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An Ankle-Foot Orthosis with a Variable-Resistance Ankle Joint Using a Magnetorheological-Fluid Rotary Damper *

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Abstract

This paper describes a prototype intelligent ankle-foot orthosis (iAFO) having an ankle joint with an adjustable viscous resistance torque. It was evaluated in a walking experiment involving a hemiplegic patient. The ankle joint was constructed with a magnetorheological (MR)-fluid damper having a changeable rotational viscosity. The iAFO consisted of an ankle joint, sensors for detecting walking, and a control box that comprised a circuit including Atmel AVR microprocessor and battery. Based on the difference in ankle joint function required in each stage of gait, walking was classified into three different phases in this study: the initial stance, mid-stance, and terminal stance/swing phases. A preliminary experiment was used to determine a method of detecting the shift from one phase to the next using the shank angle and foot contact information. The default resistance torque for each phase was configured. Using the prototype iAFO with the proposed control rules, a hemiplegic subject performed a walking experiment. While walking with the iAFO using suitable control rules, the ankle joint was maintained in dorsiflexion during the swing phase, and heel contact was achieved. In this respect, walking was better than unaided walking or walking using the iAFO with a resistance torque fixed at the highest level. The feasibility of the iAFO was confirmed.

Key words: Ankle-Foot Orthosis, Hemiplegia, Assistive Technology, Biomechanical Design

1. Introduction

People with hemiplegia often use ankle-foot orthoses (AFOs) to augment their ankle joint function when walking. AFOs offer some biomechanical benefits[(1)-(5)]. One of the main design goals of a conventional AFO is to hold the foot and ankle in a suitable position to correct involuntary plantar flexion. Therefore, conventional AFOs tend to be rigid and their mechanical properties do not change dynamically. A too rigid AFO may disturb the plantar flexion motion of the ankle joint during the initial stance phase of walking, during which smooth plantar flexion of the ankle joint is natural. Therefore, improving the hemiplegic gait should be possible by using an AFO that generates an appropriate resistance torque corresponding to the phase of the gait.

Yamamoto *et al.*[(6)] built an experimental AFO in which the rigidity and initial angle of the ankle joint could be changed arbitrarily to assess the effect of the mechanical properties

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of the AFO on the hemiplegic gait and measured the gaits of 15 hemiplegic patients using the AFO. They showed that using the appropriate assisted dorsiflexion moment for the individual hemiplegic patient's disability resulted in smooth motion at the time of the initial foot contact, which improved the gait throughout the gait cycle. Based on these results, a new AFO with an oil damper was developed that achieved sufficient plantar flexion of the ankle and mild flexion of the knee(7). Blaya and Herr(8) developed a variable-impedance active AFO with adaptive control and showed that the variable-impedance orthosis had some clinical benefits for treating a drop-foot gait compared to conventional AFOs with a zero or constant stiffness joint. Ferris *et al*.(9) developed a powered AFO using two artificial pneumatic muscles. Prototypes of active AFOs are generally heavy and the control unit and battery are not included in the current AFOs, so that the person wearing the AFO must carry these separately.

We developed an intelligent AFO (iAFO) with an ankle joint that generates an adjustable variable viscous resistance torque and evaluated it in a hemiplegic patient. The ankle joint was constructed using a magnetorheological (MR)-fluid damper for which the rotational viscosity could be changed. The iAFO includes the ankle joint, sensors for detecting walking, and a control box that consists of Atmel AVR microprocessor and battery. In addition, this study developed an algorithm to detect the gait phases and basic control rules for the iAFO and evaluated its initial feasibility.

2. Materials and methods

2.1 iAFO

The prototype iAFO shown in Fig. 1 consists of a pair of side frames and ankle joints, cuff, footplate, rotary cylinder, control box, electric angle meter, foot switches, and battery. The side frames and footplate are made of carbon-fiber-reinforced polymer (CFRP), and the cuff is a nylon belt. Commercial brace hinge joint parts (Bourbon, Tomei Brace) are used as the base of the ankle joint. Figure 1 (b) shows the variable-resistance ankle joint system. The rotary cylinder (CRB2BW20-90D, SMC) is connected to the lateral axial rod of the ankle joint and is filled with MR fluid (MRF-140CG, Lord). A polypropylene pipe connects the two cylinder chambers, which are separated by a vane, and enables fluid motion between them. The closer the magnet is to the pipe, the greater the viscosity of the MR-fluid is in the pipe, which increases the resistance torque of the cylinder.

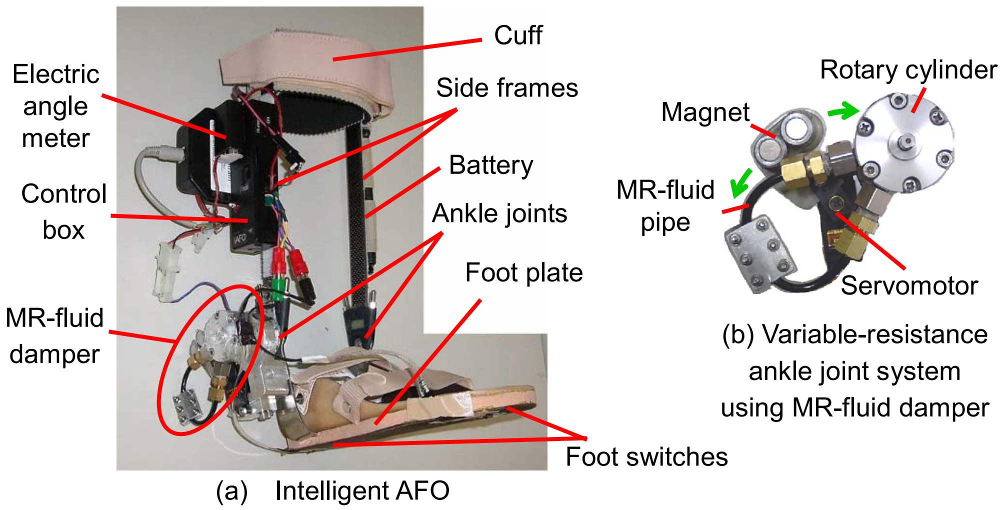

Fig. 1 The intelligent AFO (iAFO) is shown. The MR-fluid damper connected to the commercial ankle joint can vary the orthotic resistance torque.

In this study, four different viscous resistances were obtained in the ankle joint by adjusting the distance between a permanent magnet and the pipe carrying the MR fluid using a servomotor (ERG-VB, Sanwa). Modes 1 to 4 indicate the least to the greatest viscosity, respectively, based on the distance between the magnet and pipe for each control angle of the servomotor. Figure 2 plots the viscous resistance for mode 1 and Fig. 3 shows the peak resistance torque for modes 1 to 4 measured for various angular velocities using an AFO flexibility measurement device(10). These show that the resistance torque increases in proportion to the angular velocity and increases from mode 1 to mode 4. The iAFO has an electrical angle meter (FAS-G, Microstrain) attached to the lateral frame and foot switches (FlexiForce Sensor, Nitta) attached to the sole of the forefoot and heel. These sensors detect the angle of the shank to the vertical and ground contact on the affected side, respectively. Mode selection is controlled by a circuit that includes a microcontroller (AVR AT90S8535, Atmel). The AFO weighs approximately 1.3 kg.

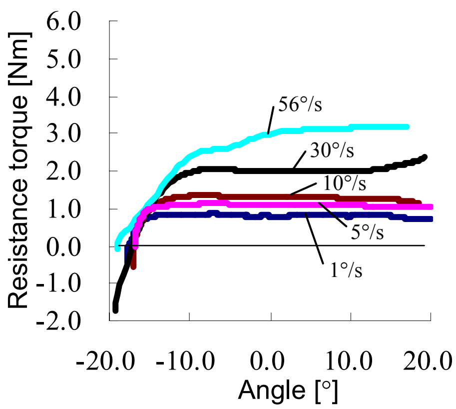

Fig. 2 The viscous resistance of the iAFO for mode 1 at angular velocities from 1° to 56°/s.

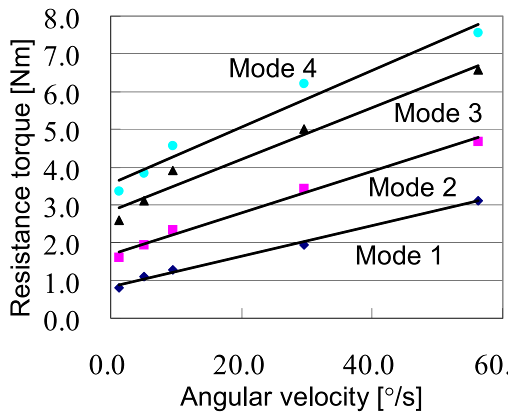

Fig. 3 The relationship between the peak value of the resistance torque and angular velocity for modes 1 to 4.

2.2 Classification of gait phase and their detection based on the sensor output

For changing the resistance modes of the iAFO, three walking phases were distinguished for a single gait cycle: 1) the initial stance (IS), 2) mid-stance (MS), and 3) terminal stance and swing (TS&S) phases. Within each phase, the desired ankle motion and position are similar, that is, generous plantar motion from the dorsal position, smooth dorsal motion from the plantar position, and the maintenance of dorsal position, respectively.

To detect the change between each phase using the electric angle meter and two foot-switches attached to the sole of the forefoot and heel, we measured the walk of a hemiplegic subject (Br.Stage IV) wearing a normal shoehorn-type AFO with the electric angle meter attached to the shank and foot switches on the sole of the forefoot and heel in a preliminary experiment. The forward/backward tilt angle of the shank (angle from the vertical) and foot contact information obtained from the sensors and plantar/dorsal flexion angle of the ankle joint on the affected side (relative angle between the shank and ankle) were recorded simultaneously using a 3D motion analysis system (MAC3D system, Motion Analysis). Figure 4 shows the changes in shank angle, ankle joint angle, and foot switch state while walking one step, beginning from toe off.

To control the iAFO, one must determine when one phase changes to the next while walking. Figure 4 shows that during the swing phase, the shank angle passes 0° once (circle a in Fig. 4). When walking at a constant velocity, the time between this event and heel contact is constant. Therefore, the time is set as a control parameter tuned to the individual.

The ankle joint plantar angle reaches a maximum when the forefoot switch is ON (circle b in Fig. 4). Therefore, the change in phase from IS to MS is defined as when the forefoot switch turns ON. The path shape of the ankle joint angle is similar to that of the shank angle when both foot switches are ON and the peak angles of ankle dorsiflexion and shank forward tilt occur at approximately the same time (circles c and d in Fig. 4). This was thought to have occurred because the forefoot rocker of the subject was weak, while the period of the ankle rocker was longer. Therefore, the time at which the ankle joint is in the maximum dorsal position is estimated from the pattern of change in the angle of the shank segment. Then, the time at which the angle of the shank segment reaches a prescribed value α is defined as the start of the terminal stance.

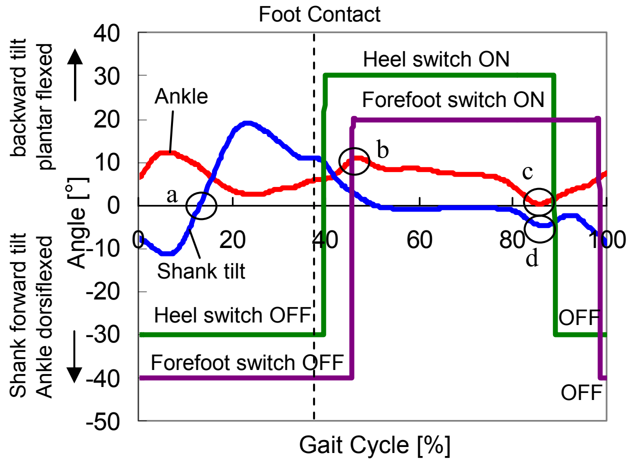

Fig. 4 The shank angle, ankle joint angle, and foot switch state while walking one step beginning from toe off.

2.3 Basic control rules

As described in the previous section, a single walking cycle was divided into three phases. The default control rules for each phase were established in advance from a healthy volunteer in a preliminary experiment. The control rules were designed so that clinical specialists, for example, physical therapists, can customize them for an individual hemiplegic patient by tuning the magnitude of the resistance torque via mode selection and the switching timing between phases after a trial of walking using the iAFO with default rules. The default control rules for phases IS, MS, and TS&S are as follows (Table 1 and Fig. 5).

1) IS: The iAFO is expected to generate dorsiflexion torque for assisting smooth plantar flexion. The default joint resistance is set to mode 2 for a moderate resistance torque. To switch the mode before heel contact, the resistance mode is switched to mode 2 after a constant time t from when the shank segment arrives at the vertical.
2) MS: During this period, the AFO must not disturb the stability of the knee. Therefore, the default mode for joint resistance is mode 1, which is the most flexible. This phase starts when both foot switches turn ON.
3) TS&S: Since foot clearance must be guaranteed throughout swing phase, the ankle joint should be held in a dorsiflexed position before swing phase. Therefore, the default is mode 4, which is the stiffest joint resistance. In the default, this period starts when the shank angle passes through a constant angle α.

Table 1 Control rule tuning parameters and default settings

Parameters	Default setting
Joint resistance in IS	Mode 2
Joint resistance in MS	Mode 1
Joint resistance in TS&S	Mode 4
Delay time t for changing to IS	0.08 s
Threshold angle α for changing to TS&S	10°

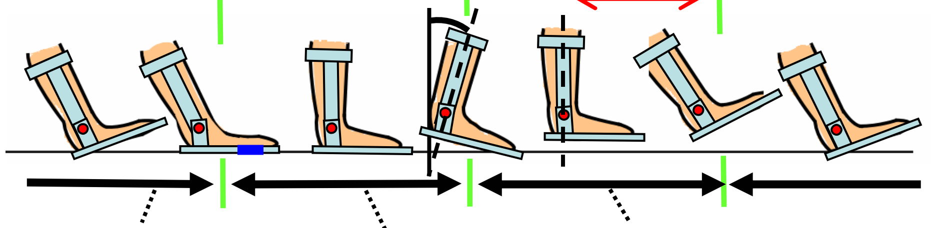

Fig. 5 Summary of the control rules, timings, operation, default settings, and tuning parameters.

2.4 Measurements

To evaluate the iAFO and proposed control rules, walking experiments were performed with a hemiplegic subject using the iAFO. The subject was a 76-year-old, 1.58-m-tall man who weighed 56 kg with right hemiplegia (Br.Stage IV and MAS 1st) due to a stroke approximately 3 months earlier. After a skilled physical therapist tuned the joint resistance mode shift threshold parameters for each phase (t = 0.03 s, α = 15°, others default), the gait was measured using a six-camera 3D motion analysis system (MAC3D system, Motion Analysis) and two force plates (Kistler) while the subject was wearing markers attached at the shoulders, hips, knees, ankle joints, fifth metatarsal heads, and sacrum. In all trials, the subject walked while using the cane that he normally used. The subject was instructed to walk at a comfortable velocity and pace. The walking motions were measured four times under each of the following conditions.

Condition 1: Wearing socks and shoes on both feet without the AFO. This condition was the same as in his daily life.
Condition 2: Affected side wearing the iAFO with the resistance torque fixed at mode 4. This was the highest resistance torque and simulated a shoehorn-type AFO.
Condition 3: Affected side wearing the iAFO using the resistance torque parameters as tuned by a skilled physical therapist.

The patients and the healthy volunteer were told that the goal of the experiment was to evaluate feasibility of the iAFO. Consent was obtained according to the Declaration of Helsinki of the World Medical Association. All experiments were done under the supervision of physical therapists of Hyogo Rehabilitation Center.

3. Results

Table 2 shows the mean values of the time cycle, speed, and stride lengths for the affected (R) and normal (L) sides and ratios of these strides under each walking condition.

Table 2 Temporal and spatial parameters of gait

	Cycle [s]	Speed [m/s]	R stride [m]	L stride [m]	R/L ratio
Condition 1	2.05	0.29	0.60	0.66	0.91
Condition 2	2.11	0.29	0.60	0.61	0.98
Condition 3	2.77	0.27	0.71	0.73	0.97

Figure 6 presents stick figures of one gait cycle: (a) condition 1, without the AFO; (b) condition 2, with the fixed mode iAFO; and (c) condition 3, with the iAFO operated using the control rules. The affected leg is drawn with a solid line and the rest of the body and other leg are drawn with broken lines. These figures show that the ankle joint angle is maintained in dorsiflexion during the swing phase (circled in the figures) using the AFO. The stride increased and the stride asymmetry improved. This confirmed that the AFO with our proposed control rules retained all the basic functions of a standard AFO, while improving the hemiplegic gait.

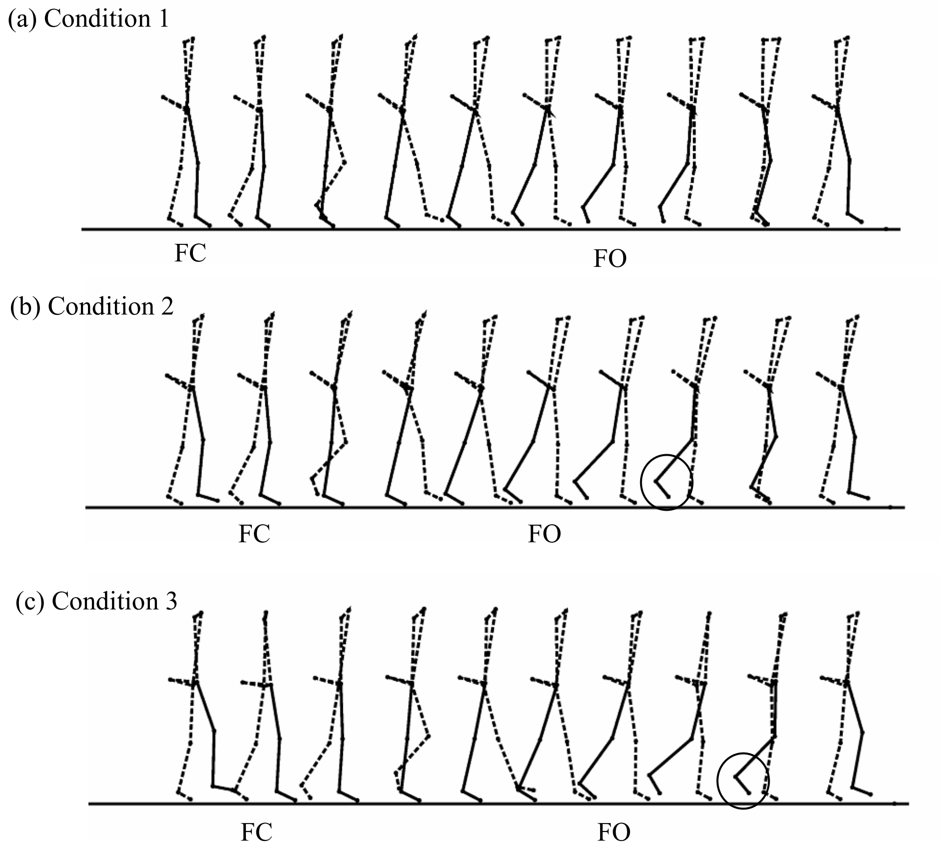

Fig. 6 Stick figures illustrating one gait cycle beginning with affected-side foot contact under each condition. Solid lines show the affected limb and broken lines indicate the rest of the body. FC and FO denote foot contact and foot off, respectively. For each cycle, ten illustrations are shown, with each offset for clarity.

Figure 7 shows the joint angle patterns of the affected side ankle (mean value ± SD) during one cycle of walking from heel contact to the next heel contact under each walking condition. At heel contact (0%), the ankle joint is plantar flexed under conditions 1 and 2, while it is dorsiflexed under condition 3. Under condition 1, the ankle joint is always in a plantar position and reaches the maximum plantar angle during the initial swing phase (ca. 75%). In contrast, under conditions 2 and 3, the ankle joint is maintained in a dorsal position after mid-stance.

Figure 8 shows the joint angle patterns of the affected side knee (mean value ± SD) during one cycle of walking from heel contact to the next heel contact under each walking condition. At heel contact (ca. 0%), the knee joint is extended under condition 1 only.

Figure 9 shows the joint angle pattern of the affected side hip drawn in the same manner as in Fig. 7. At heel contact (ca. 0%), the hip joint angle under conditions 1 and 2 is flexed 5° more than under condition 3.

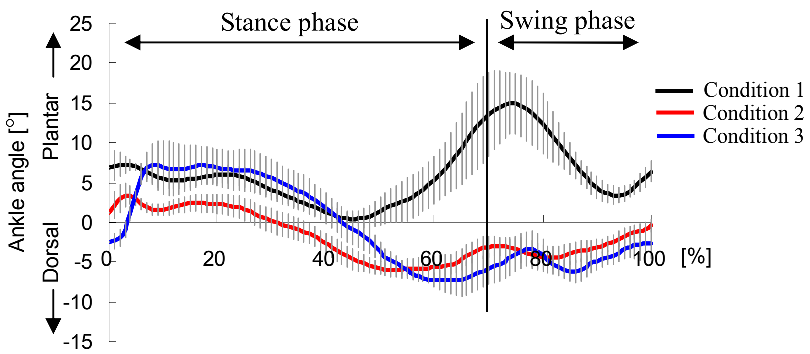

Fig. 7 Joint angle patterns of the affected side ankle (mean value ± SD) during one cycle of walking from foot contact to foot contact under each condition. The boundary between stance and swing phases is drawn using the mean value for all trials.

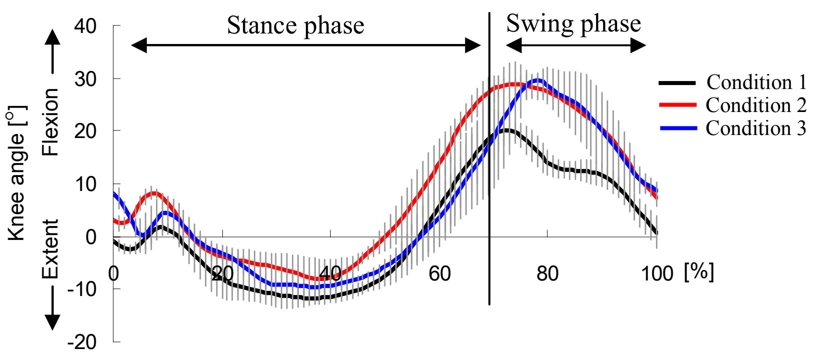

Fig. 8 Joint angle pattern of the affected side knee (mean value ± SD) drawn in the same manner as in Fig. 7.

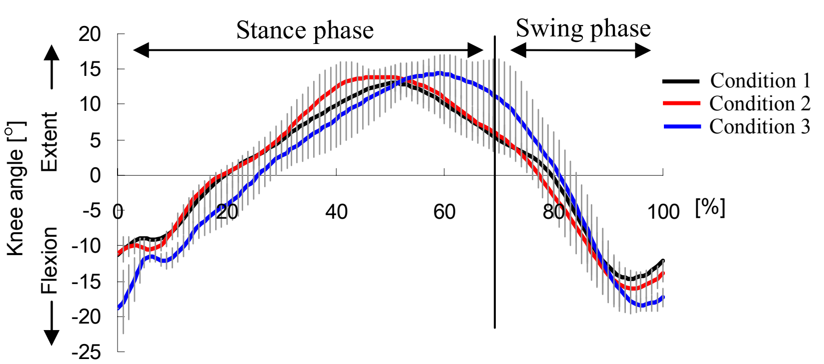

Fig. 9 Joint angle patterns of the affected side hip (mean value ± SD) drawn in the same manner as in Fig. 7.

4. Discussion

4.1 The difference in ankle joint pattern with and without the AFO

During the period from the terminal stance (ca. 45%) to the initial stance (ca. 10%), a notable difference was observed in the ankle joint pattern with and without the AFO. Under condition 1, walking without the AFO, the ankle joint was always in the plantar position and reached a peak of plantar angle during the initial swing (ca. 75%). Generally, the dorsal flexion muscle activity is weak in hemiplegia, and the subject thus had difficulty in moving the ankle in a dorsiflexed position. Furthermore, the peak angle of knee joint flexion during the swing phase was smallest under condition 1. This was likely because the subject had difficulty performing both plantar flexion of the ankle joint and flexion of the knee joint due to the synergistic extensor movement pattern(11). Under condition 1, unaided walking, the initial foot contact was with the toe, that is, heel contact was not achieved. Therefore, the loading response in the initial stance was not smooth.

In comparison, with the AFO (conditions 2 and 3), the ankle joint reached the dorsal position from mid-stance (ca. 40%). The AFO likely supported the ankle joint, preventing it from moving to the plantar position. The difference in the maximum knee flexion angles with and without the AFO was caused by the effect of both the dorsal ankle inhibiting contraction of the knee extensors due to extensor patterns and the weight of the AFO.

4.2 The effect of the iAFO control rules

The differences with the control rules were apparent, especially in the changes in the ankle joint movement pattern in the initial and mid-stance phases. Under condition 2, the resistance mode was fixed at mode 4 (the greatest viscosity), and the maximum ankle joint dorsiflexion angle was smaller and the time of peak dorsiflexion was earlier than under condition 3. The smaller maximum dorsiflexion angle of the ankle joint meant that the ankle joint at the time of heel contact (0%) was plantar flexed under condition 2, although the ankle joint resistance mode in the terminal stance and swing phase was mode 4 with both control rules. Under condition 3, the ankle was dorsiflexed at foot contact and initial contact with the heel was clearly achieved. The angles of knee and hip flexion at the time of heel contact were greater than under condition 2. Knee and hip flexion are believed to have been easier because the greater dorsiflexion angle of the ankle joint weakened the synergistic extensor movement pattern. As a result, the stride on the affected side was greater than under conditions 1 and 2. However, an excessive ankle dorsal-position during the swing phase may have obstructed knee joint extension during the terminal swing and at foot contact.

4.3 Advantages of the iAFO

AFOs are classified according to its supporting function around a disabled ankle joint, that is, it has structural elasticity, structural viscous-elasticity, elastic passive/viscous-elastic passive actuator or powered dynamic actuator, etc. It is thought that suitable function of an AFO for a patient who is with individual disabilities is different in each patient. It is desired for patients that AFOs with various and broad function are prepared as patients' option.

The iAFO is defined as one of AFOs with viscous-elastic passive actuator including adjusting mechanism of variable viscous resistance torque. As an AFO with similar functions, Blaya and Herr(8) developed a variable-impedance active AFO with adaptive control using a force-controllable series elastic actuator (SEA) capable of controlling orthotic joint elasticity and viscosity. Although it is one of the most advanced AFO because controllability for viscous-elastic property is extremely broad in principle, its feasibility is low because SEA is heavier, is needed more complicated control and consumes bigger electrical power. In contrast, with respect to the iAFO, electrical power is used for only working a servo-motor for the torque mode changing and a controller (circuit and AVR microprocessor), therefore, circuit and battery are able to become small and they are included in the iAFO. Moreover, generated torque was effective for improving one patient gait. For generating variable viscous resistance torque, using a passive actuator is one of the advantages of the iAFO for feasibility.

Tuning process of the control rules of the iAFO is that, basic control rules is prepared as a default setting, and a skilled therapist tunes the parameters depending on patient walking. Control rules proposed in this article is simple and was understandable for a clinical physical therapist. It is thought that tuning is not so more complicated than tuning of conventional AFOs in clinical situation even though some guidelines for tuning are needed. It is said that these tuning process, on which therapists could respond flexible for an individual, is a realistic option now, because patients' disabilities are various and a goal of suitable walking motion for each patient is not clear before rehabilitation therapy.

5. Conclusion

A prototype iAFO for the ankle joint was made with a variable-resistance torque mechanism. The variable resistance torque properties were measured as torque angle/angular velocity relationships and divided into four modes. A method of detecting the walking phase using the tilt angle of the shank and foot contact information was determined and basic control rules for the iAFO were proposed as default settings. These can be tuned for the user based on the clinical characteristics during hemiplegic walking. To evaluate the iAFO, a hemiplegic subject performed a walking experiment. The iAFO with the control rules tuned by a skilled therapist kept the ankle dorsiflexed during swing phase, from the terminal stance to foot contact, and achieved foot contact by the heel. This resulted in appropriate foot/ground clearance during the swing phase.

For practical use of the iAFO, its weight needs to be reduced and a protocol for setting the parameters for each subject must be devised.

Acknowledgment

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