Design of a control system for a lower-limb exoskeleton rehabilitation robot with gait phase detection algorithm using inertial sensor

Ju-hwan Bae[1], Jong-won Lee[1], Suck-jin Hwang[1], kyu-suck Kim[1]

[1]*Korea Orthopedics & rehabilitation Engineering Center*

INTRODUCTION

The aging population is a global trend, and Korea also entered the aging society in 2018. The Elderly people have difficulty performing their ADL(Activity of Daily Living) due to muscle weakness in effects of aging. And the decrease in daily activity leads to a sharp decrease in physical function. A decrease in muscle mass and strength is a representative phenomenon of the decrease in physical function. Muscle mass in elderly people is important and reported to affect lifespan, and it is recommended to exercise continuously to maintain muscle mass.

Recently, research is being conducted to combine robot technology to develop a rehabilitation apparatus for rehabilitation of the elderly and the disabled. For the development of exoskeleton rehabilitation robot, actuator design, control strategy and evaluation method for performance evaluation should be considered. In the case of actuators, various mechanisms such as pneumatic muscle, electric actuator, SMA, Bowden cable have been studied, but Cyberdyn's electric actuator is commercially available[1].

Our group has developed a robotic wearable robot platform that can be actively assisted by an electric actuator. In addition, we developed a walk assistance method using a virtual negative damping model [2]. Negative Damping control has the advantage that it can be assisted by walking speed without distinguishing between swing phase and stance phase. However, it is predicted that effective assistance will be possible if additional assistance is provided to the user in the stand position.

In this study, we implemented the algorithm that can distinguish the walking steps by using the output of the acceleration sensor and the angle of the joint. In addition, we verified the algorithm for distinguishing pedestrian steps through experiments and confirmed that pedestrian assistance was applied according to the pedestrian step.

Low-limb exoskeleton rehabilitation robot

The lab developed a robotic wearable robot platform (see Figure 1). The hip joint module and knee joint module were designed to be electrically driven. The developed lower wearable robot high-correction module can be used alone or as an integrated module integrating the hip and knee modules. Each joint module is a Kolmorgen 200W BLDC motor and a Harmonic Drive with a 50: 1 reduction ratio.

The wearable robot system consists of a user's wireless interface, PC GUI, main controller and BLDC motor driver. In this study, a separate PC GUI was implemented to verify the control algorithm (see Figure 2). The developed PC GUI can set the driving status and walking parameters of the leg wearable robot. In addition, it is possible to monitor the walking state through the joint motion image.

The walking assistance control system of the leg wearable robot is composed of gait phase recognition algorism and muscle assist torque control algorithm. Figure 3 shows the control block diagram of the walking assistance system. Gait phase recognition algorism distinguishes the stance and stature from the angle, angular velocity, and acceleration inputs. The gait assist controller generates the motor torque according to the gait assist model according to the classified mouth / embosser information. In addition, the torque controller controls the torque of the joint stage through the control of the current. The torque at the joint end is designed to remove the friction elements of the gears and the weight elements of the mechanism through friction compensation and gravity compensation.

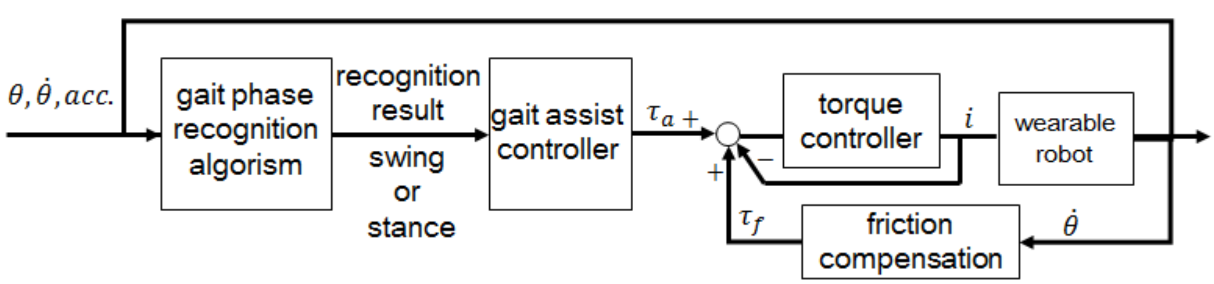

Figure 1. The control diagram of wearable assist robot

Designed of control system

The gait recognition algorithm is designed to distinguish the stature and stirrup using the 3-axis acceleration sensor value, the potentiometer for measuring the absolute angle of the hip joint, and the upper body angle estimate of the IMU sensor. Three-axis acceleration sensor values were used to detect ground contact points. In addition, as an algorithm for acceleration signal processing, a peak signal detection algorithm in which a threshold value for peak detection is actively changed according to a z-score is applied. The advantage of this algorithm is that it is advantageous for signals that are difficult to apply a given threshold because the acceleration value does not converge to a constant value during walking. Acceleration data were calculated by the vector sum acc_{xy} of Equation (1) for the x-axis and y-axis accelerations that correspond to the sagittal plane of the wearable robot. In this equation, acc_x is the frontal acceleration output on the sagittal plane, and acc_x is the longitudinal acceleration output. S_i in Equation (2) is data of acc_{xy} sampled and stored in real time, and is calculated as an average value s_i by Equation (3). In addition, z_i calculated in Equation (4) means the difference between the mean and the output value, that is, the deviation. This deviation is calculated as a variable threshold by multiplying by the constant th_r. It is output as a detection result through the condition of Equation (5).

$$acc_{xy} = \sqrt{acc_x^2 + acc_y^2} \qquad (1)$$

$$s_i = acc_{xy} \qquad (2)$$

$$\bar{s}_i = \frac{1}{l}\sum_i^{i+l} s_i \qquad (3)$$

$$z_i = x_i - \bar{s}_{i-1} \qquad (4)$$

$$\text{results} = \begin{cases} 1 & if\ z_i \geq th_r\sigma_{i-1} \\ 1 & if\ z_i \geq th_r\sigma_{i-1} \\ 0 & else \end{cases} \qquad (5)$$

Figure 5 shows the results when the accelerometer data measured after the robot is processed through the algorithm. The first graph shows the variable thresholds with and, and the second graph shows the peak detection results. In the results, 1 represents the result when the foot is in contact with the ground, and -1 is the result when the acceleration is zero while the foot is in motion. As a result of the experiment, it was confirmed that the recognition rate was about 95%.

In this study, the angle detection and acceleration conditions were used to identify the ground contact and to distinguish the gait through the peak detection algorithm. Together, these conditions were defined as the conditions for fuzzy inference as shown in Figure 6.

In this study, we designed a walking aid model for hip and integrated modules separately. In the case of the integrated module, the robot's foot closes to the ground during the stance, so the reactions to the ground, the robot and the human body should be considered. In addition, the swing assist torque and the gravity compensation torque must be simultaneously implemented in consideration of the weight of the mechanism. Fig. 6 (a) shows the model of the reaction reaction relationship between gravity compensation and ground. Equations (6) and (7) are equations for calculating the gravity compensation torques of the hip and knee modules, respectively. Equation (8) calculates the final output torque calculated from the relation function $\mathrm{G}()$ for gravity compensation and the function $\mathrm{A}()$ for walking assistance. Equation (9) represents Jacobian in relation to the torque of the hip and knee modules, and through this Jacobian, the torque required for the output of the tip of toe in Eq. (10) was calculated. The parameters for the equation are defined by the control model, where m_{sh} and m_{th} are the weights of the thigh and the thigh including the robot weight. F_x is set to the force to assist the weight of the robot and the user, F_x was set to zero.

$$T_{hip} = m_{sh}\left(l_{sh}\sin\left(\theta_{knee} + \theta_{hip}\right) + l_{th}\sin\left(\theta_{hip}\right)\right) + m_{th}l_{th}\sin\left(\theta_{hip}\right) \quad (6)$$

$$T_{knee} = m_{sh}l_{sh}\sin\left(\theta_{knee} + \theta_{hip}\right) \qquad (7)$$

$$\begin{bmatrix} \tau_{hip} \\ \tau_{knee} \end{bmatrix} = G(\theta_{hip},\ \theta_{knee}) + \mathrm{A}(\theta_{hip},\ \theta_{knee}) \qquad (8)$$

$$\mathrm{J} = \begin{bmatrix} l_{th}\cos(\theta_{hip}) + l_{sh}\cos(\theta_{hip}+\theta_{knee}) & l_{sh}\cos(\theta_{hip}+\theta_{knee}) \\ -l_{th}\sin(\theta_{hip}) - l_{sh}\sin(\theta_{hip}+\theta_{knee}) & -l_{sh}sin(\theta_{hip}+\theta_{knee}) \end{bmatrix} \qquad (9)$$

$$\begin{bmatrix} \tau_{hip} \\ \tau_{knee} \end{bmatrix} = \mathrm{J}^{T} \begin{bmatrix} F_x \\ F_y \end{bmatrix} \qquad (10)$$

In the case of the hip module, since the mechanism does not close to the ground, the reaction force against the ground is not taken into consideration. In addition, the weight of the cuff of the thigh is light, so the torque for gravity compensation can be ignored. Figure 6 (b) shows the walking aid model for hip module. The range of assisting force is designed to be adjustable within the maximum / minimum angle range.

Experiments and results

In the project, the researchers conducted their own experiments to evaluate the recognition performance of the walking phase during walking. Gait recognition and assistance were applied to the hip module. In the experiment, it was confirmed whether the recognition of the slow pace, normal pace and fast pace and the application of assistance. Fig. 8 shows the walking recognition status and auxiliary torque output result for each walking speed. In each graph, the first graph shows the hip angle, the second graph shows the angular velocity of the hip, the third graph shows the driving dock, the fourth graph shows the acceleration output and threshold, and the fifth graph shows the walking step recognition results. The recognition rate of the walking speed was 86% for slow walking, 74% for normal walking, and 81% for fast walking. As a result, it was confirmed that walking assistance is possible in most walking. In addition, an experiment was conducted to verify the walking aid function for the elderly. For this purpose, EMG was measured to verify the effect of walking aid when walking. However, because EMG signal is weak, it was difficult to see the secondary effect by EMG. Therefore, the effect of stair walking was verified to induce the occurrence of certain EMG. The measured EMG was processed as a smoothed estimate by applying a filtering technique using a moving average.

One-way analysis of variance (ANOVA) also examined the significance of differences in EMG before and after wearing. In the experiment, one elderly man (65 years old, male) performed the climbing motion five times and measured the EMG. Figure 9 shows the measurement of the right leg femoral EMG for a single step walk. The blue line represents the enveloped data of the measured EMG data. The experimental results showed no significant difference in EMG but the average of EMG peaks decreased after wearing.

CONCLUSIONS

In this study, we implemented the algorithm that can distinguish the walking phase by using the output of the acceleration sensor and the angle of the joint. In addition, the PC GUI is implemented to monitor the walking phase and walking assistance torque. In addition, we evaluated the walking step classification algorithm through experiments using hip module and confirmed the applicability of the algorithm. In addition, we conducted a clinical test on one elderly person on the integrated module, and confirmed that the defective wearable robot, which applied walking aid through electromyography, can perform a function to support walking in walking.

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