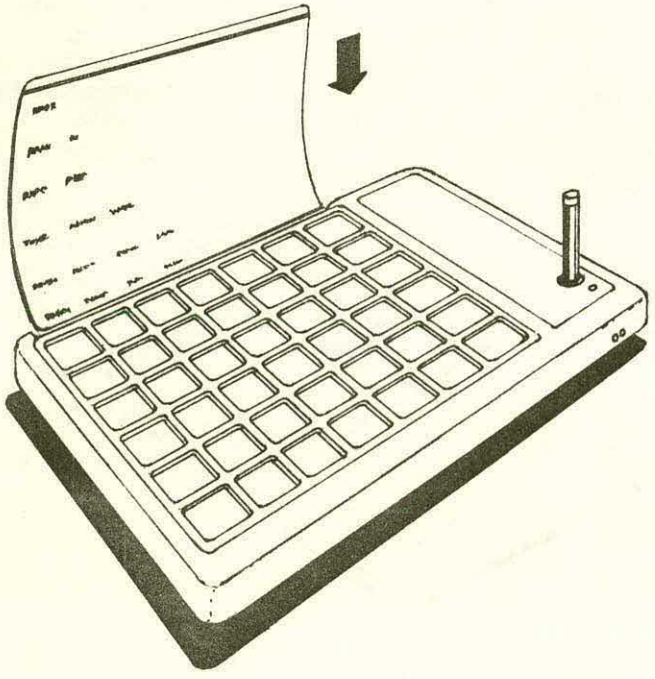


INTERFACE SWITCH : VOICE ACTIVATED PERMITS SEVERELY INVOLED INDIVIDUAL TO ACCESS DEVICES.

The design and construction of original adaptive equipment is one of CARE's primary interests. The matching of consumer needs with inovative design and current technology to create a customized communication system is depicted below.

COMMUNICATION AID:

UTILIZING PRESSURE SENSITIVE PRINCIPLE./ HAND OR FINGER/ POINTER ETC. RESPONSE IS BOTH VISUAL AND AUDITORY



CONCLUSION

The preceeding discussion and case studies demonstrates the efficacy and validity of housing a service such as CARE within an existing rehabilitation agency such as the Syracuse Cerebral Palsy Center. Disciplines such as the communication disorders field can form partnerships with rehabilitation engineering and create opportunities for the engineer and speech pathologist to work together and mold the images of the future.

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EMPLOYMENT APPLICATIONS OF COMPUTER RELATED SENSORY AIDS FOR HANDICAPPED PERSONS

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ABSTRACT

This report describes the sensory aids and adaptive devices utilized by handicapped people in computer related jobs. Sensory Aids Foundation is operating a program to Expand Employment Opportunities for Physically Disabled Individuals in Computer Related Jobs throughout the State of California. This program is funded by a grant from the Department of Health, Education and Welfare, Rehabilitation Services Administration, Projects With Industry.

At the present time there are many computer related jobs which occur in business, industry and government. These jobs will become readily available through the continuous development of microprocessor-based and other electronic devices.

INTRODUCTION

The data processing industry is a rapidly expanding one. The Department of Labor makes the following projections for the outlook of computer related positions:

Job Title	% Growth for 1985 Requirements
Programmers	49.5%
Systems Analysts	65.2
Engineers	32.8
Typists (includes Word Processing)	34.9
Airlines Reservationists (includes ticket and passenger agents)	35.8
Computer Operating Personnel	10.1
Engineering and Science Technicians	41.4

Sensory Aids Foundation (SAF) has been interested in these positions for the last 4½ years and has been involved in several different types of placements of handicapped people in employment situations.

BACKGROUND

A brief description of Sensory Aids Foundation would be helpful to provide background information. Sensory Aids Foundation has been developing new areas of employment for blind and visually impaired persons throughout the State of California since September 1975.

The Projects With Industry program expands this concept to include deaf and orthopedically disabled persons. Many of these innovative jobs require the use of advanced technology to allow access by a handicapped person. Because the data processing industry is rapidly expanding, attention will be focused on that area at this time.

Since we're all familiar with electronics and computers, we know that more and more jobs in business and industry are required to interface in some way with a CRT terminal. In the rapidly expanding job market this is also the case. Many pieces of equipment used by handicapped people in computer related jobs have cathode ray tube (CRT) screens.

INNOVATIVE EMPLOYMENT POSITIONS

Presented below is an update on some of the employment positions which SAF has been involved in using sensory aids to open computer related jobs to handicapped people.

A totally blind legal secretary and word processing operator works for the law firm of Glasby and Rice. An Optacon with a CRT lense and tracking guide are utilized with the Wang Word Processing System to provide the necessary information to the blind person. The Optacon and CRT lense provide the blind person with the exact information on the CRT screen. The Optacon allows a totally blind person to read anything in a printed format.

A partially sighted engineer is working at Ford Aerospace Corporation utilizing a closed circuit television system in conjunction with the Hewlett Packard 9825A. Since the employee could not see the LED display on the HP9825A a special configuration of a closed circuit television system was developed. The engineer can now view the computer printout as well as enter new calculations since the CCTV has a split screen capacity.

An interesting example of equipment currently in the marketplace can be utilized by totally blind people without any additional sensory aid is the Rolm CBX computerized business telephone system built by Rolm Corporation, Santa Clara, California. The system has audible tones which indicate the status of calls. For example, a double interrupted ring signals an incoming call, a single interrupted ring denotes internal calls, and a long uninterruped ring is used for call backs. After answering the call, a simple digital pad keyboard, similar to that on a conventional pushbutton telephone, gives the operator access to all extensions by pressing just three buttons.

It is interesting to note that this equipment was not developed with a blind person in mind.

There are now over 1000 of these Rolm CBX systems in service today. Because of the wide availability of this system hundreds of jobs can be opened to blind people throughout the United States. At the current time Sensory Aids Foundation has been instrumental in the placement of two totally blind people working with this equipment. One person is working at Four Phase Systems, Cupertino, California and the second at Rolm Corporation, Santa Clara, California.

Other companies are reviewing applicants at the present time. We anticipate additional placements utilizing this equipment.

SUMMARY

Placements discussed in this article represent many varied employment situations in the computer field. Computers will increasingly become more evident in the job market and now handicapped persons have the ability to perform these jobs competitively through the use of sensory aids such as the Optacon, closed circuit television and microprocessor-based systems.

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SUCCESS AND FAILURE IN VOCATIONAL REHABILITATION ENGINEERING

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ABSTRACT

Application of engineering to the vocational rehabilitation process is yet a youthful practice. Problems of timing, perception, and communication occur frequently; yet published experiences are usually positive. Believing there is much to be learned from unsuccessful applications efforts, this paper discusses several case histories and draws conclusions which are generally applicable to vocational rehabilitation engineering.

INTRODUCTION

That vocational rehabilitation engineering is effective has been demonstrated by several programs. We have the technology to adapt very severely handicapped individuals to demanding, competitive jobs. But we have not yet identified the best way for states to integrate this capability with their ongoing vocational rehabilitation services. Rehabilitation engineers willingly share information on devices and techniques that have led to successful placements. Something we have not done is discuss and share information on the heartaches and failures that we have all experienced. The purpose of this paper is to reveal what may be learned from vocational rehabilitation engineering failures. To avoid being dismal, however, we chose to alternate between a successful experience and a dismal, or failure, type of experience. It shall be shown that our failure is often related to timing or improper interaction with vocational rehabilitation service delivery people; that is, counselors. We intend to examine several case histories.

CASE HISTORIES

Dictation Assist for Quadriplegic Executive

This was a very successful vocational rehabilitation engineering venture which involved operational procedures at the worksite. Secretaries were unable to understand the client's recorded dictation. His speech was affected by a neuromuscular disease. The counselor thought that he needed a speech clarifying mechanism. Before investigating the speech clarifier or some other electronic device, we borrowed a report he had dictated when his voice was tired, and supposedly at its worst. We brought the tape to our secretaries to transcribe it. Our secretaries reported that they could understand the tape, with some difficulty, but it was certainly possible for them to type from it. This simple test result was returned to the counselor who confronted the people at the worksite and the problem disappeared. From a practical standpoint, having a professional rehabilitation engineer visit the worksite gave the counselor clout. This was a very practical solution to a co-worker acceptance problem which undoubtedly arises fairly often.

Deaf Cerebral Palsied Money Sorter

A deaf cerebral palsied client was operating a change sorter in a bank. Because the man was deaf, he couldn't detect when the machine jammed, which was a common occurrence. Naturally, we proposed to install an indicator which would light up to alert the deaf individual that the change sorter was jammed. Further observance noted that the reason for inadequate surveillance was that he was too busy handling the money that had already been sorted, bagging it, and sealing the bags. If the man had only the duty of watching the sorting machines, he could do the job well. We suggested he be provided a vise, which in effect gave him an extra hand. This would have enabled him to handle the money quicker and freed him for the surveillance operation. However, the client already felt tremendous pressures from his peers and his supervisor. He lost the job before we could intervene because of poor behavior; he reported to work under the influence of intoxicating liquor. This was a failure for vocational rehabilitation engineering. A failure because we and the counselor did not act quickly enough to solve the problem.

Low-Vision Telephone Operator

When the counselor contacted us, the client's job was in jeopardy. Because of her low-vision, she could operate a switchboard, but couldn't do filing or paper shuffling. Because the client worked slow, the employer was frustrated. Rather than fire the individual, the job was eliminated. In all likelihood, the job could have been saved had the rehabilitation engineer been able to make a site visit and talk with the supervisor. Earlier contact by the counselor might have saved the job.

Quadriplegic Police Dispatcher

The client was a quadriplegic. He was about to lose his job because he couldn't do it rapidly enough. We discovered a number of very simple things that helped him almost triple his speed. We raised the desk so he could get his legs under the worktable and closer to his work. Since the man had limited hand function, he couldn't put records in the time clock. They hung up in the machine, slowing down his productivity. To solve the problem, we acquired cards of precise width so that they did not hang up in the machine, making the job much easier. There were other simple adaptations that the counselor had already identified such as a holder for the telephone receiver. The problems were very simple and had straightforward solutions. Nevertheless, this was a successful vocational rehabilitation engineering venture and saved the man's job. Simplicity is essential.

Deaf Laboratory Technician

A laboratory technician had a severe hearing loss. She was concerned that some of the specimens would be spoiled because she would not hear timer and temperature alarms. Examination by a rehabilitation engineer determined that almost all of the alarms could be provided with visual indicators. However, the rehabilitation engineering consultation revealed very clearly to her co-workers that the client had a hearing problem; a fact she had been able to disguise. The client was understandable offended. Rehabilitation engineering was ineffective because we didn't consider the client's need. We must regard it as a failure in vocational rehabilitation engineering.

SUMMARY

Many rehabilitation professionals regard vocational rehabilitation engineering as an art, or developing science. While developing our technical expertise, we must pay attention also to the human aspects; the personal relationships between the counselor, the client, and the rehabilitation engineer. These are vital considerations that demand serious attention if we are to provide cost-effective vocational rehabilitation engineering.

AN APPLICATION OF COMPOSITE TECHNOLOGY TO ACHIEVE A LIGHTER WEIGHT WALKER

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ABSTRACT

This paper identifies two patient catagories that could benefit from a reduction in weight of current walkers in use. A material of construction and fabrication techniques which will yield significant weight reduction are discussed. A design for a walker using these techniques was developed. A prototype walker was fabricated. Functional operation is discussed and the weight savings attained are noted.

1. INTRODUCTION

There are a variety of ambulation aids available for the handicapped. These aids vary from simple canes to Quad canes, crutches, and various types of walkers. Most of these assistive devices for ambulation are fabricated from aluminum and steel. Various components of these aids are also made of rubber for specific cushioning effects. The steel used in such devices is used sparingly in those places where maximum strength is required.

As a general rule, patients who only require canes as ambulation aids have reasonably good control and strength in their upper extremities. Canes require only a small amount of material and consequently, most are sufficiently light as to not require any additional weight reduction. In addition, because of their relatively small initial weight, significant gross weight reduction is difficult to achieve. Crutches are several times heavier than canes and are required by persons with less functional ambulatory capability. Again however, the overall weight of a pair of crutches, particularly the newer lightweight aluminum crutches, is not very much. For most persons requiring crutches, the aluminum crutches are sufficiently light in weight for the needs of the patients. The maximum weight reduction that could be achieved merely by the substitution of lightweight materials is somewhere on the order of one pound. For most patients this is a marginal decrease in weight and would require a significant increase in cost because of the expense of the fabrication and raw materials involved in lightweight processes. Crutches, therefore, are not particularly appropriate candidates for lightweight, high-strength materials. It could be argued that for needs of certain particular patients who do a significant amount of ambulation and a significant number of transfers from automobiles, a small weight reduction could be significant enough to warrant fabrication of lightweight crutches. Overall, however, it is clearly a marginal situation since the cost of the crutches themselves would increase.

For those more severely disabled in their ambulatory capability, a family of aids called walkers have been developed. These generally consist of a four-legged, three-dimensional space frame within which the patient stands. This frame is used both for support and steadiness and also to permit ambulation using arm strength much as in the manner of arm crutches. Most of these walkers are fabricated from aluminum with steel tubing employed at points of stress. Most are designed to permit folding so that the patient can store the walker in the back seat or in the trunk during transportation by automobile. Nominal weights of such walkers are in the range of 2.7 to 3.6 kilograms (6 to 8 pounds). These basic walkers have been modified in numerous ways to provide additional structural support for specific patients having more severe disabilities. For example, forearm rests have been added and, in some cases, even the upper portion of crutches have been added to permit weight bearing at the shoulder.

Again, for a large segment of the population of patients who employ walkers for ambulation, the aluminum and steel walkers presently on the market are entirely adequate. The occasional user of a walker who has reasonable upper arm strength and upper body control experiences little difficulty with the weight of currently available walkers. There are two classes of patients, however, in which the weight of the walker itself poses somewhat of a problem.

First, the very active patient with good upper body strength but with significant functional loss in the legs tends many times to have a life style in which he seeks to live a relatively normal life going from place to place, holding down a job,and seeking not to compromise

his mobility to any significant degree. Such patients do a lot of ambulation and may be in and out of an automobile a large number of times each day. For such patients, a significant weight reduction in the walker would be a positive benefit and would reduce their daily energy expenditure. A weight reduction would also produce greater convenience in handling, particularly in the process of transfering from the walker to a vehicle after which the walker must be folded up and then stored by the patient while he is seated in the driver's seat of the vehicle. This group of highly active patients who require a walker for functional ambulation could benefit from a reduction in weight of the walker itself.

The second category of patients who would benefit from a reduction in weight of the walker is the patient suffering severe disability; so severe that ambulation is impossible or extremely difficult with walkers of conventional weight. Most of these patients are, of course, not frequent ambulators but would ambulate more frequently if use of the walker itself were not such a chore. In addition there are patients whose disability is sufficiently severe as to preclude the use of walkers at the present time because of the weight of present day walkers. Most of these patients fall into a particular pattern. They are usually arthiritic patients having multiple joint involvment. Frequently, additional structures are required on the walker to permit ambulation, and usually they have little strength, or little ability to exercise strength without severe pain. These are the classic "little, old Ladies" with multiple joint involvment arthritis. These patients with severe pain and motion impairment in multiple joints including arms and shoulders would benefit from a significant reduction in the weight of the walkers.

2. MATERIAL AND FABRICATION CHOICES

It is with these two classes of patients in mind that we have attempted to develop a lighter walker than is presently available on the commercial market. We have chosen to opt for the strongest, lightest-weight materials available in order to achieve maximum weight reduction. This involves the use of composite structures. We have selected a graphite fiber-epoxy resin material as an appropriate choice for this application. Composite materials generally suffer from two distinct disadvantages in the marketplace. First, the fabrication of composite materials is frequently expensive. Second, the materials used in fabrication are often quite expensive. The use of graphite-epoxy in applications such as the one which we have addressed has been greatly hindered by the cost added by the technology when compared to that of units made from conventional materials. Both factors, the cost of fabrication and the cost of materials have contributed to this problem. I might hasten to point out that this has not been a significant deterrent to the use of graphite-epoxy and other composite materials in recreational devices. For example, it is not at all uncommon to find the fisherman using a graphite-epoxy fishing rod. In addition, graphite-epoxy and other composite materials are being used in the fabrication of skis, tennis rackets, golf club shafts; others to numerous to mention are continually being developed employing these expensive technologies. This says something about the value structure of our society; however, such is not the subject of this paper. The point is that while not everyone who fishes with a fishing rod buys a graphite-epoxy rod, there are those who feel that the marginal benefits to be gained are worth the additional cost. In considering the patient population using walkers we feel that the two categories of patients discussed will realize sufficient marginal benefits to warrant the additional cost of these walkers.

In addition, there have been several encouraging developments which promise to reduce the cost of items fabricated from graphite-epoxy and other composite materials. These factors are: (1) As graphite-epoxy and other composite materials are used in more and more applications and as they find their way into the mass market, the relative cost per pound of such materials is decreasing (2) The recent development of a particular fabrication process offers the promise of significant reduction in fabrication cost. This process is called pultrusion. The pultrusion process permits the continuous forming of specific shapes from composite fiber-resin systems. For example, using the pultrusion process, it is possible to make continuous rods of almost any diameter, as well as continuous bar stock, right-angle, "U" and other shapes including tubing. The advent of the pultrusion technique promises to reduce the cost of graphite tubing.

3. DESIGN PHILOSOPHY

With the promise that these costs will be reduced, we have felt encouraged to embark on a design fabrication study using these materials in the fabrication of a walker. We have chosen to develop a design based upon the use of straight tubes of graphite-epoxy which can be pultruded. In order to construct a three-dimensional space frame from straight tubes, it is obvious that methods of joining these tubes in a reliable fashion must be found. In this program we have concentrated on methods of joining straight graphite-epoxy tubes so as to achieve the three-dimensional space frames required. Anyone who has played with an erector set or tinker toys knows that it is necessary to have connectors which will permit, at the very minimum, "T"-joints and right angle joints if rectangular frames are to be constructed and assembled to form three-dimensional space frames. The examples that I shall give are prototypes designed to illustrate the

principles and should not necessarily be construed as the final configuration of such connectors.

In most respects, the weight of the graphite walker, as designed, is larger than necessary in that most of the structural components are overdesigned. This was done for two reasons. First, the tubes employed have a larger diameter and greater wall thickness than required for the application based upon structural calculations. We felt that patient acceptance is such a significant factor that the use of smaller tubes might lead to mistrust on the part of patients, so we chose to make a very strong walker whose tubes are essentially not much smaller than that found in commercial walkers today. Second, the use of stronger components than necessary contributes to overall rigidity of the walker itself. Both of these factors: construction in a familiar size and high rigidity will contribute, we feel, to ready patient acceptance of the units.

4. DESCRIPTION OF GRAPHITE-EPOXY WALKER

Figure 1 shows an oblique view of the graphite-epoxy walker. Note that all components of the walker are of graphite-epoxy material with the exception of the rubber foot caps on the bottom of each leg. One item not visible is the bearing material used in the joints involving the front horizontal connector bar. The two upright side panels rotate in each end connector of the front horizontal bar. This connector is essentially of a clam-shell design, as can be seen from the photograph, with two pieces glued together with epoxy adhesive to form a "T"-joint. The horizontal bar is securely attached with adhesive. The front vertical member of the side frame has a piece of heat-shrink Teflon tubing over it in the area of the joint. The front vertical member is not glued to the connector so that the side frame is free to rotate in the tube formed by the two halves of the "T"-joint. The Teflon covering of the vertical tube in this area permits easy rotation and eliminates wear on the graphite-epoxy material itself. Two graphite rings are epoxied in place above and below the joint to hold it at its proper height. These are essentially stop rings.

In this prototype, two measures are employed to ensure rigidity of the walker when it is in its operational configuration. First, the top horizontal bar as seen in the photograph is attached with a slip joint to the horizontal member of the right side panel. The other end of this horizontal tube has a modified "T"-joint which is free to slip over the forward end of the upper horizontal member of the left side panel. When in place as shown in Figure 1 this bar provides a locking mechanism. It holds the two side panels perpendicular to the front spacer bar providing a rigid, U-shaped walker.

The second means of providing additional

Figure 1 - GRAPHITE-EPOXY WALKER, DEPLOYED

stiffness to the walker involves the use of a U-shaped structure consisting of three straight pieces of graphite-epoxy tubing and two right-angle joints. This U-shaped member remains suspended in front of the walker when not in use. When in use, it is rotated up and over the walker and locked into the two clips, one on each side on the rear vertical member of each side panel. Coupling to the front horizontal connector bar is achieved by means of two components formed from two sections of a right-angle joint. These are employed in such a fashion as to provide a clamp around the horizontal connector bar with a stand-off and sleeve which fits over the horizontal section of graphite tubing at the base of the "U". This permits swiveling of the "U" so that the walker can be folded. When the "U" is rotated up and over into its operational position and locked into the two clips at the lower portion of the rear vertical member on each side panel, the "U" bracket provides additional resistance to spreading of the rear portion of the walker. Tests show the walker to be strong enough without the U-bracket. The net result is that approximately 325 grams (0.75 pound) of additional material can be eliminated resulting in both a weight saving and a cost saving.

Figure 2 shows the walker in a folded configuration, and Figure 3 shows it as it would normally be carried. Most conventional walkers on the market today have a telescoping bar which holds the two side panels at a specified point. In order to fold the walker, a catch is released and both of the side panels are folded in toward the middle, one lapping over the other to achieve a folded structure. With the graphite walker, a thinner package can be achieved by folding one leg of the walker inward and by rotating the other completely around to the front. This results in a thickness reduction of the folded

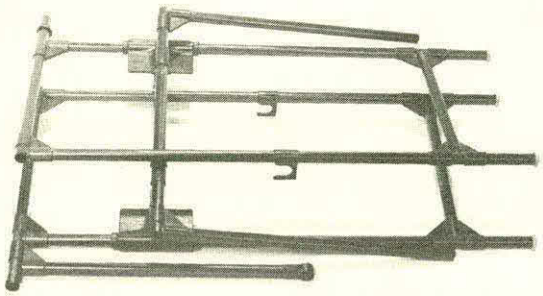

Figure 2 - WALKER, FOLDED

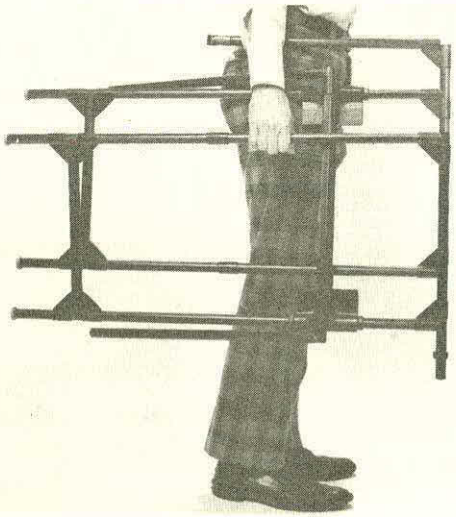

Figure 3 - WALKER, CARRY POSITION

walker of several inches. The present walker is unfolded in the following sequence. The side panel (folded to the front) is rotated around to the 90-degree position with respect to the front horizontal connector bar. The other side panel (folded to the rear) is opened up again to the 90-degree position. The U-bracket is rotated up and over the walker and lowered until the rear ends are engaged in the two stops, one on either side of the rear vertical member of the side frame. Next, the top spreader bar is rotated up into position and slid back toward the rear, engaging the sleeve on the left end of the spreader bar with the horizontal tube of the left side panel. It is pulled to the stop, thus completing deployment. Figure 4 shows the assembled walker in the normal position for use. It will be noted that no hand grips are apparent on the top horizontal bars of the side panels.

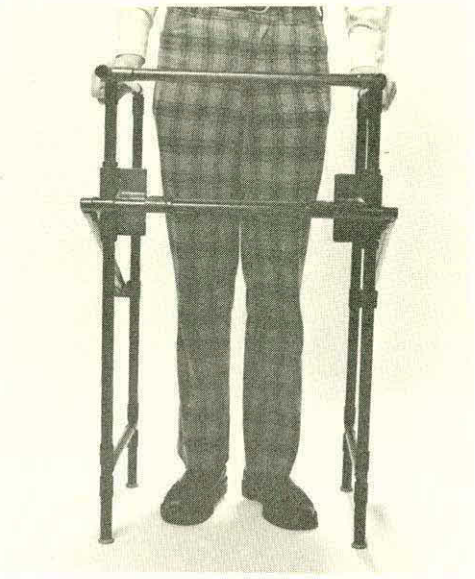

Figure 4 - WALKER, OPERATIONAL POSITION

These were not shown so as to reveal the graphite structure more clearly. In actual use, rubber hand grips will be cemented in place on the walker.

In the next prototype, the U-shaped bracket will be eliminated, and assembly of the walker will be simplified because only the top spreader bar will be involved. In the new design, it is anticipated that the top spreader bar will be broken at a point approximately two-thirds from one end. Both ends of the spreader bar will be captivated on the horizontal bar of each side panel. They will be allowed to rotate however. A graphite-epoxy sleeve will be fabricated to fit snugly over the horizontal spreader bar. To assemble this unit, the two portions of the spreader bar will be rotated upward to the horizontal position and then the sleeve on the longer piece of spreader bar can be slid over the junction of the two to a stop point, thus locking the spreader bar in the horizontal position and providing both the lock and the stiffness required.

The walker as shown in the photographs has a weight of approximately 2.0 kilograms (4.5 pounds). This represents a 1.1-1.4 kilogram (2.5-3.0 pound) reduction over that found in conventional walkers fabricated out of aluminum and steel. Elimination of the U-shaped bracket will reduce the weight by an additional 0.3 kilogram (0.75 pound). The net result will be an extremely lightweight, portable walker.

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REDESIGNED PARAPODIUM

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The hip and knee locking configuration of the parapodium designed at the Ontario Crippled Children's Center is simple in function but can be difficult for young children to operate. Particularly, the simultaneous release of both locks using both arms is sufficiently frightening to prevent the disabled child from learning unaided sitting maneuvers. Without an ability to sit and stand readily, the child is inhibited from participation in other classroom experiences. The observed benefits of a parapodium in the development of the young paraplegic are sufficient to warrant an effort to extend the usefulness of this orthosis. An alternate design has been completed and an initial evaluation confirms the design objectives.

Figure 1

The redesigned parapodium has independent knee and hip locks, each with single lever release action. Locking occurs automatically on extension. This frees one arm to aid in balance control. During the process of sitting or standing, the child can flex his hips while the lower portion of the orthosis remains rigid and supportive. The design uses commercial hinges and components that can be fabricated and assembled in an equipped orthopaedic shop. Two views of the new parapodium are seen in Figure 1. Figure 2 shows the child extending his knees while the hips are flexed. The hip release bar is visible in the unlocked position under the right forearm. The knee release is a fiber strap shown encircling the knee extension assist bar.

This parapodium design has been fitted successfully on four young patients, ages 3 to 10 years, with meningomyelocele ranging from L5 to T12. Each child developed the facility for unaided sitting and standing. Increased confidence from this new freedom to move within a classroom apparently allowed the development of other skills and participation in new activities.

Partial support for this work has been provided by the J.M. McDonald Foundation, Cortland, NY.

Figure 2

THE HEYER-ABADIE "TALKING" MOUTHSTICK

ARTHUR HEYER

EXTENSIONS FOR INDEPENDENCE

To me, as a quad with no use of hands, the mouthstick I use means life. Thanks to this mouthstick, plus some peripheral equipment I use, I am functional for work and study as I always dreamed of being.

In this paper I am describing this mouthstick, now commercially available, after having spent a full year making prototypes to experiment with new and better ideas. In our new design, the following factors have been combined:

1. A mouthstick to enable the user to talk and swallow while in use.
2. A modular design where the mouthpiece is a separate unit and different mouthstick "bodies" are available for the type of functions each individual wants to perform.
3. A mouthstick "body" available with interchangeable implements for quad independent operation.
4. A simple and sturdy light-weight design.
5. Custom fit possible by mail.

For the sake of clarity, this paper has been composed in the form of questions and answers.

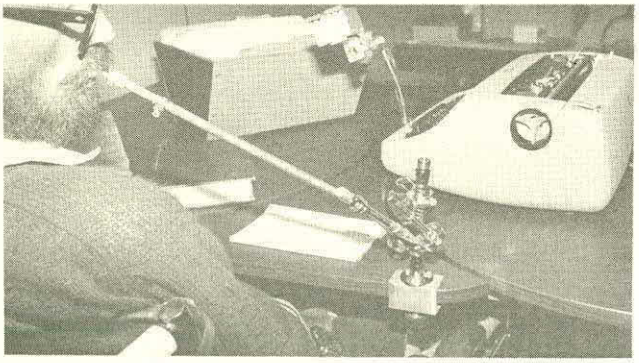

Fig. 1. The Heyer-Abadie "Talking" Mouthstick

Your mouthstick looks different from others I have seen. How can you talk with it in your mouth?

That's why we named it "Talking" mouthstick. My father designed it. He is a professor of phonetics and he was very concerned about this feature when he first made me one. Observe that the mouthpiece is flat and "V" shaped (see Fig. 2). In my mouth, I hold it with my back teeth and my tongue is free to move. My front teeth remain slightly open, giving a free path for air to flow out of my mouth. These are the features that allow me to talk and swallow while I use it (Fig. 3).

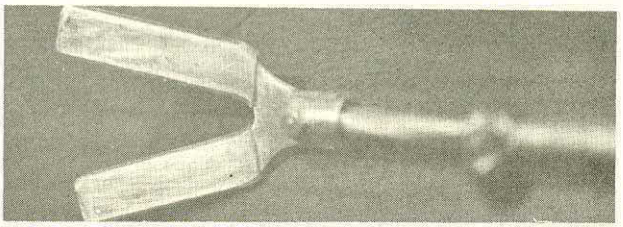

Fig. 2. Flat, "V" shaped mouthpiece.

I see only metal on this mouthstick.

....And plastic sleeves on the mouthpiece for a comfortable grip (Fig. 2). The mouthpiece is most important. We custom make it to conform to the size of the customer's bite and mouth. The prongs are tapered to the end following the natural angle between the maxilla and the mandible that is formed when biting the mouthpiece. This tapering of the prongs eliminates a concentration of pressure on the back teeth which otherwise would occur. The plastic sleeves make a good cushion for the teeth, and only in cases of severe bite malformation would it be necessary for a dentist to do additional work on the mouthpiece for a custom fit.

Fig. 3. When holding the mouthstick in the mouth, the front teeth remain separated, allowing air to flow out of the mouth for talking.

How often does a dentist have to custom fit a "Talking" mouthpiece?

We made a study by having 50 persons of all ages, from 5 to 72 years old, bite on wax sheets, and we found out that 82% of the 50 subjects had normal or acceptably normal bites suitable for using our mouthsticks as we make them; 14% were doubtful, with probability of needing some dentist work, either on their teeth, or on the mouthpiece. These were cases of subjects with molars missing on one side, or with irregular teeth. The two remaining subjects, a man 32 years old, and a woman 62 years of age, would definitely necessitate the dentist's intervention in order to operate our mouthsticks comfortably. Most of the subjects studied, 45 of the 50, ranged from 5 to 40 years old with a uniform distribution of ages. A concentration of doubtful or bad bites was found among the people over 40 years of age.

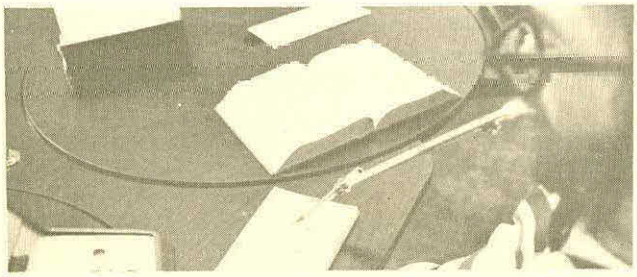

Fig. 4. Pen is angled for optimum use.

I notice a very peculiar position of the pen on your mouthstick.

It points down and towards my left side. I found this position best. Notice that this is the same position with respect to the paper that a pencil or a pen is normally held with the right hand. In this manner the mouthstick user can see the tip of the instrument, what he is writing, and what he has already written.

Do you make all the mouthsticks the same?

No. The shape and size of the bites and mouths of different people vary. And as I said before, we have to customize the mouthpiece. We do this by having our customers bite on a piece of wax sheet that we mail them and they return together with information related to length of mouthstick, amount of bending at the mouthpiece, and type of mouthstick as related to its function.

For example, if all a person wishes to do is to turn pages of a book and nothing else, we provide him with an extremely simple and light-weight mouthstick which is actually a very light aluminum tube that attaches to the custom mouthpiece, and has a rubber tip. On the other hand, for the more sophisticated user we have a telescoping mouthstick which can be combined with a simple rubber tip for turning pages only, or with our basic tool holder with Quad Independent Release Mechanism (Fig. 6) for independent interchange of basic tools and attachments.

What are these basic tools and attachments?

We call BASIC TOOLS the pen, the mechanical pencil and a hook designed to pull objects, like books. We call ATTACHMENTS the implements that attach on the basic tools to change their function. For example; we have an eraser attachment that fits on the pen or the pencil and that the mouthstick user can pick up or leave independently. In a work situation, this operation may sometimes be done dozens of times in one hour. For example, a writer who uses the pencil to edit his typed work and the eraser to erase or to handle the papers.

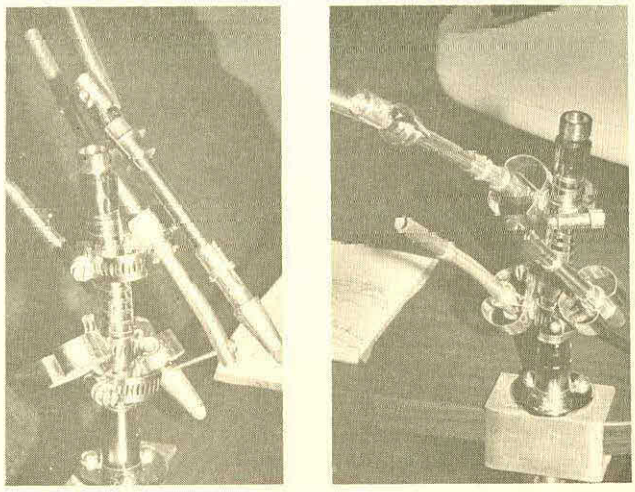

Fig. 5. (Left) Moving the sleeve to raise the clip. (Right) Inserting an eraser attachment into a gripping clip.

How do you change attachments?

The attachments are thin-wall aluminum cylinders that fit snugly on the pen or mechanical pencil. Into the other end of these attachments, different things, like an eraser or a brush, can be inserted. The attachments can be pulled off or pushed onto the pen or mechanical pencil by means of a two-step docking procedure: If, for example, the attachment is initially on the pen, the mouthstick user simply inserts the attachment into a gripping clip of the docking system...and pulls. (Fig. 5)

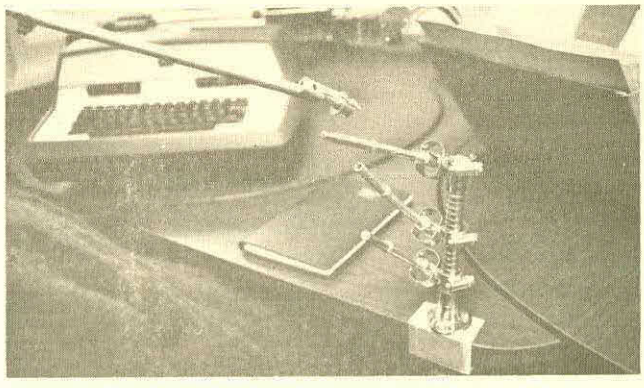

Fig. 6. Ready to pick up a Basic Tool.

The attachment stays behind in the clip because of the double-ring design of the attachment's body which keeps it from sliding out of the clip.(Fig.7). To pick it up, one inserts the pen or mechanical pencil in the attachment...and makes a lateral motion out of the gripping clip.

How do you interchange basic tools?

The BASIC TOOLS fit smoothly into the BASIC-TOOL HOLDER, which is a short aluminum cylinder attached to the end of the body of the mouthstick.

When a Basic Tool is inserted into the holder, a clip attached to the holder automatically hooks it.

A short stainless steel sleeve with a wide flange is so designed as to allow the mouthstick user to slide it by pushing the flange to raise the clip and release the Basic Tool. We call this the QUAD INDEPENDENT RELEASE MECHANISM. The mouthstick user uses the gooseneck tool docking holder (Fig. 5), a chin control arm, or any fixed large object as a fixed point of support for pushing the flange of the sleeve.

If the user wants to interchange the pen for the mechanical pencil, he first picks up an attachment, then he operates the Quad Independent Release Mechanism, and...once the Basic Tool (the pen in this instance) is released or unhooked, the mouthstick user inserts the attachment into the gripping clip (Fig. 5) and pulls gently. The pen will be pulled out of the Basic Tool Holder because the friction between the attachment and the pen is greater than the friction between the pen and the Basic Tool Holder (Fig. 6).

To pick up a Basic Tool, the mouthstick user gets the tool inserted into the holder and pulls. As was described previously, the clip of the Basic Tool Holder automatically hooks the Basic Tool when this is inserted into the holder. The Basic Tool is picked up, and the attachment remains inserted in the gripping clip. Any number of Basic Tools, and of attachments, can be made easily accessible to the mouthstick user with this system.

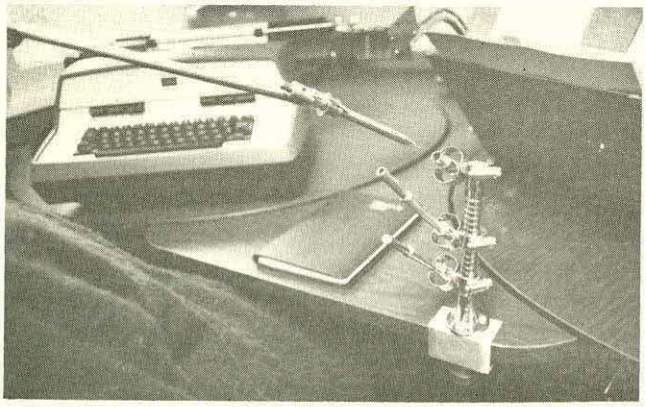

Fig. 7. The attachment stays behind in the gripping clip.

How long can you work with your mouthstick without feeling soreness of the muscles of your neck?

Most likely soreness is a symptom of muscle building up. After 15 years of mouthstick use an average of 8 hours a day, my neck muscles are very strong and I never get tired at my neck before I get tired of sitting.

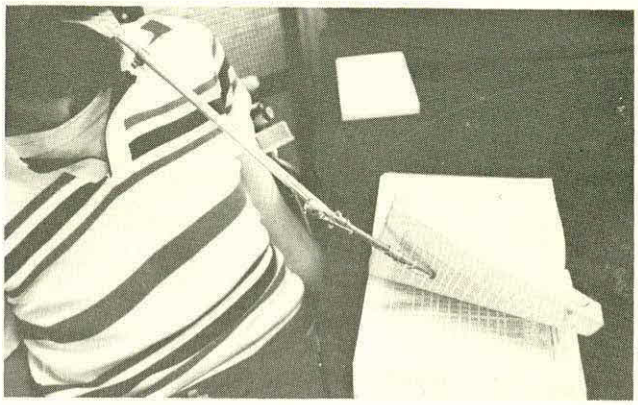

Fig. 8. Turning pages of a book.

Do you think that any quad can learn to use a mouthstick well?

I cannot tell. There are so many factors that may contribute. In general, the successful use of a mouthstick may depend on the following:

1) Motivation.
2) A well designed mouthstick, with emphasis on the design of the mouthpiece.
3) Amount and quality of neck and shoulders muscle strength and control.
4) Sitting stability, position, and tolerance.
5) Accessability of work area and adaptations or peripheral equipment.

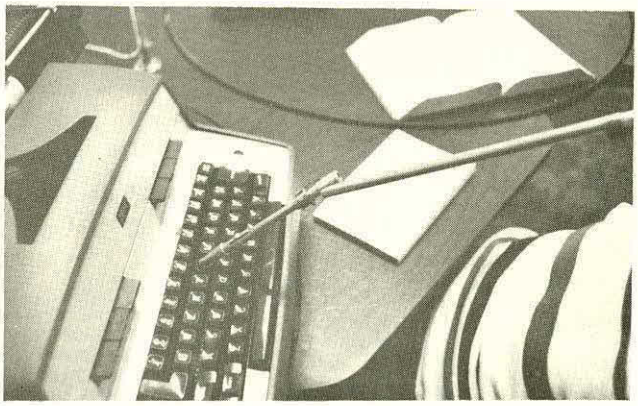

Fig. 9. An eraser attachment is used for typing.

How are the muscles of the shoulders related to mouthstick use?

It was not until recently that I realized that while working with my mouthstick, I was using not only the muscles of my neck, but also those of my shoulders and my biseps of which I have some control. The role these latter muscles play in the use of a mouthstick seems to be that of maintaining equilibrium.

PERIPHERAL EQUIPMENT*

A mouthstick alone has limited use if the work area is inadequate. Four pieces of equipment supplement my work area for me to be able to work 100% independently in an office situation:

Fig. 10. Self-contained workstation.

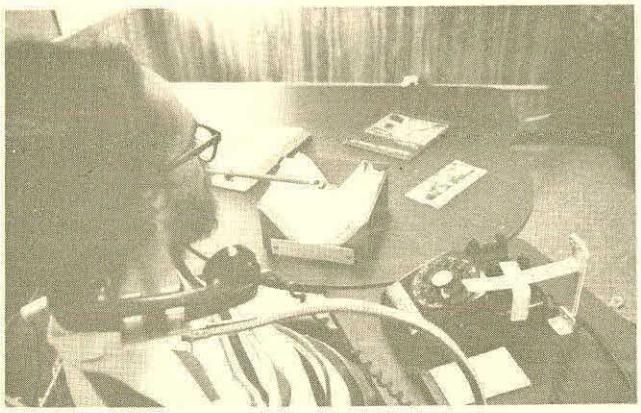

Fig. 11. Using the File Tray while talking on the telephone.

Fig. 12. Portable Desk under Twin-Turntable Desk.

*A description of this equipment is found in Ref.#6

1. A Twin-Turntable Desk (Fig. 10) with which I can bring a large work area within my reach by rotating the turntables with my mouthstick.
2. A Portable Desk (Fig. 12) that attaches onto my lap-board.
3. A Telephone Adapter (Fig. 11) that enables me to use a regular telephone independently.
4. A Quad-Operation File Tray with slanted sides to allow the folders to maintain themselves open when so needed, and with an open front to facilitate the filing of papers with a mouthstick.

CONCLUDING REMARKS

The strong needs that I had for functional equipment motivated me to spend thousands of hours analysing the needs and looking for solutions. I hope that the time and effort spent in coming up with solutions to solve my own physical limitations serve to decrease the effort of others in their endeavor of making the handicapped more functional.

ACKNOWLEDGEMENTS

I owe much to my father who invented the basic tools for me to start my way back to a productive life, to other members of my family and friends who have helped in one way or the other, to Dan Richardson for his invaluable assistance in the production of evaluation samples,to Dr. Bill Utley, Director of the Dental Department of Rancho Los Amigos Hospital, for his assistance in obtaining a statistical analysis of bite sizes and shapes, and Iespecially thank Henry Abadie, who since retirement, has been most valuable in the development of this mouthstick and other very meaningful devices for the disabled.

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AIDS FOR THE HANDICAPPED AND THE GENERAL PUBLIC
or
"YOU DON'T HAVE TO BE DISABLED TO USE THESE AIDS"

Daniel Barak

Weizmann Institute of Science, Rehovot, Israel

Generally, aids for the handicapped are either specially designed or are made by suitable modification of standard existing equipment.

Here we describe the opposite situation.

Technical aids, which were originally designed for specific problems were discovered to be "convenience products" for the general public. In fact, we only realized how attractive these could be when colleagues who saw them said, "Hey, that's a good idea. Would you make one for me too, even though I'm not disabled?

Among the aids which were originally designed for the disabled and which were found to be suitable for general use are: toothpaste squeezer, sit-up bed, slide-out refrigerator shelf, soaping device, automobile distress signal, finger exerciser, book holder and remote electric switch. Some of these are described in detail.

BACKGROUND

The Scientific Services Department at the Weizmann Institute of Science in Israel, is a service department made up of approximately 50 engineers and technicians whose primary function is to design and build scientific equipment according to the needs and requests of the Institute's scientists.

This background was found to be most suitable for the development and production of technical aids for the disabled. In the last few years a variety of such aids have been made and given free of charge to a number of disabled people, who have found them to be a help in their daily lives.

This activity is carried out on a voluntary basis, and costs are covered, in part, by government funds, and to a lesser extent by private contributions.

MODUS OPERANDI

Behind each and every aid we have made, is the story of one disabled individual who came to us with his personal problem.

Our approach is as follows:
1) Meet with the disabled person to define the problem and decide on a technical solution.
2) Check to see if the needed aid can be purchased. If not,
3) Seek simplest solution, utilizing existing equipment, wherever possible.

Much of our activity is influence by the fact that Israel is far from the main sources of supply of daily-aids for the handicapped.

AIDS FOR THE "ABLED"

As mentioned above, we always design aids on a one-to-one basis. However, in a number of cases we later found that more than one disabled person can be helped by the same aid. We have even found that these aids can be used by a larger population such as the aged, hospital patients and even the general public. A number of such aids are described herewith.

Sit-up Bed

Problem: With the exception of the partial use of one hand, Y.Z. is nearly totally paralyzed. He paints by mouth and is a member of the international group of foot and mouth painters. For health reasons he must have his bed raised and lowered periodically, and of course, he was unable to do so independently. Because he shares a double bed with his wife, and also for esthetic reasons, the idea of a hospital bed was rejected.

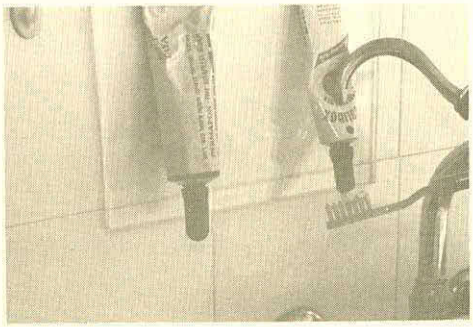
Solution: An electric-pneumatic system which meets the above-mentioned requirements was installed in the bed.

General Use: Because of its small dimensions, ease of installation and operation, low cost and particularly the ease with which it can be adapted to any bed, this aid is attractive not only to paraplegics, but also to other bedridden persons, the elderly and even healthy people who would enjoy the convenience it affords. With suitable adaptation, the foot of the bed can also be raised and lowered.

Toothpaste Tube Squeezer

Problem: Upper extremity amputees and those who suffer from partial paralysis of the hands cannot squeeze toothpaste (or any other) tubes, nor can they operate the screw-top cap.

Solution: The tube is mounted upside-down on a perspex or wooden board and hung on the wall. The screw cap is replaced with an eye-dropper-type rubber bulb which is slit across the top. Leaning or pressing on the tube causes the contents to be squeezed out of the slit in the rubber bulb. It seals itself when the pressure is released.

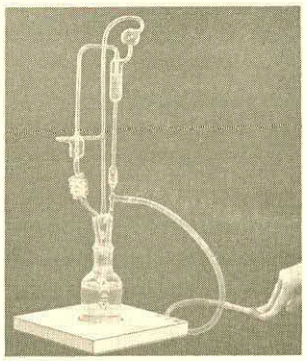


General Use: This device is enjoyed by the children of the disabled, who no longer have to be told by their mothers, "Don't forget to put the cap back on the toothpaste tube". It also serves as an attractive, neat, clean and permanent toothpaste holder.

Finger Exerciser

Problem: To provide an interesting and effective exercising device for a person with weak finger muscles.

Solution: An attractive "game" was devised from a complex system of glass tubing through which colored water is made to flow, by using finger pressure on rubber bulbs.

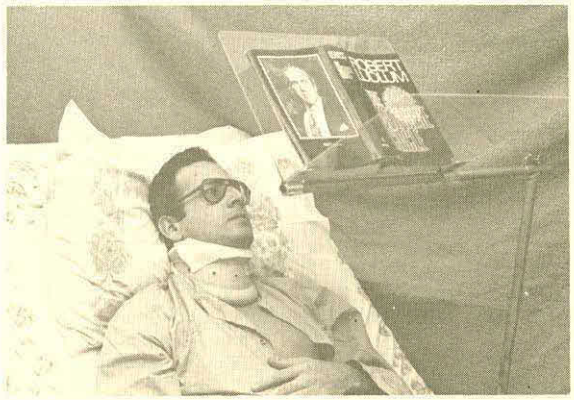


General Use: Quite unexpectedly, we discovered that people were attracted to this device and that it is a success as a "desk game".

Book Holder

Problem: A bedridden patient who had to lay in a prone position was unable to read, due to the difficulty in holding a book in a reading position.

Solution: A book holder was made of aluminum tubing and a sheet of transparent perspex. The device was fitted to the bed. Its positions and angles are adjustable. The pages are turned by the user himself.



General Use: Any person who needs (or just wants) to lay in bed and read can be assisted by this convenient, simple, inexpensive device.

Distress Signal for Disabled Drivers

Problem: Disabled drivers needed a means of summoning assistance in the event of an automobile break-down.

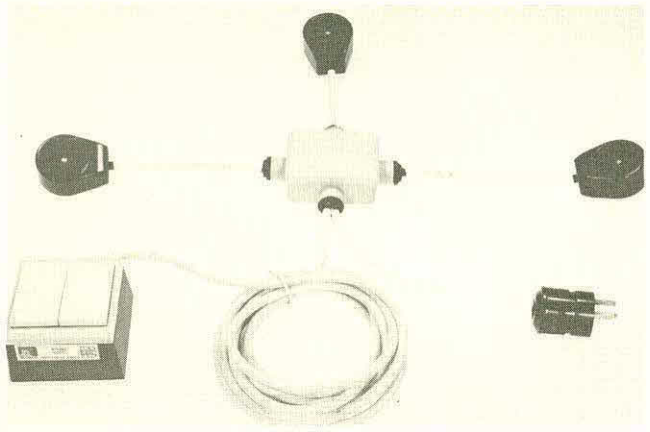
Solution: Two lamps, one with the symbol of the handicapped, and the other with the printed message, "DISABLED DRIVER NEEDS HELP" were installed in the rear window of the car. They blink alternately, thereby attracting attention, when activated by a switch on the dashboard.

General Use: The same system, with different symbols and/or messages, would be useful to a wider public.

"Octopus" Switch

Problem: A disabled bedridden patient was unable to operate electrical appliances which were not within his reach.

Solution: A long electric cord was fitted with a double switch, which the patient kept next to him in bed. He could then activate two electrical devices whenever he desired, at the touch of a switch.



General Use: Naturally, this aid was found to be useful to other bedridden persons. I only became aware of its universality when my mother-in-law suggested that I make one for her. She now enjoys the convenience of being able to switch off her light and television set, without having to get out of bed. (In Israel, remote-controlled television is not yet generally used.)

S. L. Garber, T. A. Krouskop, E. Hemmen

The Institute for Rehabilitation and Research

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The design, development, fabrication and evaluation of highly specialized adapted equipment for use by persons with physical disabilities is a major objective for the rehabilitation engineering team. These devices are used in all aspects of an individual's life and assist him in all personal, vocational, educational and social pursuits. A number of such devices have been developed at the Texas Rehabilitation Engineering Center and three will be discussed in detail. They include a helmet device for the operation of a camera by a quadriplegic, a guitar modified for use by a bilateral upper extremity amputee, and an ADL device to enable a bilateral upper extremity amputee to put on his own prostheses.

INTRODUCTION

The major objective of the Rehabilitation Engineering Centers is the "improvement of the quality of life of the handicapped through broad interdisciplinary collaboration among medical, engineering and allied health professions". (1) The development of equipment that enables a person with a physical disability to maximize his potential has been the focal point of the efforts of many rehabilitation experts. Hundreds of devices are used daily by the physically disabled and provide them with increased function in all personal, recreational, social, vocational and educational activities. Many of these devices are commercially available. However, there are those situations that necessitate highly specialized design and fabrication of equipment. The collaborative efforts of therapists, engineers, physicians and the patient himself have resulted in unique devices that enable the person with a physical disability to achieve his highest level of physical independence. More than twenty such devices have been developed and are in use by the persons for whom they were designed; the three considered in this communications are the latest contributions.

1. Helmet Device for a Camera. Painting is an activity that can be useful and rewarding hobby or vocation for persons with high spinal cord injuries. However, these artists are often limited in what they can paint by how long it takes them to work and how willing an attendant or friend is to carry the needed equipment to the appropriate scenic spot.

Several enterprising artists have attempted to overcome these shortcomings by using a camera to capture the scene they wish to paint. But this approach also has a drawback; the artists' attendent is required to take the picture and the artist does not then have control over the photograph's composition. This requires that the artist rely on memory to compensate for deficits in the photograph.

In order to overcome these problems a system has been designed and developed that permits high SCI persons to photograph scenes, controlling the picture's composition, and work independently. This system, Figure 1, consists of a modified motorcycle helmet, mounting bracket and a Polaroid SX 70 Sonar camera. Using the snaps on the helmet that are normally used to hold a visor in place, it is possible to attach a jointed camera mount that permits the user to view the scene being photographed. By moving his head and looking at the scene, the artist can adjust the photo composition until it suits his neck. Problems of focusing are overcome by using the Polaroid Sonar model camera. Thus the photographer has freedom from having someone else focus the camera and set exposures.

Shutter control is provided by modifying the commercial remote controller so that it is activated by a lever rather than push button as shown in Figure 2. Figure 3, shows a client with the system in position for use.

This system is successfully being used by a former patient from TIRR. He is now able to capture the scenes he wishes to paint as shown in Figure 4. The system is relatively inexpensive and once put in place by an attendant, it does not require adjustment while the artist is in field. By using the Polaroid camera, it is possible for the artist to evaluate each picture's composition, make adjustments while at the

scene, and reshoot the picture if necessary.

Thus far, experience with the system is quite reliable and should not require maintenance other than routine camera maintenance and occasional washing of the helmet. It thus appears to be a viable solution to a problem that has limited the creative efforts of SCI artists and could be used by a larger number of SCI persons who wish to photographically record their experience while on trips.

2. Device for a Bilateral Upper Extremity Amputee to Put on His Own Prostheses.

For the bilateral upper extremity amputee, functional independence begins with his being able to put on and take off his prostheses. This is frequently impossible if the individual lives alone. For the individual living with other people, this task represents an accomplishment that enables him to do all other desired activities at his highest level of functional independence. A device has been designed and developed to enable the bilateral upper extremity amputee to put on his prostheses. It consists of a wooden box with a stiff, but flexible foam that grips below the elbow portion of the prosthesis. The box may be clamped or permanently attached to a dressing table, desk or other conventional flat top surface. To take off the prosthesis the arms are put in a flexed position and pushed into the molded foam cutout, the client then ducks his head under the harness and gently pulls his residual limbs out of the prosthesis. To put on the artificial limbs the process is reversed with the individual pressing his residual limbs into the prosthesis and ducking his head under the harness system. This device is simple, inexpensive and requires only several hours of training to be used effectively.

3. Guitar Adaptation for Upper Extremity Amputee.

A device has been developed that can be used by an upper extremity amputee to chord on a guitar. This device is particularly important to a young man recently referred to TIRR since before his accident he was an accomplished guitarist. The devices has provided him with the renewed hope to redevelop his skil and provide self enjoyment and enjoyment to others. The adaptation consists of a slide bar that is clamped to the neck of the guitar and a small cross member that pivots and may be slid up and down the frets. This device permits the user to produce major chords and some minor chords by varying the hook interface. Different types of prosthetic terminal devices and ranges of motion may be accomodated by changing the interface. Based on the success of this device a second generation system is being designed that will enable the performer greater latitude in selecting chords by activating small solenoids.

CONCLUSIONS

Adapted equipment enables the person with a physical disability to reach his highest potential. Through cooperation among rehabilitation specialists, highly sophisticated equipment will present new opportunities for the disabled to live more meaningful and productive lives.

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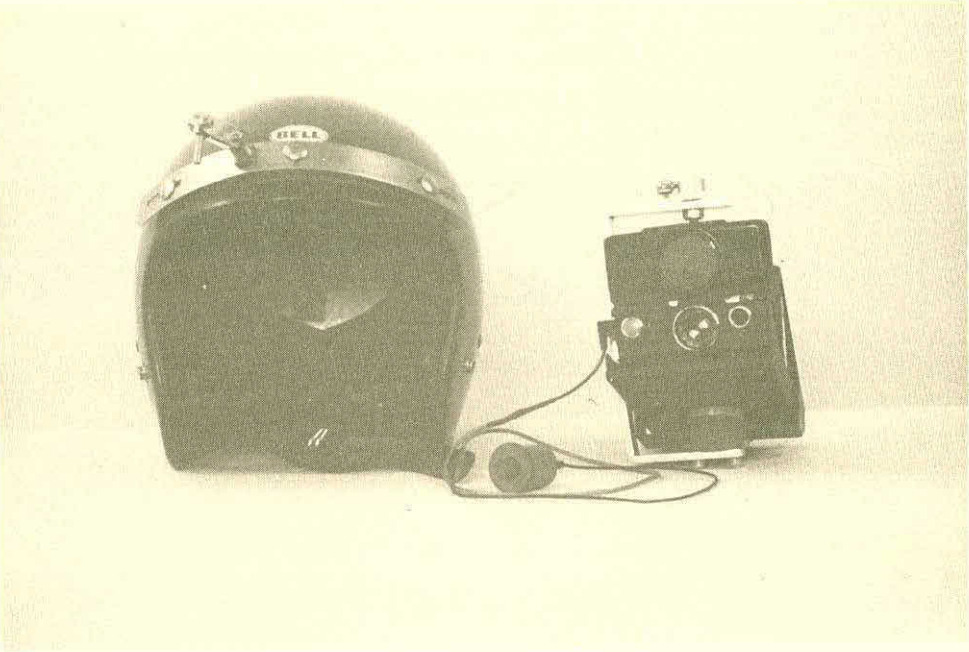
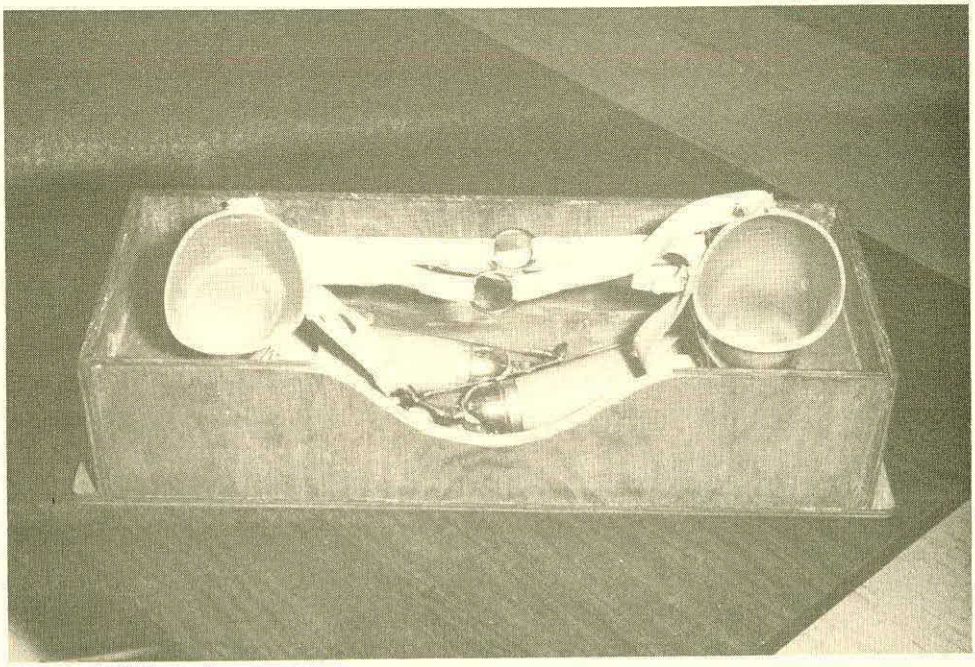


Helmet Mounted Camera System



Prosthesis Holder

EMERGENCY ALERT SYSTEM FOR USE WITH INDEPENDENT LIVING

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ABSTRACT

Independence in day-to-day living poses risks for both the able-bodied and the handicapped individual. To minimize the risk and to still allow the independence of being mobile outside buildings unattended, the system reported on in this paper was developed.

The system developed consists of commercially available transmitters, receivers, and a customized display. Four receivers, all set for the same frequency and digital code, are placed strategically throughout the property at The Timbers. Each receiver has a defined area of coverage with the response of the receivers decoded at the display. Thus, when an individual tenant has an emergency, the transmitter can be activated. This causes all receivers within receiving distance to be activated, the appropriate region on the map corresponding to the tenant's position to be lit up, and the alarm to be sounded. Thus an attendant can respond to the call for help.

This system became operational in The Timbers housing program for the handicapped in Wichita in February 1980.

INTRODUCTION

Independent living is a concept which is taken for granted by the able-bodied individual. He assumes this mode of living as his accepted way of life without regard, in most cases, for the risks which accompany this independence. However, as the risk increases, ways are sought to minimize these risks while still allowing independence, as long as the cost involved in minimizing this risk does not exceed the individual's ability to pay or the worth of the particular freedom to the individual. Thus, many times unknowingly, the relative merits of independence --cost--risk are assessed by the individual when deciding on the proper course of action to be taken on individual freedoms.

To the handicapped individual, however, the concept of independent living has been either unavailable or only recently available and not an established way of life. As a result, due to the newness of the concept and to the haste exercised to make this freedom available to all individuals, the risk associated with these freedoms are not always assessed and properly balanced with the cost involved to make the risk acceptable. In addition, the risk involved often changes with external conditions such as weather. For example, having a wheelchair become immovable under occupant control while outdoors may be an acceptable risk if the temperature is 80°F and the weather good but it may become life threatening if the temperature is 0°F with a wind chill index of -20°F. It is to this balance independence and risk and the associated cost to reduce the risk as applied to the freedom of external mobility that this paper is addressed.

PROPERTY DESCRIPTION

The emergency alert system described in this paper was developed for use in "The Timbers", a 100-unit housing project developed by the Cerebral Palsy Research Foundation of Kansas featuring independent living for the handicapped. This project consists of 10 sixplexes, 4 fourplexes and one 24-unit complex housing tenants of all degrees of handicap from very mild to very severe involvement. The tenants with the most severe involvement rent the apartments in the 24-unit complex. This complex also contains a staff room from which attendant services can be requested. The other fourplex and sixplex units border this complex on a 10-acre plot. All units are wheelchair accessible with the tenants free to come and go at will.

SYSTEM DESCRIPTION

A high degree of mobility exists with The Timbers for the tenants. All offices and apartments are wheelchair accessible with a high degree of privacy similar to any other apartment complex. This means that tenants are outdoors in almost any weather with the risk of becoming immobile always present.

With this ever present high risk in bad weather, it was considered necessary to design a system to serve as an emergency alert system to reduce this risk. Because of the variety of disabilities among the tenants, the system was to be operable without any verbal capability and with very little physical agility being required. The system must sound an audible alarm when it determines that a tenant is in difficulty and it must be capable of determining the location of the tenant with enough precision so that the tenant can be found within five minutes. It must also provide complete coverage of The Timbers regardless of buildings or other obstructions but not necessarily outside the property boundaries. It

also must not generate any interference for any existing communication system in the area nor should it give any false alarms, if possible, due to any communications in the area such as citizens band transmitters, etc.

Receivers and Transmitters

In order to achieve the desired coverage independent of building obstructions, radio transmission was chosen with multiple receivers to determine location. Various methods of determining location utilizing multiple receivers were considered too costly since they required a computer and very precise receivers and are subject to too much error over long term usage.

It was considered desirable to have a system which would exhibit high reliability with little maintenance. With this in mind, a digital system was selected which operates similar to a garage door opener. Each time the transmitter is activated, an eight bit digital code is sent by modulating a 310MHz carrier with this code. The receivers are tuned to this same carrier frequency and, when they receive a signal on this frequency, the digital code is separated from the carrier and compared to the preset code which would be sent by the transmitter which is a part of this system. If the digital code matches the preset code, a relay is activated closing a set of contacts which are normally open. Any other digital code is rejected. Since only one code out of a possible 256 codes is accepted, a high degree of isolation is obtained from any external transmission.

The determination of location can be accomplished by utilizing multiple receivers which have limited range of reception and are all set for the same carrier and digital code. The receivers are physically placed so that their circular region of reception overlap. Thus if two receivers are used, three regions are defined; the region where only receiver A is activated, the region where only receiver B is activated, and the region where both are activated. In the same manner, if three receivers are used, seven regions are definable relative to areas of overlap of their circular regions of reception. However, depending on the relative placement of the receivers, some of these regions may be too small to be of any practical use. The system designed for The Timbers utilizes four receivers with seven well defined and reasonably equal content areas. This yields, on the average, 15 apartments per area. This resolution is adequate to insure, with reasonable confidence, that any tenant in trouble will be located within five minutes after the alarm is sounded.

The transmitters utilized with this system are very similar in size and in function to those used with garage door openers. The major difference is that the activation switch is a protruding plate approximately 2 inches on a side. This allows easy activation with very little manual dexterity required. The switch makes battery contact when depressed thereby eliminating battery drain when not in use.

The transmitters and receivers described were purchased as off-the-shelf items from Microlert[1] for use with this system. The receivers were purchased at a per unit cost of $250 and the transmitters at a per unit cost of $72.[2] The receivers were modified slightly to add a battery backup power supply, to disable the latch on the internal relay, and to disable the audible alarm. These modifications were required to make the receivers utility service independent and to interface these units with the custom made display.

Display Panel

The display panel was made at Wichita State University REC laboratory. The visable surface of this panel consists of a frosted plastic plate with a scale layout of The Timbers on it. The seven distinct areas as discussed earlier in this paper are outlined on this layout for easy identification. Each of these areas has two lights which, when selected, will light that area. The panel also contains the logic necessary to decode which area should be lighted based upon which receivers are activated. It also contains an audible alarm which sounds anytime any receiver is activated. Once activated, the setting of the lighted regions will hold until a reset button is pressed.

SUMMARY

The system described has been installed at The Timbers and became operational February 7, 1980. The receivers and transmitters chosen to be a part of this system were chosen because the company which manufactures them demonstrated their ability to perform the task as specified and because they were available within two weeks. It was of extreme importance that the system become operational as quickly as possible, and while total cost was very important, it wasn't the overriding condition. As a result, other transmitters and receivers are currently being evaluated relative to cost and performance.

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[1]Microlert Systems International, Burbank, California, 91504.

[2]These prices were those quoted in December 1979 and are subject to change.

A MECHANICAL DEVICE FOR TRIPLE AMPUTEE DRIVERS

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ABSTRACT

The object of this study was to design and fabricate a mechanical device for a triple amputee driver. The device is fastened to the vehicle dashboard and connected with the amputee's left leg stump. The motion of this stump commands vehicle acceleration and braking mechanisms. The normal left hand is used for vehicle steering. A slip-proof leg harness was designed and fabricated. This leg harness fits over the subject's left leg stump and connected with the mechanical device. Extensive accelerated life-cycle tests were conducted and the device was found safe for a minimum of five years operation. The subject is presently driving the device-equipped vehicle and has obtained her Texas Vehicle Driver's License.

INTRODUCTION

Background Information

Vehicle motion control requires a complex interaction of the driver and his vehicle. Even though the physical abilities of the driver may vary drastically, a constant level of control of the vehicle motion must be maintained. One of the most difficult challenges is an approach for a prospective driver who has only one functional arm with other extremities either missing or paralyzed. Triple amputees make up a sizable minority of the physically handicapped. There are a number of commercially available hand control systems for people with two functional hands, which are designed so that both the brake and throttle can be operated with one hand and the steering with the other, but for people with only one hand there is no commercially available equipment which will allow them to drive.

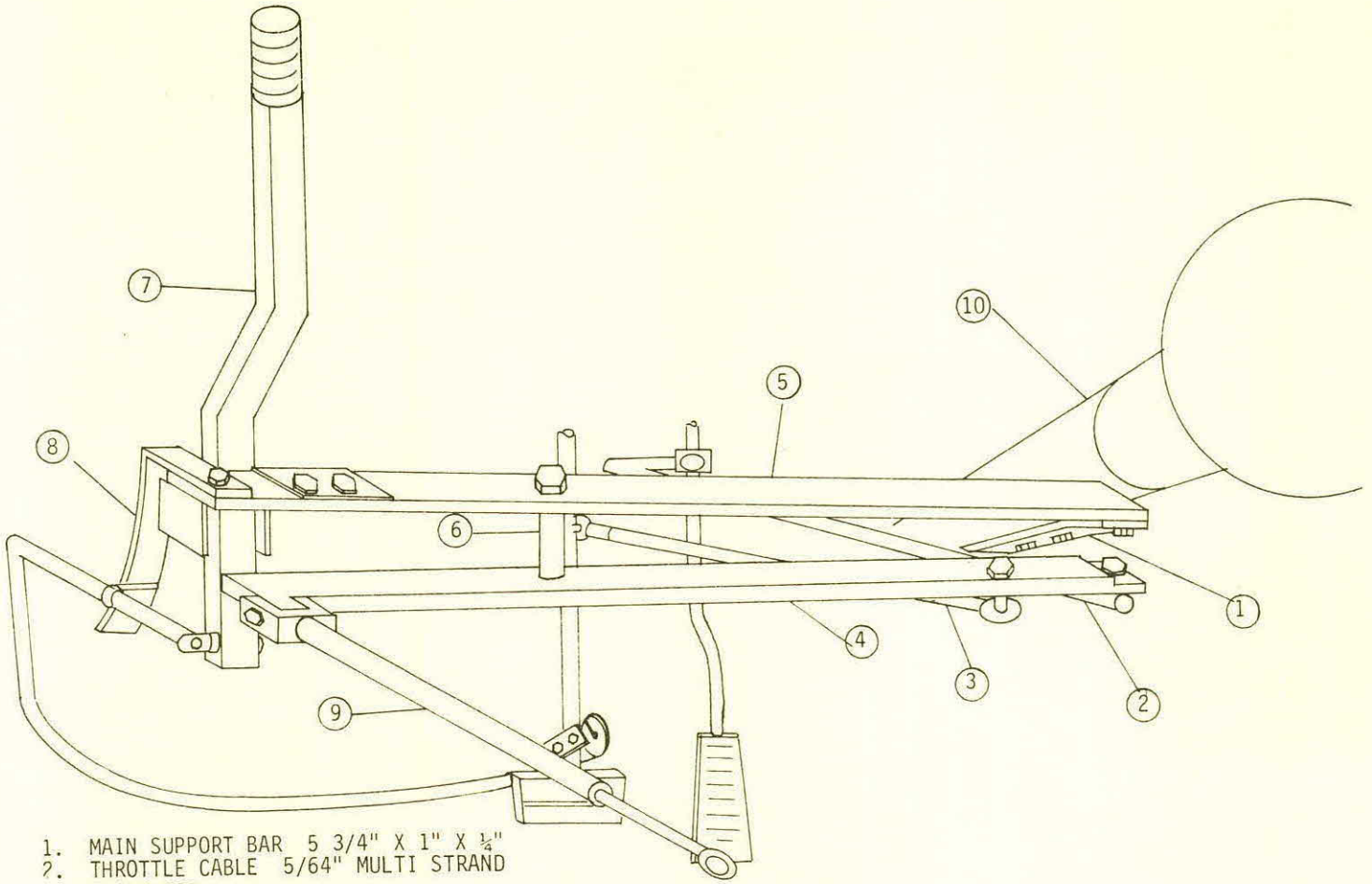
Under contract to Texas Rehabilitation Commission, Texas A&M University designed, fabricated and evaluated an experimental adaptive device which enables a triple amputee to drive. A congenital triple amputee who has a functional, normal left arm and hand and who is already a client of the Texas Rehabilitation Commission was selected by the commission to provide consultation to the project and to test the operation of a vehicle equipped with an adaptive control.

The client had never driven a motor vehicle and does not have an operational prosthesis for her missing right arm; she occasionally uses leg prostheses. She has successfully completed the classroom portion of a high school drivers education course and her left arm strength and finger dexterity are quite adequate for operating the various controls of a motor vehicle such as the gear shift lever, parking brake release, etc. She has sufficient strength and dexterity in her left arm to steer a car equipped with power steering, provided she has a good quality seat belt of the fixed adjustment type and a spinner knob on the steering wheel. She can also perform gross holding tasks with her right arm stump. The difficulty arises in using some other part of her body to provide throttle and braking effort.

Statement of the Problem. The overall purpose of this research was to design a throttle and brake control for use by a triple amputee driver, construct a prototype, test this control by destructive and non-destructive testing simulating one year's operation, and equip a motor vehicle with the control for long term evaluation of the adequacy and reliability of the design.

PRELIMINARY DESIGN

It was found that commercially available adaptive control equipment could not be modified for our purpose, therefore, a custom-made control needed to be fabricated. A device which would use a kind of yoke that would fit over the left leg stump was decided upon, the client's left leg was stronger and longer than her right, but the design is equally adaptable to right stump operation. This yoke would be connected by levers and cables to the brake and throttle pedal of the car so that leg stump motion to the left (towards the door) commands throttle and motion to the right commands brake. The motion conventions were chosen by consideration of movements envelopes behind the steering wheel. Lateral motion to the left is restricted by the door surface, whereas movement to the right can be increased by turning the torso. Since hard braking or



1. MAIN SUPPORT BAR 5 3/4" X 1" X ¼"
2. THROTTLE CABLE 5/64" MULTI STRAND
3. BRAKE ROD
4. YOKE BAR 16" X 3/4" X ¼"
5. YOKE SUPPORT 12½" X 1" X 5/16"
6. SLEEVE 5/16" ID X 1"
7. EMERGENCY BRAKE
8. SUPPORT BAR 7" X 5/8" 1/8"
9. SLIDER ROD ½" ID X 1/16" THICK 3/8" ROD
10. STEERING WHEEL

FIG 1. The Adaptive Control

possible malfunction of the braking system might result in longer travel of the control yoke, the decision was made to allocate braking to the right lateral control motion. Full throttle takes much less movement and is rarely called for in any case, thus left motion of the yoke was given to throttle. In other words, motion of the leg stump from side to side on the vehicle seat controls the speed of the car while steering is handled by the functional left hand.

A two door 1973 Chevrolet Impala was initially used to mount the control prototype for testing and feasibility evaluation. A preliminary idea to attach the accelerator cable by a system of pulleys and wires was abandoned because it decreased the sensitivity of the accelerator and increased the chances of malfunctioning.

Since the standard power brake requires a substantial amount of strength, more than what the subject could exert with her leg stump, the brakes were modified and made extremely low effort. Tests with the subject proved this modification to be satisfactory. A similar modification was made in the throttle by replacing the stock throttle return spring by a lighter spring.

THE MECHANICAL DEVICE

Figure 1 shows the design of the mechanical device. The device is supported by a main support bar (1) which is bolted to the steering column, and another support bar (5) attached to the dashboard. Brake rod (3) and throttle cable (2) are connected to a yoke bar (4). The yoke bar is pivoted on the yoke support (5) and is capable of rotating freely about the pivot. When the yoke bar (4) is pushed under the dashboard it actuates brake and when it is pulled away from the dashboard the throttle is actuated. The motion of the yoke bar is controlled by means of a leg harness which slides over the driver's leg stump. The leg harness is connected to the yoke bar through a slider rod (9). Thus, when the leg stump is moved to the right the yoke bar is pushed in, thereby, commanding the brake, and when the leg stump is moved to the left the yoke bar is pulled up, commanding the throttle. A hand brake (7) for emergency is located at the end of the yoke bar, thus, if the leg harness slips off the leg stump, the emergency brake could easily be used. See Fig. 2 for the driver-device interface.

FINAL DESIGN

For safety and appearance, the whole design was reviewed and appropriate changes were made. A secondary emergency brake was installed, this is an off-the-shelf item and was modified with special lever to make it hand operated instead of the foot pedal operation.

Prosthesist A. L. Muilenberg was consulted for a suitable leg stump harness. Two custom designed harnesses with special features were purchased. The harness has an embedded harness lock which is capable of receiving and locking a 1/4 inch quick release pin. The harness can be used either on a prosthesis or on the leg stump.

Fig. 2

The angle of the clevis extension piece was changed from the 90 degree angle to a 102 outbound angle. This angle restricted throttle travel while allowing more movement for braking, and allowing full normal braking with the left leg stump parallel to the steering column. This yoke bar was lengthened to give more mechanical advantage and its thickness was increased to allow for full braking at high speed without failure or bending.

A change in the slider rod was also made; the end connecting to the prosthesis was machined to 1/4 inch diameter to allow for failure in an accident situation, thereby avoiding a potential hazard of injury due to the equipment.

The exposed components of the device were treated with "plasti-dip" coating. This material forms a protective plastic coating for safety, appearance, and corrosion resistance on the exposed metal.

FAILURE ANALYSIS

The device was tested for reliability and fatigue strength by simulating two years operation. A test stand was designed and constructed to hold a crank-and-slider mechanism. It is assumed that when installed for actual driving the device will be used, on the average, 100 cycles per day. Therefore in one year it will go through a total of 36,500 cycles. For initial testing the adaptive control without the leg

harness was cycled 41,040 times. When no major problem developed the leg harness was installed and a total of 170,160 cycles of braking and acceleration were performed which simulates approximately 5 years of use. The results of this accelerated life cycle test indicate that the adaptive control should give excellent service.

Conclusions and Recommendations

The device is an effective and reliable piece of equipment and will enable numerous triple amputees to drive for the first time. However, the device should not be considered as the ultimate in triple amputee adaptive control as there are many improvements that can be made in the existing device and the vehicle device interface. Foremost of these is the in-dash structure of the vehicle, which should be reinforced with a structurally stronger plate to which the device can be fastened. At present, the device is fastened to 1/8" thick metal plate which is susceptible to either a fatigue failure or - in case of sudden overload - ultimate strength failure. No other vehicle structure is accessible which is stronger and could be utilized for this purpose.

It is possible to improve upon the existing device. First, by reducing the weight and frictional forces in the device; this will require an extensive experimental stress analysis for light weight components. Secondly, by keeping the operational principle of the device as it is, and using a servo mechanism, this will eliminate the frictional forces and the bulkiness in the device.

ACKNOWLEDGEMENT

The authors would like to express their appreciation to Mr. Al Muilenberg for his valuable advice and help in designing and fabricating the leg harness used in this study.

A WHEELCHAIR RESTRAINT SYSTEM FOR HANDICAPPED DRIVERS AND PASSENGERS

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A motorized wheelchair tie-down system has been developed for use in vehicles by the wheelchair bound driver or passenger. This system provides for automatic securement of the wheelchair to the vehicle by operation of an electrical switch. The design also provides a way of anchoring the seat belt to the wheelchair without loading the wheelchair frame with occupant restraining forces during a crash. Sled impact tests using a 50th percentile male crash dummy demonstrate that this device provides effective restraint of the wheelchair even when all the occupant restraint is provided by a lap belt only. More effective occupant protection is provided if an upper torso belt is used in conjunction with this system.

INTRODUCTION

Providing for mobility of handicapped adults requires not only development of equipment and devices which will allow independent operations of private vehicles, but development of adequate restraint systems to protect these handicapped drivers and occupants in vehicle collisions. Until recently, little concern and effort has been devoted to the latter area. As a result, handicapped individuals fortunate enough to have found a means for operating a vehicle are, in most cases, completely unprotected in the event of a crash.

While designing for impact protection of able-bodied vehicle occupants is a complex and difficult engineering problem, designing for the protection of handicapped individuals, with their special equipment and assisting devices as well as strength and mobility limitations, presents additional problems in crashworthiness design.

In an effort to provide an effective and automatic means of wheelchair securement and occupant protection for the wheelchair bound person, a motorized wheelchair tie-down device was developed by Creative Controls, Inc. in conjunction with the University of Michigan Rehabilitation Engineering Center. This paper describes the design concept of this device and reports on its performance evaluation based upon sled impact tests conducted at the Highway Safety Research Institute of the University of Michigan.

DEVICE DESCRIPTION

The conventional wheelchair in use today has not been designed with vehicle transportation and crashworthiness criteria as a primary concern. Consequently, it's structural members cannot, in general, be relied upon to hold up under the large inertial forces required to restrain both the wheelchair and its occupant under vehicle crash decelerations. Yet, for the wheelchair bound person, who has limited strength and mobility, it is advantageous to provide occupant restraint anchorage points on the wheelchair itself. It is also important for the person who wants to operate a vehicle from his wheelchair without assistance, to have a restraint system which is actuated automatically or by the simple action of a switch. It is with this perspective that the Creative Controls wheelchair tie-down and restraint system illustrated in Figure 1 was developed.

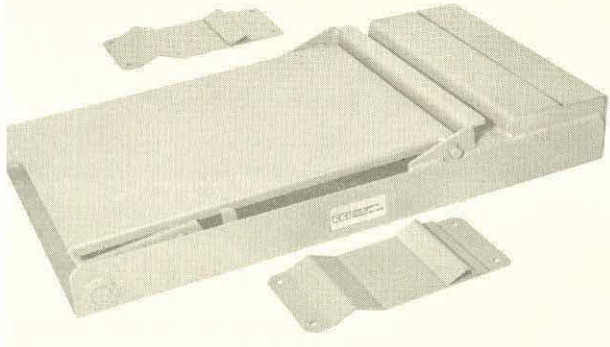

Figure 1. Creative Controls wheelchair lockdown platform

This work was sponsored in part by the University of Michigan Rehabilitation Engineering Center supported by Grant No. 23-P-59227/5 of the Rehabilitation Services Administration, Office of Human Development of the Department of Health, Education and Welfare.

The conventional wheelchair is modified to include two triangular shaped side plates shown in Figure 2. These plates are easily installed by removing the large wheels and clamping the plates to the tubular frame by two sets of U-bolts. The front and rear lower corners of these plates are provided with holes through which two steel retaining rods are inserted and held by means of retaining pins. These rods can be easily removed to allow folding of the wheelchair. Holes located in the upper part of these plates provide solid anchoring points for the lap belt so that the belt is properly positioned across the occupants' pelvis.

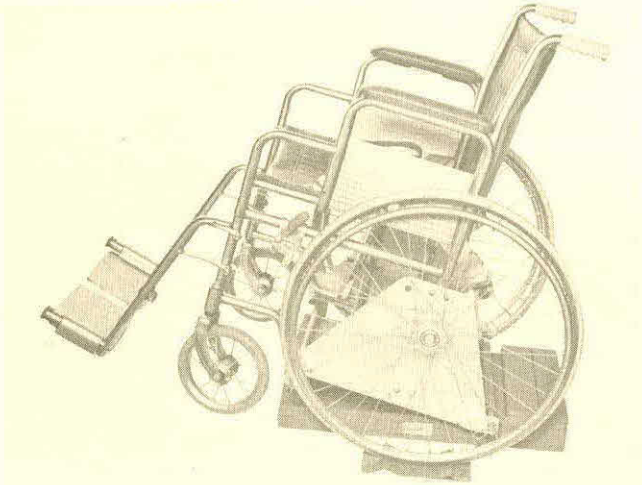

Figure 2. Wheelchair modified with triangular plates secured by CCI platform

Equipped with these plates and retaining rods, the wheelchair is secured in position by the locking platform shown in Figure 3. When the wheelchair is positioned over the platform, an electrical switch will trigger a motorized actuator linkage which grabs and lifts the front retaining bar and, in the process, pulls and tilts the wheelchair so that the rear retaining bar seats securely into its attachment position as illustrated in Figure 3. In the locked position, the chair and occupant are completely supported by the triangular side plates and bars and therefore the wheels of the chair are not subjected to any loads during operation of the vehicle.

For passengers and drivers below the average male stature, the system can be used as shown in Figure 2. In this case the locking platform is simply bolted to the vehicle floor with a plate under the vehicle to distribute the loads. For taller drivers the chair must be lowered for better driver visibility and access to steering and hand controls. This is accomplished by the platform illustrated in Figures 4 and 5. With this system, the wheels of the chair are lowered into wheelwells as the powered linkage grabs and lifts the front retaining bar and tilts the chair and

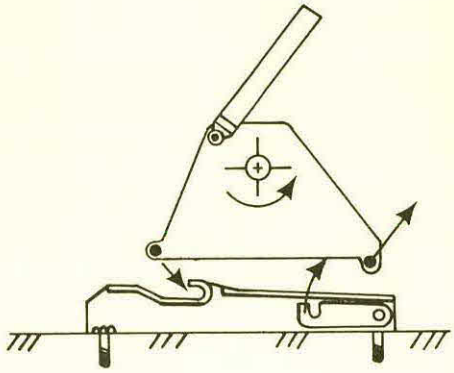

Figure 3. Illustration of Creative Controls wheelchair lockdown mechanism

rear retaining bar into the locked position. With this system also, the chair is completely supported by the retaining bars and in the final locked position the occupant is in a position similar to that provided by a standard vehicle seat. This driver restraint platform must be securely welded in place after cutting out an appropriate section of the vehicle floor.

Figure 4. Wheelchair ready for securement by driver restraint system

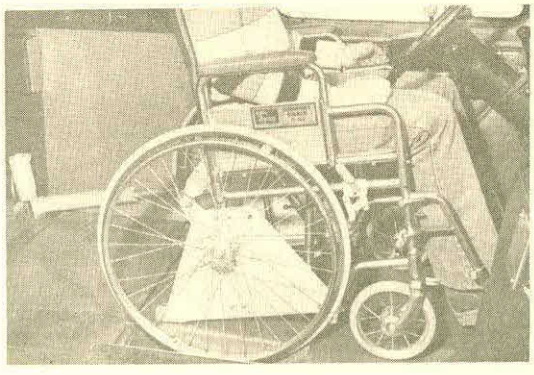

Figure 5. Wheelchair locked into position by driver restraint platform

IMPACT TESTING

An important aspect in the design of any new restraint system is the evaluation of its potential crash effectiveness by dynamic testing. The Creative Controls platform previously described has been subjected to a series of impact tests at the Highway Safety Research Institute rebound sled facility. These tests were conducted using a standard Everest and Jennings wheelchair modified with the triangular plates and a 50th percentile male crash dummy (HSRI) weighing 165 lbs. The sled impact pulse was a 20 mph velocity differential generated by a 16 G (16 times gravity) constant deceleration pulse. Tests were conducted for both frontal and 33 degree oblique impact directions.

Figure 6 shows the test set-up for the initial tests in which the dummy was secured by only a lap belt anchored to the steel triangular plates. Figure 7 shows the post test result in which it is seen that the chair was effectively restrained during the impact. The rear retaining rod absorbs the majority of the inertial forces as indicated by the bending of the rod shown in Figure 8. The only damage to the wheelchair itself was bending of the front castor shafts caused by downward loading on the chair by the dummy and the fact that the front retaining rod pops out of its retaining bracket as the rear bar deforms during impact. Similar results were obtained for the 33 degree oblique impacts and good wheelchair restraint was provided with little damage to the wheelchair itself.

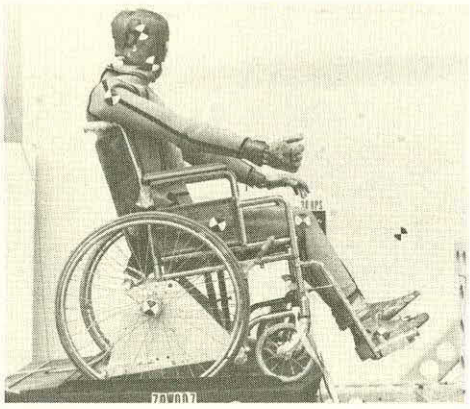

Figure 6. Sled test set up for CCI platform with 165 lb. restrained by lap belt

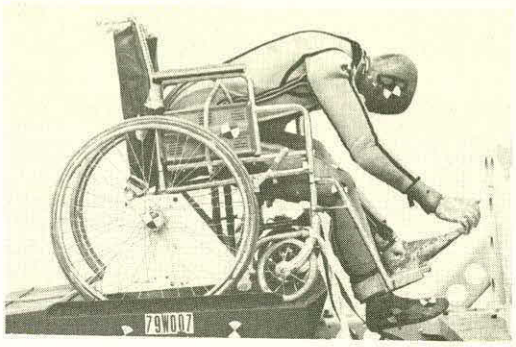

Figure 7. Photograph of post-impact conditions

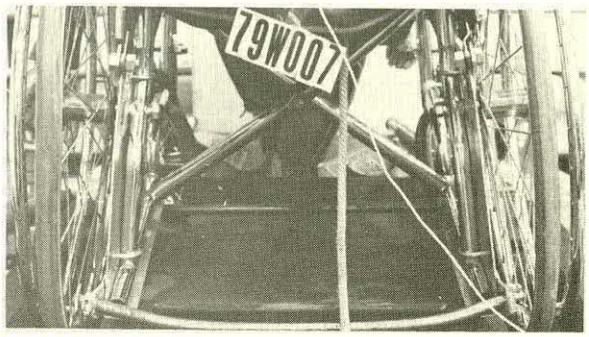

Figure 8. Rear view of wheelchair showing bent retaining rod after impact

FORCE ANALYSIS

Figure 9 shows a simplified free body drawing of the forces generated during a frontal crash and points out the advantage of utilizing these triangular plates attached to the wheelchair frame. As shown, the primary forces during the impact are the forces on the seat belt caused by deceleration of the occupant (F_{belt}), the force of the occupant downward on the seat cushion (F_{seat}) caused by pivoting action of the occupant around the belt, the inertial force of the wheelchair (F_m), upward forces on the front wheel castors (F_w), and the forces exerted on the rear restraining bar by the locking platform (F_{bar}). No forces are shown on the front retaining bar since this pops out during impact causing the vertical loading to be primarily taken by the front castors. If we make the assumption that F_{seat} and F_w are primarily in the vertical direction (i.e., frictional forces are negligable) then the only significant forward restraining forces imposed on the wheelchair frame are due to the mass of the chair and these are distributed at the interface with the triangular plates. The more substantial loading due to the occupant's mass is transferred from the lap belt (F_{belt}) through the triangles nearly directly to the rear retaining bar (F_{bar}) and is not imposed on the wheelchair frame.

The forces on the rear retaining bar can be roughly calculated from a resultant force diagram shown in Figure 10 in which it has again been assumed that F_w and F_{seat} are vertical and equal. F_{bar} then becomes the summation of the F_{belt} and F_m vectors. For the tests described above a typical peak lap belt tension was 920 lbs. If we assume that F_{belt} forms an angle of 45 degrees with the horizontal, a wheelchair mass of 40 lbs. and a peak wheelchair deceleration at the time of peak belt force of 12 G's (it will be less than sled G's due to deformation of rear retaining bar),

then the force on each end of the rear retaining bar is approximately 1100 lbs.

$$\vec{F}_{bar} = \vec{F}_{belt} + \vec{F}_{m}$$

$$F_{bar} = \sqrt{(\Sigma Fx) + (\Sigma Fz)}$$

$$F_{bar} = \sqrt{[650 + \tfrac{1}{2}(40 \cdot 12)]^2 + (650)^2}$$

$$F_{bar} = \sqrt{(650 + 240)^2 + (650)^2}$$

$$F_{bar} = \sqrt{890^2 + 650^2}$$

$$F_{bar} = 1102 \text{ lbs.}$$

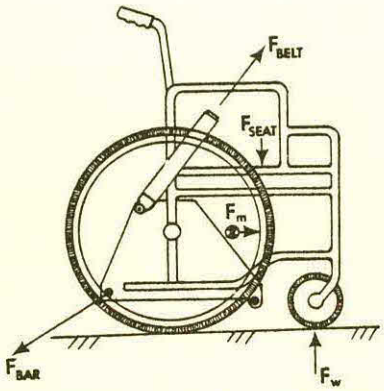

Figure 9. Simplified free-body diagram of forces on wheelchair and restraint system during a frontal impact

In contrast the force imposed on each side of the wheelchair frame by F_m is only:

$$F_{frame} = \tfrac{1}{2}(40 \times 12) = 240 \text{ lbs.}$$

and this is distributed at the two U-bolts and over much of the frame tubing by frictional forces.

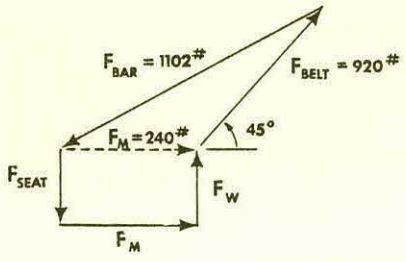

Figure 10. Force diagram of loading during frontal impact

UPPER TORSO RESTRAINT

While the Creative Controls lock-down platform provides effective restraint for the wheelchair under the conditions described above, improved impact protection for the occupant can be achieved through the use of an upper torso restraint which will prevent serious injury due to contact of the head and chest with vehicle structures. Figures 11 and 12 show two approaches to providing this additional restraint. In Figure 11 a single shoulder belt is anchored at the upper end to the vehicle and at the lower end to the wheelchair. In Figure 12, a Y-yoke harness is anchored to the vehicle roof above and behind the occupants' head and to the two triangular plates on the wheelchair. Both systems provided excellent upper torso restraint during frontal impact testing, and, together with the Creative Controls wheelchair restraint, can provide effective impact protection for wheelchair bound persons in frontal crash situations.

It is also interesting to note that peak lap belt tensions for tests with upper torso restraint were in the range of 400 lbs. compared with 920 lbs. when the lap belt alone was used. The advantages in terms of less severe occupant loading and reduced forces on the wheelchair restraint are obvious.

The need for automatic or passive upper torso restraint for many individuals, in addition to the automatic wheelchair tie-down, is well recognized and work is currently underway to determine an effective way to achieve this.

Figure 11. Dummy restrained by lap belt and shoulder belt

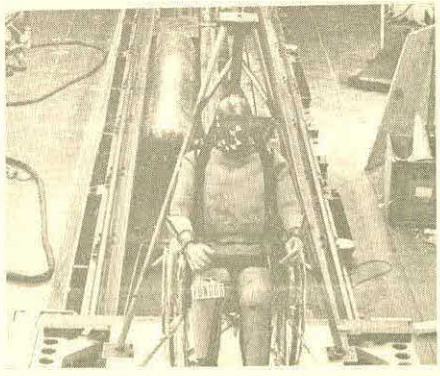

Figure 12. Dummy restrained by lap belt and Y-yoke harness

DRIVER'S TRAINING PROGRAM: A METHODOLOGY FOR DETERMINING THE TRAINING NEEDS OF INDIVIDUALS WITH PERCEPTUAL DISABILITIES

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ABSTRACT

This study examines the potential of interactive driving simulators for assessment and training purposes. The results indicate that simulators do not adequately assess every individual's driving skills. However, the simulator does have the strong correlation with on-the-road driving performance. It is concluded that simulators may be useful for (1) appropriately placing an individual in training programs and (2) providing individualized training.

INTRODUCTION

The use of testing aids for assessment of an individual's capabilities is a topic of growing concern in the scientific community and among persons who feel that assessment tools may someday be used to deny people the right to participate in necessary activities. Assessment efforts at the University of Michigan for driving activities have therefore begun by reviewing the information available on the development of driving or perceptual-motor skills to obtain an objective viewpoint. Several facts stand out in the literature, including: (1) feedback is necessary for learning to occur; (2) perceptual motor skills require a substantial amount of time to learn; (3) initial performance on a perceptual-motor task is not a good predictor of final performance after substantial training, and (4) transfer of learning increases with increasing similarity between the training device and the automobile (Salvendy and Seymour, 1973; Welford, 1968).

The necessity of feedback for the acquisition of perceptual motor skills may be understood by considering that a person cannot learn to perform better unless she or he knows what was done wrong. Knowledge of what was done wrong allows a person to try alternative strategies and select the strategy which is most appropriate. Although this fact is clear and unrefutable, many high schools across the nation use driving "simulators" which do not respond interactively to the student's control inputs at the steering wheel, accelerator or brake. This is a violation of the most basic principle of learning, i.e., feedback or knowledge of results.

The fact that learning perceptual motor skills takes substantial training is well documented. Studies of such "simple" jobs as dipping cherries in chocolate show that people are still learning after several years (Crossman, 1959). The rate of acquisition of perceptual-motor skills may be so slow as to be imperceptable. Furthermore, a handicapped individual may not have had the opportunity to develop some of the skills commonly acquired in childhood which may transfer to the more complex driving task. Thus, we should expect and not be surprised that training may take a substantial length of time for some handicapped students, and this fact should be incorporated into the design of our training programs.

It is well known that initial performance on a perceptual motor task is a poor predictor of final performance. For years psychologists have tried to predict an individual's performance based on assessment tests. For perceptual-motor skills initial performance typically explains less than twenty-five percent of final performance after substantial training (Salvendy and Seymour, 1973). This is a small amount and certainly cannot be used to deny an individual access to well designed training programs. Now critics may claim that this figure is false since it is based only on the so-called normal population. While this is true, it is equally true that there has been no proof that assessment tests will be any better for the handicapped population while thirty years of study have failed to produce a reliable predictive assessment test for perceptual motor skills. Thus, it has been shown that the only way to find out if a person will be able to perform a perceptual-motor task is to train that individual for a substantial period of time and then assess his skills. Reliable prediction cannot be made prior to training. Therefore, the only tenable position is that assessment tests must be used only to determine the proper training procedures and appropriately place an individual.

It is apparent that transfer of learning should increase as the training apparatus more closely resembles an automobile. Perhaps less obvious are the important characteristics which must be included in the training apparatus to assure that the appropriate skills are learned.

Interactive feedback, of course, is necessary. This means that the simulator must take the trainee's control inputs, process these inputs in a manner similar to an automobile and instantaneously display these results on a television or other display unit. Redundant or additional kinesthetic information may be supplied through lateral motion of the simulator, but the visual feedback is a minimum. This is sufficient for the trainee to develop a "feel" for the control. Control "feel" is just the ability for the driver to correctly predict the future response of the automobile relative to the road when a control action is implemental at the wheel. The person has to "know" what the car will do several seconds down the road. To learn control feel a person has to be able to try alternative strategies and immediately see the response in the simulator.

PURPOSE

The use of driving simulators as an assessment and training tool is based on the assumption that the perceptual-motor skills required or developed on the simulator are similar to the skills required to drive an automobile. The purpose of this study is to determine the degree to which driving performance in a simulator is related to driving performance in an automobile. A strong relationship between performance in the simulator and the automobile would indicate that the driving simulator may be useful as an assessment and training tool.

METHODOLOGY

A driving simulator (Boydstun and Kessel, 1980) and road test (Sivak, et al., 1980) were used to assess the driving skills of sixteen subjects. The driving simulator is a fixed base simulator. Fixed base simulators are less expensive and more reliable than moving base simulators and are therefore more likely to be chosen for assessment and training purposes.

The road test was a ten-mile course in a suburban area. A qualified driving instructor directed the driver while a passenger recorded observations of driving errors. Driving errors include incorrect stops at stop signs, fast or slow driving speed, failure to check the rearview mirror prior to a lane change and unsafe maneuvers.

Eight subjects were student volunteers and eight subjects were handicappers with perceptual disabilities. Perceptual disabilities included cerebral palsy, spinabifida and Friedreich's disease. Six of the eight handicappers had some training with a qualified driving instructor prior to participation in the study.

RESULTS

A plot of the data (Figure 1) indicates a high correlation ($r = 0.88$) between the simulator and road test error scores. The simulator appears to assess current driving skills and may be useful in assessment and training programs.

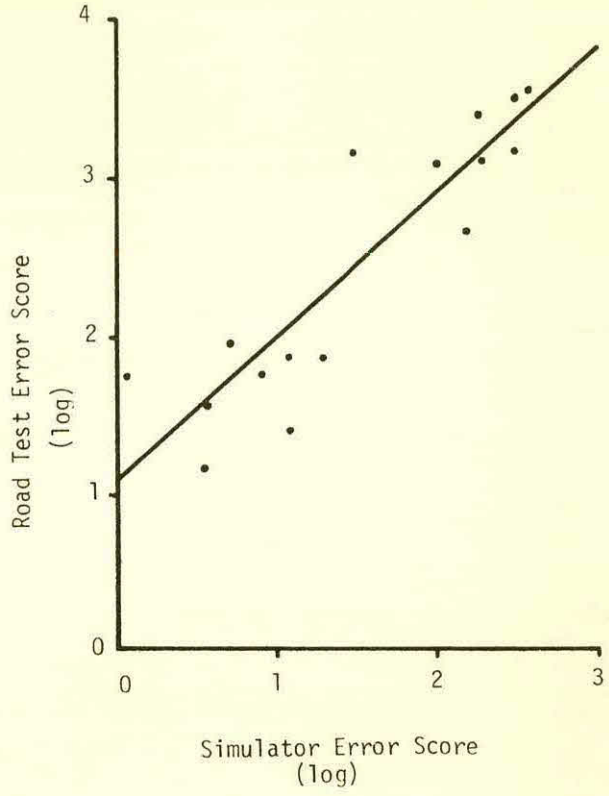

Figure 1: Relationship between road tests and simulator performance scores

It should be noted that two subjects (who were not included in the analysis) did not complete some portion of the experiment. Thus, the simulator will not provide a good assessment for every individual. For example, one of the subjects who was not included in the analysis performed well on the road test but did not complete the simulator portion.

DISCUSSION

The results of this study indicate that the simulator provides an assessment of current driving skills. This assessment should be used to select an appropriate training program for the individual. The training program may include detailed instruction of driving rules, training in a simulator and parking lot tests. The more advanced student may proceed immediately to on-the-road training. Any error in assessment will not penalize the individual and rapid progress by a student will be evident if he or she is inappropriately placed in an extensive training program.

The training benefits of the simulator have not been determined although the simulator appears promising based on the results of this study. Driving simulators are also attractive since a large amount of individualized training may be conveniently provided for each student. Simulators are commercially available and provide a safe learning environment.

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REHABILITATION ENGINEERING AS THE CROW FLIES - SUMMARY

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Abstract

The general goal of rehabilitation engineering, the main results which must be aimed for and what information areas are significant for problem solving are outlined. A framework for planning activities and using time is provided with an example case. How the technologist differs from the clinical practitioner is indicated in a discussion of time-use.

Engineering principles and procedures applied to the problems of disabled people and their supporting services constitute rehabilitation engineering. The process integrates a wide variety of inputs ranging from clinical to industrial according to the needs at a particular stage in the process. Rehabilitation Engineering starts with the identification of a problem and ends with the establishment of a device, technique, procedure or recommendation which advances the manner in which services to the persons in need are delivered.

To meet the goal of rehabilitation engineering activities, one or more of these results must be achieved:

(a) the target group must experience improved quality of life.
(b) personnel serving the target group must function more effectively
(c) there will be improved cost effectiveness of services
(d) there will be new insights which have use-potential.

For such outcomes to be possible, rehabilitation engineering must be applied so that populations of disabled persons and the service system supporting them are affected. The clinical practitioner, who deals with specific persons as clients may not appreciate this factor. To affect groups or populations of disabled persons the rehabilitation engineering function must be applied to a variety of disability situations in an environment large enough to offer a suitable population base for support of such an effort. Approximately one technologist per 200,000 of general population is a reasonable ratio, and the setting would be one in which the process would be carried out in cooperation with the clinical people of other specialties and in cooperation with other technical people.

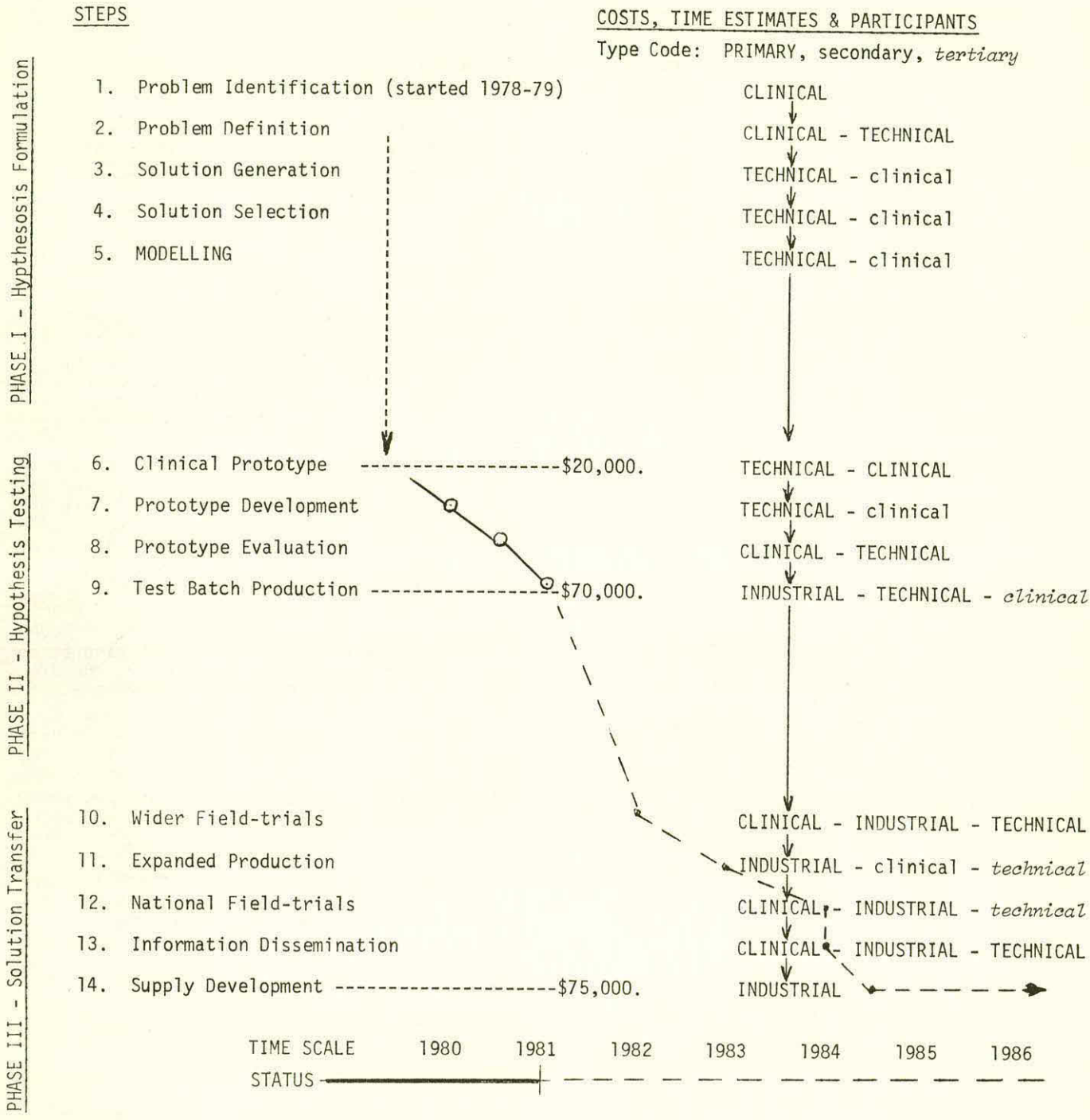
The technical team must be sufficiently large to have within it electrical engineering skills, skills in mechanical engineering, probably skills in Kinesiology as well as having access to or inclusion of related technician skills, including skills in prosthetics and orthotics. The skills must be of the highest order and it must be possible for the people of various disciplines to cross each other's boundaries effectively and without constraints so that dependence on specific people for specific functions within the group wont become problematic. When well integrated and technologists operate cooperatively with each other and with clinical persons dealing with the disabled, the potential for effective problem-solving exists. The obligation of the various people involved in the process, technical and clinical is to learn from and teach each other. For such prospects to be fully realized, it is best that the technologists have sufficient autonomy to permit them free movement across institutional boundaries. They must be able to function wherever there are problems because of the highly specialized nature of the rehabilitation engineering function. Such autonomy must not mean segregation however, but rather the assurance of a sufficient population base to warrant the service and access to sympathetic and clinical people wherever they can be found.

Considerable institutional support is needed however. There must be a base or bases at which various functions can be carried out. There must be access to clients in an ethical way. There must be access to other skills. There must be access to various services such as are offered by machine shops, electronic shops, plastic shops and the like. There must be suitable facilities available for information derivation and processing. The administrative structure which will allow this is one that takes

Figure 1

TIME-SCALE PROBLEM-SOLVING FRAMEWORK

PROJECT..EXAMPLE.....



into account the unique and specialized nature of innovation and research as part of problem-solving. There are at least six areas into which enquiries may be needed in order to make the problem-solving process effective:

(a) SHAPE of anatomical parts
(b) MOTION of body segments around joints
(c) FORCE against body parts
(d) NEURAL function
(e) MATERIALS properties
(f) PSYCHOLOGICAL factors

Such information will be sought firstly in the literature and from other team members. When such information is crucial to the solution of problems and is unavailable, then sufficient research to satisfy the problem-solving operation might be required, or intelligent estimates. This view conflicts with the traditional view that the technologist is a researcher (which has evolved from the origins of the application of engineering to problem-solving for disabled people). That does not mean to say that there is not a scientific basis in the proceedings of rehabilitation engineering. In fact, a suitable line of attack on problems includes the usual steps - hypothesis formation, hypothesis testing and information dissemination - balanced within a reasonable time frame and tailored to a realistic budget. The budget and time scale will suit the nature of the problem, but the steps toward resolution of the problem will approximate the problem-solving framework shown in Figure 1.

In the illustration provided, you see depicted the shifting responsibilities between the three principle groups participating in the example - clinical, technical and industrial. You see how $20,000 has been spent already to reach Step 6 of the first phase, that of hypothesis formation. $70,000 available is intended to bring the project through Phase II to Step 9. After that, estimates are that an additional $75,000 will be required to complete transfer (Phase III) of the solution into manufacturing for clinical use and that this will be completed by the end of 1984. During Phase III, when the solution is being transferred into practice and is being made available for general clinical use, the various participants will be reporting in appropriate publications on the results achieved so that the information is made generally available.

While the TIME-SCALE PROBLEM-SOLVING FRAMEWORK layout shown in Figure 1 relates specifically to a device, adjustment of the terms and selection of a particular phase block could be a part of planning a different operation, such as the evaluation of an existing system. Similarly, re-cycling can be done, as when new technology, new ideas, or new information can be taken advantage of profitably. But at all times, persons who are responsible for planning, or for whom the work is being done, or by whom it is being funded and others who are also engaged in rehabilitation engineering or treatment, will know just where a project stands in time and costs as well as where it can be expected to be. What stands out clearly in such a time scale is just how protracted rehabilitation engineering processes can be and how costly they can be. Such skills can only be fitted into the clinically familiar 20 minute time slot when all the facts are in and the means finalized. Then action is no longer the function of rehabilitation engineering, but must be in the hands of clinical people as part of their routine processes. At such a stage, all involved groups concentrate on disseminating information in relation to their own aspects while rehabilitation engineering processes are brought into focus on new problems.

While costs, such as those indicated in the example, seem great in relation to the population of disabled to be helped, when a good solution has been generated and brought into clinical service, it will endure for many years, bringing the cost of the engineering aspect down to some reasonable proportion of the rehabilitation costs. In the particular example serving as a model for this discussion, cost of not having a solution far outweighs the cost of developing the solution, and the proposal to develop it then displays both technical competence and social responsibility.

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A SYSTEMATIC APPROACH TO REHABILITATION ENGINEERING DESIGN

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ABSTRACT

Rehabilitation Engineering design problems require the rehabilitation team to work together to reach suitable solutions. A systematic approach to rehabilitation design problems permits all the team members to work together towards effective problem solutions. The suggested procedures build upon engineering design principles with emphasis upon their particular application to the interdisciplinary problems of rehabilitation engineering. Idea generation, product liability, and group dynamics are given special emphasis. It is recommended that the rehabilitation professional and consumer should work together throughout the whole project to produce acceptable solutions to rehabilitation engineering design problems.

INTRODUCTION

The purpose of the paper is to provide a systematic approach for all the rehabilitation team members to follow in solving rehabilitation engineering problems. Since the rehabilitation team is interdisciplinary it is important for all the team members to be familiar with the design process. The suggested procedures build upon engineering design principles, but with emphasis on their application to the interdisciplinary problems of rehabilitation engineering. The main phases in the design process are: problem definition, literature review, generation of ideas, synthesis and analysis, and solution presentation. Within the basic framework the designer must continually evaluate, document, plan, control, and consider economics and product liability.

PROBLEM DEFINITION

Problem definition includes problem identification and formulation. Rehabilitation engineering problems have psychological, social, physiological, and technical aspects, therefore the whole rehabilitation team and the consumer must be involved in problem definition. The rehabilitation team must determine what is the cause, or merely a manifestation of the problem, the generality, specificity, or components of the problem, the available information about the problem and the resources available to solve the problem. For problem definition and formulation, there must be very good communication between all the members of the rehabilitation team, including the consumer, so that there is a clear understanding of the total problem from all viewpoints. Needs assessment of the user population, which is the first stage of the evaluation process, is also begun at this time to determine the number of potential users of the problem solution and consequently the scale of the project.

LITERATURE REVIEW

Once the problem is understood, a literature review is done to determine the current state of the art, or if there is an available or apparent solution to the problem. There is a broad spectrum of sources of rehabilitation engineering information, but many investigators have found that due to a lack of proper documentation of research and poor cross-referencing it is very difficult to retrieve the technical information. Additional information is also available in many commercial catalogues. Recommendations have been made to the new National Institute for Handicapped Research for centralization and organization of information. This combined with the use of computers should improve the dissemination of rehabilitation engineering information in the future.

IDEA GENERATION

When a literature review does not provide an immediate solution to the problem, creative and innovative ideas must be generated to solve the problem. Four methods of idea generation: check-lists and attribute listing, a systematic search for design parameters, brainstorming, and synetics(1) are commonly used in rehabilitation engineering. Checklists and attribute listing are useful for the modification of existing solutions. Checklists provide a list of questions such as how can ideas be improved in quality or what is the technical base of the idea, and are there other technical bases that might work equally well. Attribute listing or value analysis is another technique of improving existing solutions. Value analysis is based on the assumption that most ideas are merely extensions or combinations of previously recognized observations, and this involves: listing the key elements and main features, functions or significant attributes and finally systematically modifying or eliminating each feature.

Design Parameters

A systematic search of design parameters considers all possible combinations of given conditions or design parameters. The success of this technique depends upon the identification of the significant parameters that affect the design. The steps in the search are as follows: (i) make the description of the problem as general as possible, (ii) select the major independent variable conditions required to describe the characteristics and functions of the problem, (iii) list the alternative methods that satisfy each independent variable condition, and (iv) establish a list of all the possible combinations of the conditions.

Brainstorming

For problems that have not been encountered before and require a novel approach, brainstorming and synetics are useful techniques for generating new ideas. The objective of brainstorming is to stimulate ideas and not to generate complete solutions to given problems. Brainstorming involves compiling all the ideas a group can generate while deferring any judgement on their worth. This releases the imagination of all the participants from restraints such as fear. The participants should suggest new ideas or combine, alter, modify and add to ideas as they are suggested. In rehabilitation engineering, the brainstorming team is made up of consumers and their families, therapists, doctors, nurses, teachers, social workers and engineers. Therefore it is important that all judgement or evaluation be kept out of the session, so that the participants do not feel inhibited.

Synetics

Synetics is a technique that attempts to bring about a solution by drawing unrelated ideas together and forcing them to complement one another. The synetics participant tries to imagine himself as the personality of the inanimate object. The key to this technique is the leaders' ability to make the team members "forcefit" or combine unrelated ideas into a new and useful solution. Synetics uses various psychological states that are involved in creative acts and usually draws the team members from diverse fields of learning so that the group spans many areas of knowledge. The members must be imaginative but not necessarily experienced, since a novice has fewer preconceived ideas. However the team should include at least one expert who fills in the facts and is often the "devil's advocate" since he can point out weaknesses of an idea. In rehabilitation engineering problems there is a psychological advantage for a potential consumer to be involved in both the brainstorming and synetics sessions, since he is actively participating in and contributing to the creative aspects of a solution to a problem and alternatively has much more concern for the success of the project.

SYNTHESIS

Once a variety of ideas has been generated it is then necessary to determine the optimum combination of ideas. This very difficult part of the design process is called "synthesis" and must take into consideration medical, social, economic, and psychological aspects of the problem in addition to the technical aspects in order to bring all parts of the problem together such that they compliment one another to comprise an effective solution. For this phase, synthesis, assumptions are made which link the reality of the problem to models. Models are simplified versions of complex systems and are used to make visualization, mathematical analysis, or testing more practical. Models are constructed and tested by using the technical and medical tools of analysis and diagnosis respectively to determine the optimum combination of ideas. The "art" of making a model consists of selecting the most appropriate degree of simplification and assessing the importance of the effects that have been neglected. Standard engineering modeling techniques are used for most aspects of rehabilitation engineering design problems.

TESTING

Tests and experiments must be performed on the initial assumptions, the prototype, and the materials. Four types of tests that are useful in rehabilitation engineering design problems are performance testing, quality assurance tests, life endurance and safety tests, and human acceptance tests. Human acceptance tests are very important for rehabilitation engineering. The designer must be concerned with the interface of the device as well as the physical, mental and emotional needs and limitations of the human operator. A design must be acceptable and satisfy the technical and economic constraints of the problem.

ADMINISTRATION

Throughout the whole project planning, control and accounting must be done to administer the project and account for the appropriation of funds. For rehabilitation engineering design projects the economics are determined by low production volume and specificity of design. A budget for rehabilitation projects must include the regular items but provision must also be made for the added costs of technical experts, consultants, evaluators, legal advice, information dissemination, documentation, inspection and testing. Ongoing documentation of the project should be done so that all the team members can be familiar with all aspects of the project at any period of time for the duration of the project. Evaluation, which begins with the needs assessment and is carried on until an impact study of the manufactured product on the user is completed, is an integral part of the design process. There are six steps within the design

process where evaluation should be done: an initial needs assessment, impact studies of the device on users, prototype design development, testing and evaluation of a prototype with regard to user impact and industrial potential, marketing studies, and the evaluation of impact on users over a long period of time. The most important aspect of evaluation is to ask the correct questions.

LEGALITIES

In rehabilitation engineering design problems there must be careful attention to government regulations, details and proper engineering standards to produce both efficient and effective designs that reduce the risk of liability due to negligence. Liability is a very important facet of professional responsibility. A rehabilitation engineer is no longer protected from liability either by a physician's prescription or a hospital's release form. A designer or any member of the team may be liable for negligence resulting from any forseeable or unforseeable use of the product. The far reaching implication of a professional's liability must be understood by all members of the rehabilitation team throughout the whole design process and not just those concerned with the final product.

TEAMWORK

In rehabilitation engineering design problems it is necessary for the rehabilitation team to work together to reach a suitable solution and therefore, there should be consideration for the personal interaction of all the team members. The team members assume certain "roles" as a result of professional stereotyping and labeling, and consequently there are certain expectations and prejudgements of the contributions of the team members. It is important for the team members to have reasonable expectations of the "roles" each can assume and the contributions which can be made to the problem definition/solution. For example an engineer may unrealistically expect a physician or a therapist to be able to describe a disability in quantitative terms, or a medically trained person may expect an engineer to provide answers to a broad range of non-trivial technical problems. Thus, it is important for the team members to have a knowledge of and respect for each others' area of expertise in order to maintain a cooperative relationship. It is suggested that the team members initially discuss their background, roles and the scope that they see for themselves as well as those of others on the team. Open lines of communication are essential in order to minimize hard feelings which can develop when one member sees another member over-stepping his "boundary" or not carrying out his share of the work. There is often an innate hierarchy within the rehabilitation team (which may be real or imagined by one or more team members). The hierarchy may produce domination and subordination complexes, which should be minimized in order to avoid an adverse effect on the contributions made by the various team members. The interaction of each team member is determined by the particular personality and the atmosphere produced by all the personalities. The protocol of a team depends upon the personality and context of the team as a whole. The protocol or ground rules, which all members should be aware of, determine how the team members will interact. It is the interaction of the team members and their ability to communicate and share resources which will determine the success of the systematic approach to design.

THE CONSUMER

An aim of rehabilitation engineering design is to produce designs of products that are profitable and thus attractive for manufacturers to manufacture. It is recommended that recognition of the particular needs and concerns of the potential consumer should be applied throughout the whole design process to produce solutions to rehabilitation engineering design problems that are acceptable and satisfy the technical and economic constraints.

ACKNOWLEDGEMENTS

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A PROPOSED EVALUATION METHODOLOGY FOR REHABILITATION ENGINEERING DECISION-MAKING

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ABSTRACT

What evaluation data does a potential manufacturer actually require to make the decision to produce and market a rehabilitation engineering research item? Inadequate, incomplete evaluation of research and development items has resulted in few items being adopted for manufacture by industry and consequently produced and made available to the ultimate consumer for which it was originally conceived-the handicapped population. This paper proposes a model for the comprehensive evaluation of a research item which will produce decision - making information for the ultimate manufacturer as well as the developing agency, and will thus expedite the flow of research technology from the laboratory to the marketplace.

INTRODUCTION

The medical and engineering technological advances of the past decade have produced a variety of new devices, new products, and new techniques which have potential to enable handicapped individuals to improve their quality of life. Funding from the National Institute for Handicapped Research in 15 Rehabilitation Engineering Centers (12 in the United States and 3 overseas) has provided the impetus for developing methods of applying available technology to increase levels of functioning for the handicapped. Other publicly and privately funded organizations and university engineering departments are also involved in ongoing research to develop relevant technology for the handicapped consumer. Similar programs have been established in Canada, Sweden, United Kingdom and several other western countries.

A critical gap exists, however, between the research laboratory setting of technological development and the actual delivery of manufactured devices/products designed with the ultimate purpose of making the handicapped person more independent and employable. Few research items are actually adopted for manufacture by industry for the following reasons: (1) adequate evaluation of technology providing necessary data for decision-making has not been executed by the researching agency; (2) mass production and marketing of aids is not economically feasible because statistics indicating the number of persons who need comparable devices are unavailable; (3) the restraints of governmental regulatory agencies, limitations dictated by grant agreements of federal funding agencies, and the complexities of securing licenses, patents, etc., plus the possible liability and fear of lawsuits from the user population, have made the cooperative efforts of developers, research agencies and manufacturers most difficult to execute. While some of these items of technology are relatively simple and inexpensive, others are quite expensive and complex - all of them must be evaluated because of their profound effect on the lives of thousands of individuals and must reach the handicapped as soon as possible. A comprehensive approach to conducting the evaluation of research and development must include a critical needs assessment of user population; laboratory testing to determine such things as strength and durability to verify specifications and technical performance, suitability, acceptability and durability for specific patient applications - first with small numbers and then, if satisfactory, with larger scale external field evaluations; and finally, adequate reporting of findings at different levels of development. Evaluation should be cooperatively conducted with developers so that ultimate production specifications may be met. This evaluation process must be accomplished at the local REC level. The model developed by the University of Tennessee Rehabilitation Engineering Center and utilized in the evaluation of the Modular Plastic Insert (MPI) seating system is proposed as an example of one answer to the problem of providing adequate data for relevant decision making at all levels.

EVALUATION DEFINED

Evaluation is an aid to rational thought and action within the decision-making process. It includes specifying an information need and collecting, analyzing, and reporting information to alleviate that need. Those individuals in decision-making capacities in the development of rehabilitation technology may be the developer, the directors of the REC, the staff (physicians, engineers, technicians, and therapists) of the

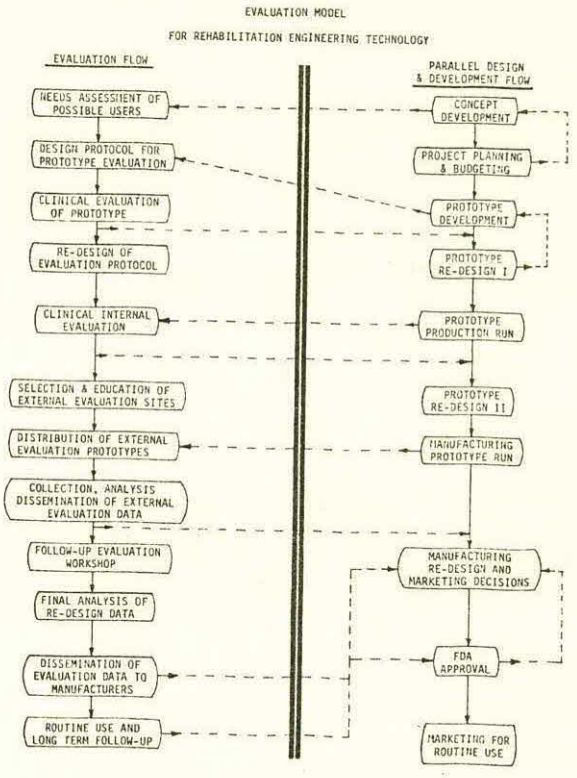
researching agency, those persons participating in the internal and external evaluation of the device/product, actual users, potential manu-facturers, funding agencies, and the professional and local community. Each of these groups makes decisions at different levels, and thus, requires information in different formats and at different times during the development of a device/product. Evaluation must be a continuous process of reappraisal of approaches and policies so that rational decisions can be made between often conflicting alternatives. Evaluation is not a product, but a means for enhancing the use of available resources and provides the information required for planning and developing technology. A comprehensive evaluation model must be one that is information-based; that is, with the reference points being the information users for the evaluation data and the information needs of each. Capitalizing on these two reference points, the evaluation model is designed.

DESCRIPTION OF EVALUATION MODEL

The evaluation design procedure relies heavily on the interaction between the evaluation team and potential information users. A design conference was held at the UT-REC to facilitate this interaction where relevant information users identified their information needs and relative evaluation questions were posed. When certain potential users could not be present, a role-playing technique was used quite effectively to project their needs. The design conference resulted in a final blueprint or protocol for conducting the evaluation.

UTILIZATION OF THE EVALUATION MODEL

A twelve (12) step evaluation process has been developed for use in assessing all University of Tennessee REC developments. It is illustrated by outlining the steps in the evaluation flow, the parallel design and development flow, and its execution in the development of the Modular Plastic Insert Seating System which is now considered essentially complete with the refined system ready for manufacturing negotiations. The evaluation model is presently utilized in the assessment of the Spherical Thoracic Support(STS), the Travel Base, the Foam-In-Place (FIP) Seating System, and the Plastic Upright Positioner (PUP) - all devices in various phases of development at the center. (See illustration).



This twelve step evaluation model is outlined as follows:

Step 1: Needs Assessment of Possible Users - The concept for development of a product/device is often based on needs indicative of a small population of persons who have been seen in the clinic. The developer bases his/her parameters for creation of the concept on this need. However, it is critical that needs and analysis data be available for project planning and decision-making. Surveys among professionals in the rehabilitation field; critical review of final reports, catalogs, brochures, descriptive material, and abstracts from other sources in the area of rehabilitation systems and devices; descriptions of other commercial solutions; and various census information are possible sources of relevant needs data. Paralleling the assessment of need for a modular plastic seating system, the concept development for the MPI system and the related project planning and budgeting for development of the system took place with project staff.

Step 2: Design Protocol for Prototype Evaluation- Following the needs assessment to determine feasibility for concept development and the development of initial prototype design, a prototype evaluation is developed involving the project staff and the evaluation team. This evaluation step parallels

the prototype development. The MPI evaluation protocol was determined and designed in this cooperative manner and included the four following components: (1) a clinical description of the client/user (Child's Profile); (2) a description of the user's performance utilizing the MPI (Therapy Evaluation); (3) a description of the technical aspects of the MPI (Technical Evaluation); and (4) an assessment of the MPI by those persons directly involved in the use of the system (Parent/Child Evaluation). The evaluation questions were formulated in such a way as to be answered by a simple checking response with space provided for comments. Following a prototype evaluation using this protocol, the MPI system was ready for the clinical evaluation phase.

Step 3: Clinical Evaluation of Prototype -
Following the development of usually a single prototype, the evaluation of the clinical value of the prototype is accomplished using the protocol previously described. This step in the evaluation assessment flow provides data for prototype redesign based on experience gained in the clinical testing with research subjects. This step in the MPI evaluation included clinical testing to determine such things as verification of specifications, technical performance, and determination of acceptability to user. Laboratory testing and planning will also begin at this time.

Step 4: Redesign of Evaluation Protocol -
The utilization of the established protocol in the clinical setting with the prototype is the basis for redesign of the evaluation methodology making additions or deletions to the evaluation instrument as needed. This protocol redesign takes place again with project staff who have had the responsibility for completing certain sections of the evaluation questionnaire and are cognizant of problems encountered in utilizing the instrument. The MPI protocol was redesigned and expanded considerably to include therapy evaluation at two distinct periods of use for comparison of effect: one at the time of initial fitting and another after four months of use.

Step 5: Clinical Internal Evaluation-
Once the experience of utilizing the prototype in the clinical setting is completed, a number of prototypes are produced for use within the clinical setting of the REC to gain further experience utilizing the second generation prototype. At this point additional data can be collected from the laboratory testing, plus the clinical performance related to suitability, acceptability and durability. This evaluation step follows the production run of prototypes and determines factors for the second redesign of the prototype based on a larger sample. The internal evaluation of the MPI seating system indicated definite advantages related to durability, aesthetics, therapeutic design, firmness of support, and range of sizes and components required. Problems included the fatigue fracturing of the plastic under heavy use, the need for more allowance for growth modification, and the lack of color choice.

Step 6: Selection and Education of External Evaluation Sites -
Following the internal experience, a decision must be made as to whether or not to conduct an external field evaluation. It was decided to take this step to determine if the MPI system had applicability to a significant number of children with special seating needs and what refinements to the system should be made to suit it for wider clinical application. The sample for evaluating the MPI system included forty-two individuals - six research subjects from seven clinical facilities in the United States and Canada. Teams from these clinical facilities came to Memphis for a 2½ day orientation session on the fabrication, patient selection, and fitting of the MPI system. The evaluation protocol called for fitting the six MPI systems at each center following the instruction manual provided to the external evaluators. Those prototypes were provided free of charge to the seven centers.

Step 7: Distribution of External Evaluation Prototypes -
External evaluators must be provided with sufficient prototypes and related parts for fitting research subjects. As external evaluators of the MPI system, the seven centers were provided with parts from which to construct prototypes for April, 1978 through January, 1979. This evaluation step must, of course, be coordinated with the manufacturing run of prototypes for use in the external evaluation. As many of the centers were connected through the Electronic Information Exchange System (EIES), it was possible to receive immediate feedback from participating evaluation centers concerning needs for parts, problems encountered etc., and also to transmit and receive ongoing status data regarding the evaluation process.

Step 8: Collection, Analysis, Dissemination of External Evaluation Data -
The compiling of statistics and writing of the external evaluation document provides critical redesign data. MPI data was collected utilizing the timetable for evaluation set forth in the orientation session. The date of completion for each of the four components of the evaluation protocol was as follows: Child's Profile - when child is initially brought to the clinic; Therapy Evaluation - at the time initial fitting and again after four months use; Technical Evaluation - after four months use; Parent/Child Evaluation - after four months use. A summarization of results from each of the components of the evaluation protocol indicates the status of a specific segment of the field-tested prototypes. The evaluation of the MPI was positive with definite attributes acknowledged by therapists, technicians, and parents. It was determined by project staff to make suggested refinements to the system to ready it for final redesign and consideration by possible manufacturers.

Step 9: Follow-up Evaluation Workshop -
A decision may be made to hold a final evaluation workshop for external evaluators to discuss design shortcomings and evaluation results and to set

priorities for further design refinements. In August, 1979, prior to the International Conference on Rehabilitation Engineering in Atlanta, a meeting was held with representatives from the seven external evaluating clinics to facilitate final redesign of the MPI system.

Step 10: Final Analysis of Redesign Data - Priority designations for design refinement may be made at the follow-up evaluation workshop and results collected, analyzed, and distributed to developers. At the August, 1979 workshop for MPI evaluators, redesign data was collected from participants concerning their first, second, and third priorities for redesign and priority design report resulted. These redesign refinements were incorporated into the MPI system and were completed in December, 1979. The comprehensive evaluation of the MPI system was then considered complete, and the refined system deemed ready for manufacturing negotiations.

Step 11: Dissemination of Evaluation Data to Manufacturers - Government supported research mandates that all information related to those developments resides in the public domain and should be available to all interested parties on an equal basis. Through conferences, presentations, publications, preparation of A/V materials, and visitors through REC centers, local research developments are exposed to a number of people - many of which have manufacturing and marketing interests. Interest has been shown by a number of companies in the MPI system. This presents the problem that a mechanism does not exist to expedite the formation of a working relationship between a developer and a single manufacturer without threat of demonstrating favoritism towards one manufacturer at the expense of others. At this time complete mechanical drawings, evaluation data, etc., have been made available to several potential manufacturers. A bid process will be initiated in order to market the product in keeping with the goals of the Rehabilitation Engineering Center.

Step 12: Routine Use and Long Term Follow-up - It is necessary that documentation results with any further local fittings be continued even after formal research on the development has been completed. Following the manufacture and marketing of the device/product, a longitudinal study of product suitability of the MPI will be conducted through a follow-up of sample users.

CONCLUSIONS

The utilization of the UT-REC evaluation model in assessing the development of the MPI Seating System has demonstrated both the reliability of the evaluation model in providing necessary data for decision-making at all levels and the subsequent decision that it is feasible to provide custom seating for the multiply handicapped using modular plastic components. It is felt the utilization of this evaluation model has resulted in a more effective and coordinated process for expediting the flow of new developments from research & development towards routine clinical use.

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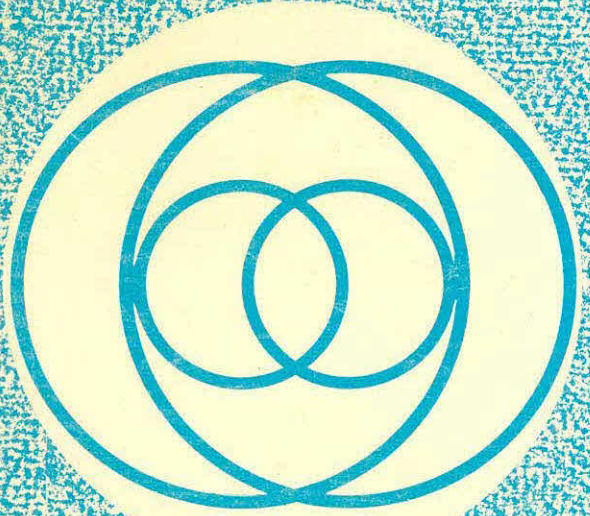

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