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Development of Control Model for Intelligently Controllable Ankle-Foot Orthosis*

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Abstract— **We have developed an intelligently controllable ankle-foot orthosis (i-AFO). In this paper, we formulated a new control method for the i-AFO. In the method the sensor system of the i-AFO estimates walking speed of user and decide optimal drop speed of foot at the duration between initial contact and foot flat. We conducted the pretest for eight healthy subjects to make a control rule for the drop speed. Then we conducted the modeling test for one patient to make an estimation rule for walking speed. Finally we conducted the evaluation test for the proposed method. Despite the walking speed estimation show errors, the i-AFO successfully controlled the foot motion depending on the gait states.**

I. Introduction

Locomotion is an important function in daily living, and its drawback is a severe barrier for comfortable life. Therefore, gait training is given a high priority in rehabilitative trainings. Normal gait is cyclic and can be characterized by the timing of the foot contact with the ground. An entire sequence of functions by one limb is known as the gait cycle [1].

Orthoses are supportive tools that are attached on the external surface of the human body to improve body functions, restrict, enforce, and support body segments [2]. Lower-limb orthoses are generally used to improve the gait functions of patients, and to assist ankle functions, ankle–foot orthoses (AFOs) are used to restrict involuntary plantar flexion. Main focus of this paper is a development of a controllable AFO.

Several kinds of powered knee orthosis can be found in commercial products, e.g. E-MAG, and C-Brace produced by ottobock. However, powered AFOs are still challenging research topics [3]. Researchers have proposed several kinds of powered AFO with different types of actuators, e.g. pneumatic actuation systems [4], ball screw drive systems [5], and series elastic actuators [6]. However, safety countermeasure is must for these active orthoses.

One of the potential solutions for safe controllable AFO is brake type controllable devices. Passive orthotic devices have been suggested as alternative design concepts for the controllable orthoses. In particular, drop foot can be controlled by using the passive devices alone. Passive controllable AFOs also have considerable advantages in terms of cost, safety, and miniaturization. Berkelman et al. [7] developed a completely passive orthosis with a parallel linkage mechanism alone; however this device requires extensive adjustment for the needs of the individual user. Farris et al. [8] suggested a joint-coupled orthosis that uses wafer-disk friction brakes to control the torque at hip joints and knee joints. However, the response time of their brake was about 150 ms, which is insufficient to assist the rapid dynamics of ankles; for example, the loading response normally ends after 10% of a gait cycle (about 100 ms).

To improve controllability of the passive controllable AFO, we have developed intelligently controllable ankle foot orthoses (i-AFO) by using compact magnetorheological fluid brakes (MR brake) [9]. Fig. 1 shows outer view (left) and inner view (right) of the i-AFO. We put the compact and rapid responsible brake (time constant: 10 ms) on its ankle joint to control ankle torque on the basis of the gait states. An accelerometer and potentiometer were used to identify the gait states. A spring unit was used to return the ankle angle to the initial position because the i-AFO does not have any actuator to do so.

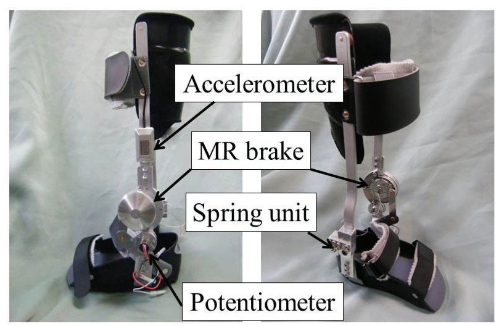

Figure 1. Intelligently controllable ankle foot orthosis (i-AFO)

The sensor system of the i-AFO has an algorithm to identify the current gait state of Fig.2 [9]. Each state was defined as follows:

State 1: from initial contact (IC) to foot flat (FF),

State 2: from FF to heel off (HO),

State 3: from HO, through toe off (TO), to the next IC.

The compact MR brake attached on the ankle joint of the i-AFO generates reaction torque. The brake is controlled with the different control rules for the states. In the state 1, the angular velocity of plantar flexion of the ankle joint is kept constant (reference speed: ω_{ref}) to support smooth loading response of the body segment. In the state 3, the ankle angle is kept constant to prevent abnormal drop foot during swing phase of gait. The brake does not generate any torque in the state 2.

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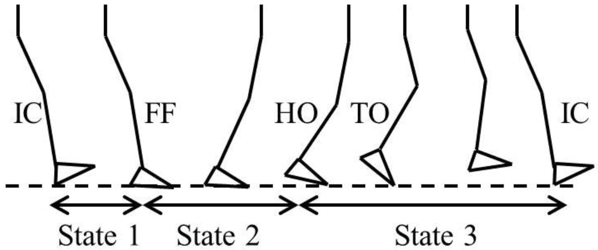

Figure 2. Definition of gait state for i-AFO

In the previous research [9], we experienced that comfortable value of ω_{ref} seems to link to the walking speed of users. However we have not developed a method to decide the optimal value of ω_{ref} for each walking speed yet. Such a rule must be useful to develop an automatic adjustable controller of the i-AFO for different walking speeds. In this paper, we originally developed a method to estimate the optimal value of ω_{ref} as a new control model of the i-AFO.

II. Pretest and Suggestion of New Method

A. Definition

In this paper, we use some gait parameters as key inputs for gait evaluations. The step length is a distance from the position of a foot to that of another foot. The stride is a length from the current position of a foot to the next position of it. The stride width is a transverse distance from the left foot to the right foot. These values are surely important to evaluate stability of the gait in practice [10]. Then these values have a potential to be used as indicators of the suitability of the control method of the i-AFO. In addition, the stride length is used to model user's gait as mentioned in the following section.

Furthermore, drop speed is defined as an average angular velocity of foot drop about ankle joint during the state 1 of Fig. 2. Velocity control of the drop speed is one of the main functions of the i-AFO [9] and its automatic adjustment is a main topic of this paper.

B. Setup

In order to check the relationship between walking speed and stride length or drop speed, we investigated gait experiments for 8 healthy male subjects (22-32 years old). The experimental setup is shown in Fig. 3. We developed a measurement system for the gait parameters mentioned in the section II-A by using Microsoft Kinect sensor [11]. The Kinect sensor contains a two dimensional distance sensor. We developed original software to measures distances from the sensor to the left foot and right foot independently, and calculated the step length dynamically. In addition, we also developed adjustable sensor attached AFOs for both feet so as to be wearable for many subjects to measure the ankle angle and gate states during waking (Fig.4). We put foot switches under the sole of the AFO to know the current gait state.

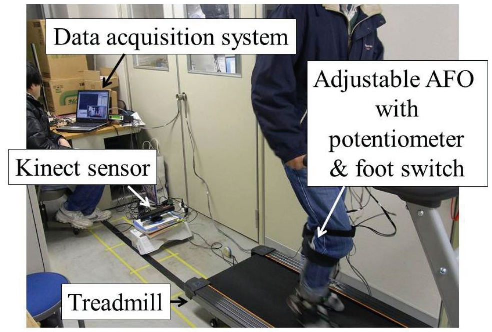

Figure 3. Setup for pretest

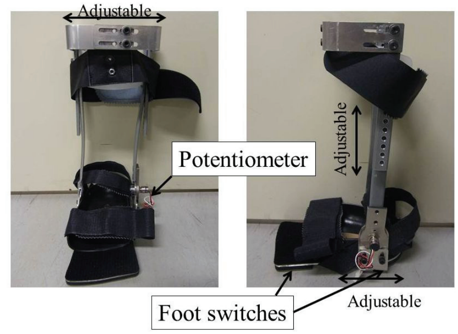

Figure 4. Setup for pretest

C. Results

Fig.5 shows the experimental results on the relationship between walking speed and step length of the gait. Each mark shows the result of each subject. In the same way, Fig.6 shows the experimental results on the relationship between walking speed and drop speed of foot in the state 1. According to the Spearman's correlation analysis, significant correlations were found in both between walking speed and step length, and between walking speed and drop speed. In several papers [12], the relationship between the walking speed and step length is reported to be a linear relation. The solid line in Fig.6 is the average of all data and adopted as a model for estimation.

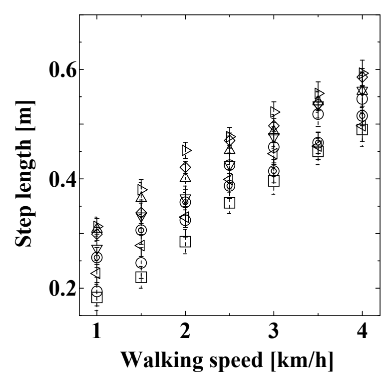

Figure 5. Walking speed vs. step length of healthy young adult (each mark shows each subject)

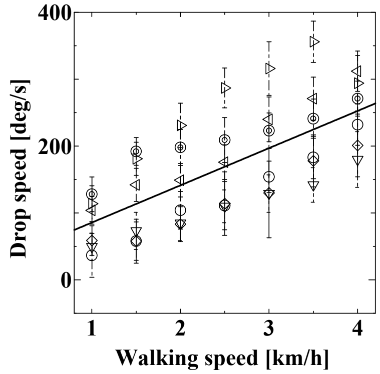

Figure 6. Walking speed vs. drop speed of healthy young adult (the solid line is the regression line of all data)

D. Suggestion of a method for automatic adjustable i-AFO

By utilizing facts we found in the results of the pretest, we suggest a new control method to realize the automatic adjustable control method of the i-AFO for different walking speeds.

Fig. 7 shows the summary of the control method. The i-AFO has already had a main function to identify the gait state (state 1 to 3 of Fig.2) from its sensor input [9]. The new controller counts the gait cycle (Δt [s]) as a duration from the former state 1 to the next state 1.

We model the relationship between walking speeds (V [m/s]) and stride lengths (L [m]) as a linear function as follows:

$$L = aV + b \tag{1}$$

According to the definition of Δt, it can be calculated as follows:

$$\Delta t = L / V \tag{2}$$

By combing Eq.(1) and (2), we can estimate V only from Δt, if we know the modeling parameters a and b.

$$V = \frac{b}{\Delta t - a} \tag{3}$$

As the next stage the controller decides the optimal value of ω_{ref} as a linear function of V as follows:

$$\omega_{ref} = cV + d \tag{4}$$

The ultimate purpose of the i-AFO is recovery of the normal gait for patients. From this view point, the regression line of Fig.6 is a potential candidate for the optimal model of ω_{ref}. In this case, c and d are 55.6 rad/m and 30.1 rad/s, respectively. In this paper, we use these values in the decision model.

Once the sensor system of the i-AFO decides the optimal drop speed, it controls the drop speed of user's foot with a feedback velocity controller. The details of the control methods are described in [11].

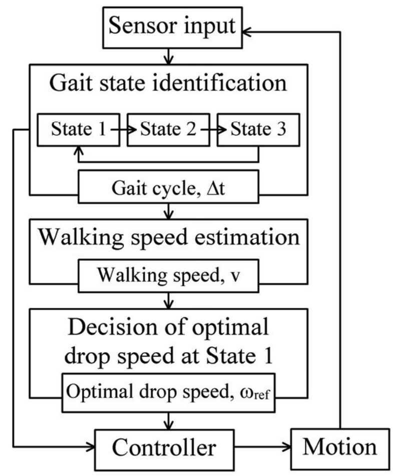

Figure 7. Concept of automatic adjustable control for i-AFO

III. MODELING TEST FOR REAL PATIENT

A. Subject

The subject was a male patient with post-Guillain–Barre syndrome (age: 39 years; height: 183.0 cm; mass: 83.0 kg). The subject has difficulty in voluntary movements of the peripheral part of the lower limb, especially the ankle joints and toes of both legs. He shows drop foot in walking. In addition, he shows considerable atrophy of the disused muscles. He usually attaches a plastic AFO to support the ankle function. The passive range of movement (ROM) of the ankle joint of the plantar flexion was 45°, and that of the dorsal flexion was 0°. He therefore had particular restrictions in dorsal flexion. We obtained the standard form of informed consent according to ethical guidelines.

B. Setup

The subject wearing plastic AFOs walked on the treadmill at some different speeds. The stride length was measured by the Kinect sensor mentioned in the section II-B.

C. Result

Experimental result is shown in Fig.8. The relationship between walking speeds and stride was almost linear as we expected. We can use the Eq.(3) as a model to estimate walking speeds. The modeling parameters, a and b were calculated from Fig.8 by root mean square method and they are 0.69 s^{-1} and 0.41 m, respectively.

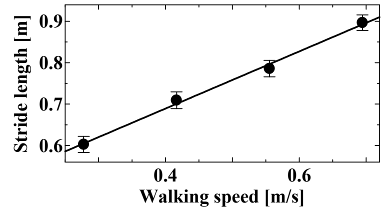

Figure 8. Result of modeling experiment

IV. Evaluation Test for Real Patient

A. Subject and setup

The same subject of the section III was recruited in this test. The subject walked on the treadmill with the i-AFO on his left foot. We implemented the new control rules in the i-AFO. The modeling parameter, *a* and *b* were set to be the values mentioned in the section III-C. Walking speed is controlled by the treadmill at 1.0, 1.3, 1.6, 1.9, 2.2, and 2.5 km/h. The subject wore the black suit with many optical markers and the infrared cameras acquired the position of each marker. The motion analysis was conducted with musculoskeletal model based motion analysis software (nMotion, nac Image Technology Inc.).

As control test, the subject walked with bare feet. For his safety, the subject walked on the flat floor without the treadmill. The walking speed was controlled with a metronome and his best effort. The same setup of the evaluation tests for the motion capture system was also used in the control tests.

B. Result

The experimental results were shown in Fig.9-12. The vertical lines show the starting point of each gait state. The dashed lines, single dot-dash lines, and double dots-dash lines show the start of the state 1, 2 and 3, respectively. Plantar flexion is defined to be positive in these figures. The horizontal line in Fig.12 shows the reference velocity of the feedback velocity control of foot at the state 1.

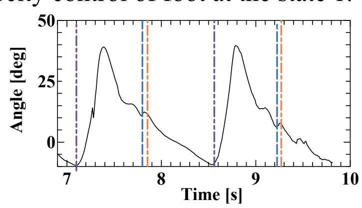

Figure 9. Ankle angle with bare foot at 1.6 km/h

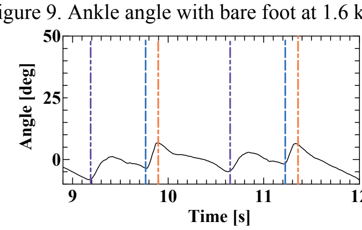

Figure 10. Ankle angle with i-AFO at 1.6 km/h

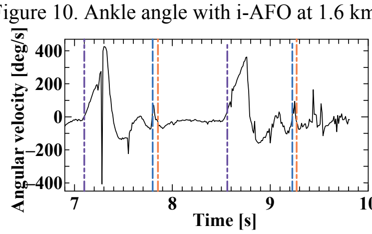

Figure 11. Angular velocity of foot with bare foot at 1.6 km/h

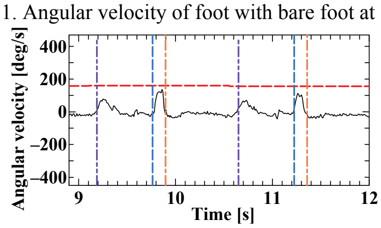

Figure 12. Angular velocity of foot with i-AFO at 1.6 km/h

C. Discussion

The gait states between the dashed lines and the dot-dash lines are the state 1. The duration of the state 1 is clearly increased by using the i-AFO. The i-AFO controlled the drop speed of foot at the state 1 and this increased the duration of the state 1 so as to produce the normal plantar flexion after IC. In addition, the i-AFO adequately prevented the abnormal drop foot at the state 3 (see Fig.9 and 10).

V. Conclusion

In this paper, we suggest a new control method for the i-AFO. In the method the sensor system of the i-AFO estimates walking speed of user and decide optimal drop speed of foot at the duration between initial contact and foot flat. We conducted the pretest for 8 healthy subjects to make a control rule for the drop speed. Then we conducted the modeling test to make an estimation rule for walking speed. Finally we conducted the evaluation tests.

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