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Development of an active ankle foot orthosis to prevent foot drop and toe drag in hemiplegic patients: A preliminary study

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Abstract. We developed an active ankle-foot orthosis (AAFO) that controls dorsiflexion/plantarflexion of the ankle joint to prevent foot drop and toe drag during hemiplegic walking. To prevent foot slap after initial contact, the ankle joint must remain active to minimize forefoot collision against the ground. During late stance, the ankle joint must also remain active to provide toe clearance and to aid with push-off. We implemented a series elastic actuator in our AAFO to induce ankle dorsiflexion/plantarflexion. The activator was controlled by signals from force sensing register (FSR) sensors that detected gait events. Three dimensional gait analyses were performed for three hemiplegic patients under three different gait conditions: gait without AFO (NAFO), gait with a conventional hinged AFO that did not control the ankle joint (HAFO), and gait with the newly-developed AFO (AAFO). Our results demonstrate that our newly-developed AAFO not only prevents foot drop by inducing plantarflexion during loading response, but also prevents toe drag by facilitating plantarflexion during pre-swing and dorsiflexion during swing phase, leading to improvement in most temporal-spatial parameters. However, only three hemiplegic patients were included in this gait analysis. Studies including more subjects will be required to evaluate the functionality of our newly developed AAFO.

Keywords: Active ankle-foot-orthosis, dorsiflexion, plantarflexion, foot drop, toe drag

1. Background

Foot drop and toe drag are symptoms of muscular weakness secondary to paralysis of the neural system [5, 7, 16]. Affected patients demonstrate abnormal gait patterns in which dorsiflexion and eversion of the ankle do not occur voluntarily. Due to a spastic plantarflexor, the sole or the forefoot, rather than the heel, strikes the ground at initial contact, resulting in a shortened stance time and triggering toe drag during the swing phase. Such inefficient gait patterns result in decreased walking speeds and increased energy consumption [3, 14]. There are two ways to improve gait patterns in patients with foot drop and toe drag: the use of an ankle-foot orthosis (AFO) and functional electrical stimulation (FES). However, conventional AFOs do not always yield satisfactory results. Carlson et al. compared gait patterns in cerebral palsy (CP) patients while not wearing and wearing a conventional AFO, and reported that the plantarflexion moment increased during ini-

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tial contact and during terminal stance [6]. However, walking speed and step length were not improved in these patients, and ankle power was reduced during late stance. Lehmann et al. reported that use of an AFO prevents toe drag during the swing phase in hemiplegic patients but does not prevent foot slap during the stance phase [13]. Additionally, conventional AFOs do not yield an adequate ankle plantarflexion moment. These shortcomings suggest the need for an AFO that addresses these disadvantages.

FES, which uses momentary electrical pulses to induce muscle contractions, exhibits promise as a potential permanent aid to remedy gait deficiencies. However, extended use of FES is still limited in practice [4, 10]. This modality requires the provision of a personalized custom device for each patient that is programmed based on the results of continuous trial and error data and is limited by the fact that it induces muscle fatigue. Automatic FES also encounters difficulties in gait phase detection and when being adapted to different walking speeds and patterns.

To prevent foot slap after heel strike, the ankle joint must minimize forefoot collision against the ground. In late stance, the ankle joint must remain active in order to function during push-off. The purpose of this study was to develop an active ankle-foot orthosis (AAFO) capable of maintaining dorsiflexion/plantarflexion in order to prevent foot slap and toe drag. We designed an AAFO that is capable of detecting and responding to such gait events, and evaluated our newly-developed AAFO during use by three hemiplegic patients.

2. Methods

2.1. Design of the AAFO

The AAFO is composed of a polypropylene AFO with a hinged ankle joint, a sensor unit, a controller, and a series elastic actuator (Fig. 1). The sensor unit detects gait events during walking and the controller controls ankle dorsiflexion/plantarflexion based on output from the sensors. The series elastic actuator moves the ankle joint according to signals from the controller. The AAFO had a total mass, including the series elastic actuator, of 2.8 kg.

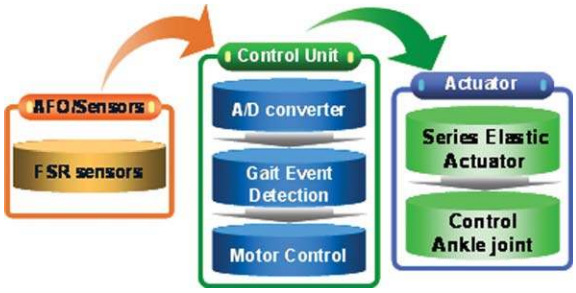

Fig. 1. Block diagram of the AAFO.

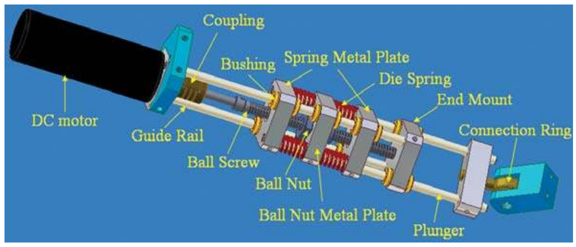

Fig. 2. Series elastic actuator (SEA).

2.1.1. Series elastic actuator (SEA)

A 24 V DC motor (RE30, Maxon Motor, Switzerland) was coupled with the SEA to control dorsiflexion/plantarflexion as the motor was battery-powered and easy to control. However, the motor was too large to attach to the ankle joint, and therefore the SEA was implemented. As is shown in Fig. 2, the SEA includes a coupling, two spring metal plates, a ball nut metal plate, an end mount, four compression springs, six bushings, one ball screw, one ball nut, two guide rails, two plungers, and a ring connecting the apparatus to the orthosis [2]. The ball screw and the ball nut convert motor rotations into translational motions [11]. The ankle joint of the AAFO was controlled by length changes of the SEA that were controlled by motor rotations. Four compression springs in the SEA were used to implement these length changes more smoothly.

2.1.2. AFO

An AFO was fabricated for each of three subjects by the Department of Prosthetics and Orthotics, Hanseo University, Korea. A hinged metal ankle joint was used in each AFO, and was designed to allow for dorsiflexion/plantarflexion of the ankle joint. However, this joint restricted all other directions of ankle movement. Figure 3 is a picture of the newly developed AAFO.

Fig. 3. The newly developed AAFO.

2.1.3. The range of motion (ROM) of the AAFO

Table 1 demonstrates the ROM of the AAFO. The maximum plantarflexion was approximately 21.5° when the device was set to the shortest length of the SEA. The maximum dorsiflexion was approximately 11.9° when the device was set to the longest length of the SEA.

2.1.4. Sensor

In order to detect gait events, force sensing register (FSR) sensors (MA-152, Motion Lab System Inc., USA) were used (Fig. 4). An FSR sensor is a small flat resistor in which resistance changes nonlinearly in relation to the applied force. The FSR sensors were used as on/off switches to indicate ground contact by measuring the voltage drop across the sensor when it was connected in a voltage divider circuit [12], since foot switches are simple and highly accurate sensors for detecting gait events. A total of four FSR sensors were used and were placed on the heel, the hallux, the first metatarsal head, and the fifth metatarsal base. The FSR sensors placed on the heel and the first metatarsal head were intended to detect gait events during normal gait. The first metatarsal sensor may not contact the ground during gait in hemiplegic patients, and, therefore, FSR

Table 1
The ROM of the AAFO

	Max. ROM	Max. plantarflexion	Max. dorsiflexion
Ankle joint angle (°)	33.4 ± 1.4	21.5 ± 1.4	11.9 ± 1.0
Motor rotation (number)	28	18	10

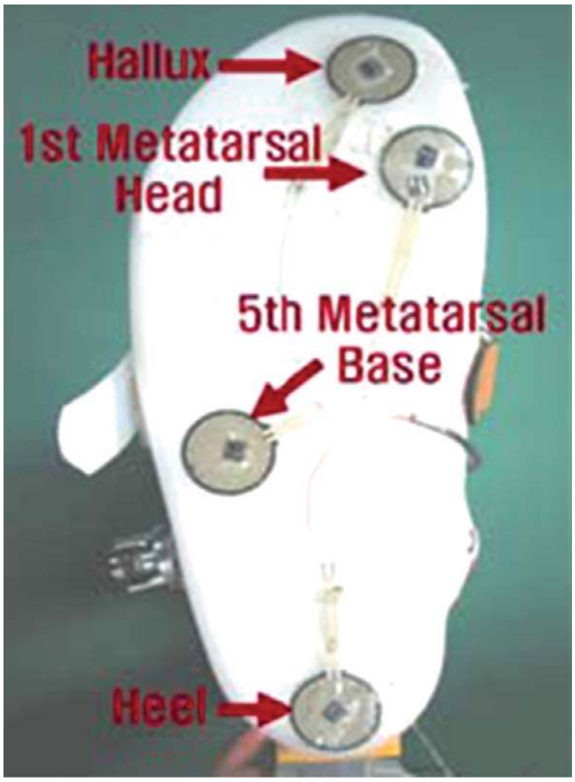

Fig. 4. The placement of the FSR sensors.

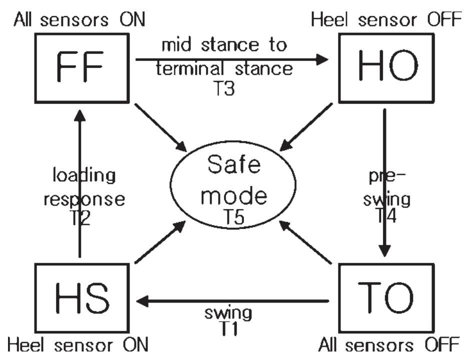

Fig. 5. Flow chart of gait event detection.

sensors were also placed on the fifth metatarsal base and the hallux.

2.1.5. Control unit

Figure 5 is a flow chart of the program used to control the AAFO. The control unit is composed of a microprocessor that detects gait events and controls the rotation of the motor, and a motor controller to control the motor based on the determined rotation of the motor. Sensor outputs are used as input signals for the microprocessor (PIC16C73, Microchip Technology Inc., USA) after the signals pass an amplification circuit, when the microprocessor performs A/D conversion of approved input signals. Gait events are then determined by a gait event detection algorithm, and the rotation of the motor is controlled by the actuator control algorithm. The motor rotation signal is composed of the rotation direction signal and pulse width modulation (PWM) signal. The PWM signal in turn controls motor rotation velocity. The motor controller used in the AAFO (LM18298, National Semiconductor, USA) transfers the motor rotation signal to the DC motor. The control unit and motor are powered by eight cell lithium polymer batteries. The AAFO microprocessor and battery pack were attached to the waists of the experimental subjects.

2.2. Control algorithm

2.2.1. Gait event detection algorithm

The gait cycle was detected using foot contact signals from the four FSR sensors. Four different gait events were defined, as shown in Fig. 5: heel strike (HS), foot flat (FF), heel off (HO), and toe off (TO) [1]. HS was recognized when the heel sensor was activated and all other sensors were inactivated. FF was recognized when the heel sensor and one of the other three sensors were activated after HS. HO was recognized when the heel sensor was inactivated after FF. TO was recognized when all sensors were inactivated (Table 2).

2.2.2. Actuator control algorithm

After HS, the controller shortens the actuator to induce plantarflexion, so that the foot is placed flat on the ground to support the body's weight. During FF,

Table 2
Gait event detection algorithm

	Gait event	Sensors			
		Hallux	Meta 1	Meta 5	Heel
T1	TO $\Rightarrow$ HS	OFF	OFF	OFF	ON
T2	HS $\Rightarrow$ FF	ON	ON	ON	ON
T3	FF $\Rightarrow$ HO	ON	ON	ON	OFF
T4	HO $\Rightarrow$ TO	OFF	OFF	OFF	OFF

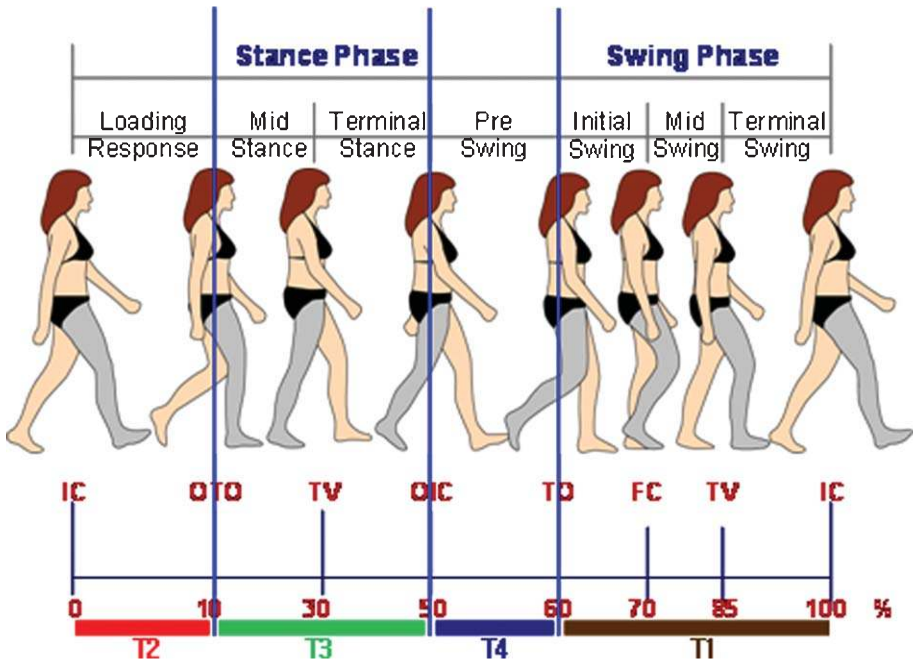

Fig. 6. Actuator control algorithm.

the actuator is lengthened to induce dorsiflexion. HO occurs after FF, during which the controller shortens the actuator rapidly to induce plantarflexion to aid in push-off. Then TO occurs, and the controller lengthens the actuator to induce dorsiflexion to prevent toe drag (Fig. 6).

2.3. 3D gait analysis

Three dimensional gait analyses were performed on three male hemiplegic patients (age; 51 ± 2.3 years, height; 163.5 ± 4.2 cm, weight; 63.5 ± 5.7 kg) using a 3D motion analysis system (Vicon 612, Vicon, UK). Sixteen reflective markers 14 mm in diameter were attached to anatomical locations following the Davis protocol [8]. Three different gait conditions were compared: gait without use of AFO (NAFO), gait while using a conventional hinged AFO that does not control the ankle joint (HAFO), and gait while using the newly developed AFO, which controls the ankle joint (AAFO, Fig. 7). The three subjects were provided with four weeks of AAFO gait training before data were collected for analysis. 3D analyses of each gait condition were performed once a week on different days for each subject. All subjects received at least 30 minutes of AAFO gait training prior to 3D analysis. Five repetitive measurements of each gait condition were made and then averaged. Statistical analysis was performed using a one-away ANOVA of temporal-spatial parameters.

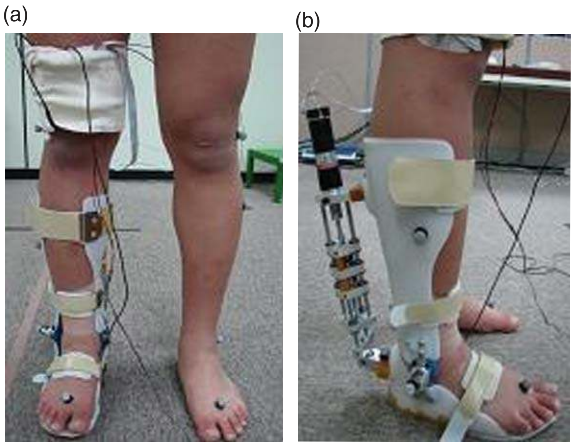

Fig. 7. Gait analysis: NAFO, HAFO, and AAFO.

3. Results

3.1. Temporal-spatial parameters

Figure 8 outlines the temporal-spatial parameters of the three different gait conditions (NAFO, HAFO, and AAFO). After four weeks, virtually all temporal-spatial parameters related to all three gait conditions increased. Step length and walking speed on the healthy side increased significantly for each patient when using the AAFO compared to the HAFO. In addition, the cadence and walking speed on the hemiplegic side increased ($p < 0.05$). After four weeks,

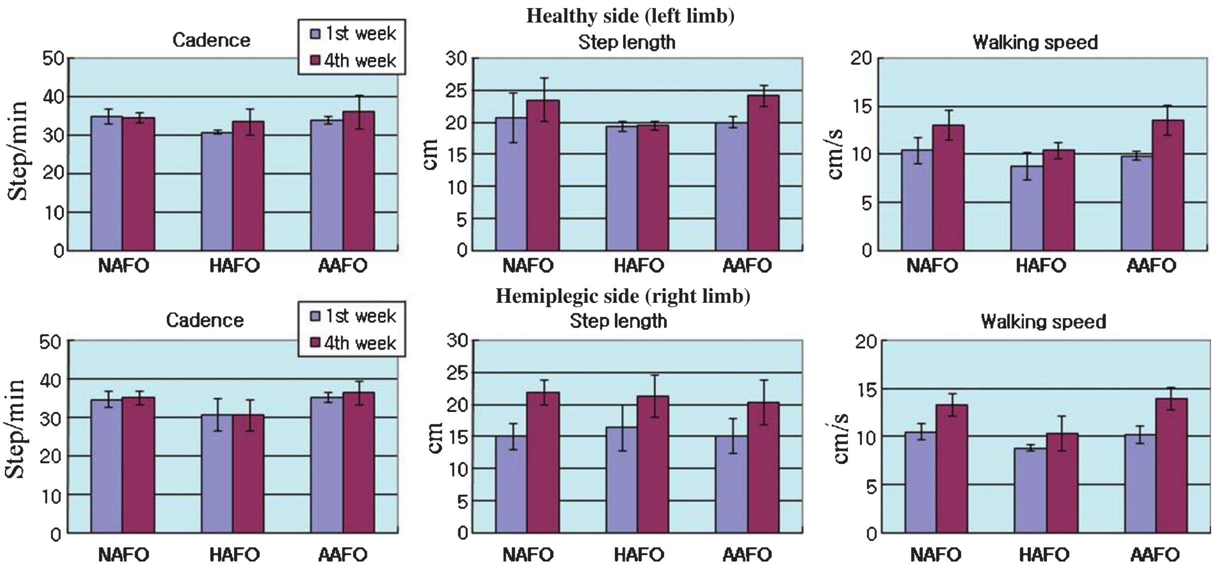

Fig. 8. Temporal-spatial parameters.

average walking speed during AAFO assisted gait had increased compared to NAFO, although this difference was not statistically significant.

3.2. Joint angles

Figure 9 shows the joint angles of the ankle during normal, NAFO, HAFO, and AAFO gait. During loading response, normal and AAFO gait both demonstrated plantarflexion, but plantarflexion was not observed during NAFO and HAFO gait. Until terminal stance, all gait conditions were characterized by dorsiflexion. During pre-swing, AAFO assisted gait demonstrated rapid plantarflexion while NAFO and HAFO assisted gaits demonstrated relatively slow plantarflexion. During swing phase, the normal and AAFO gaits demonstrated dorsiflexion while the HAFO assisted gait did not.

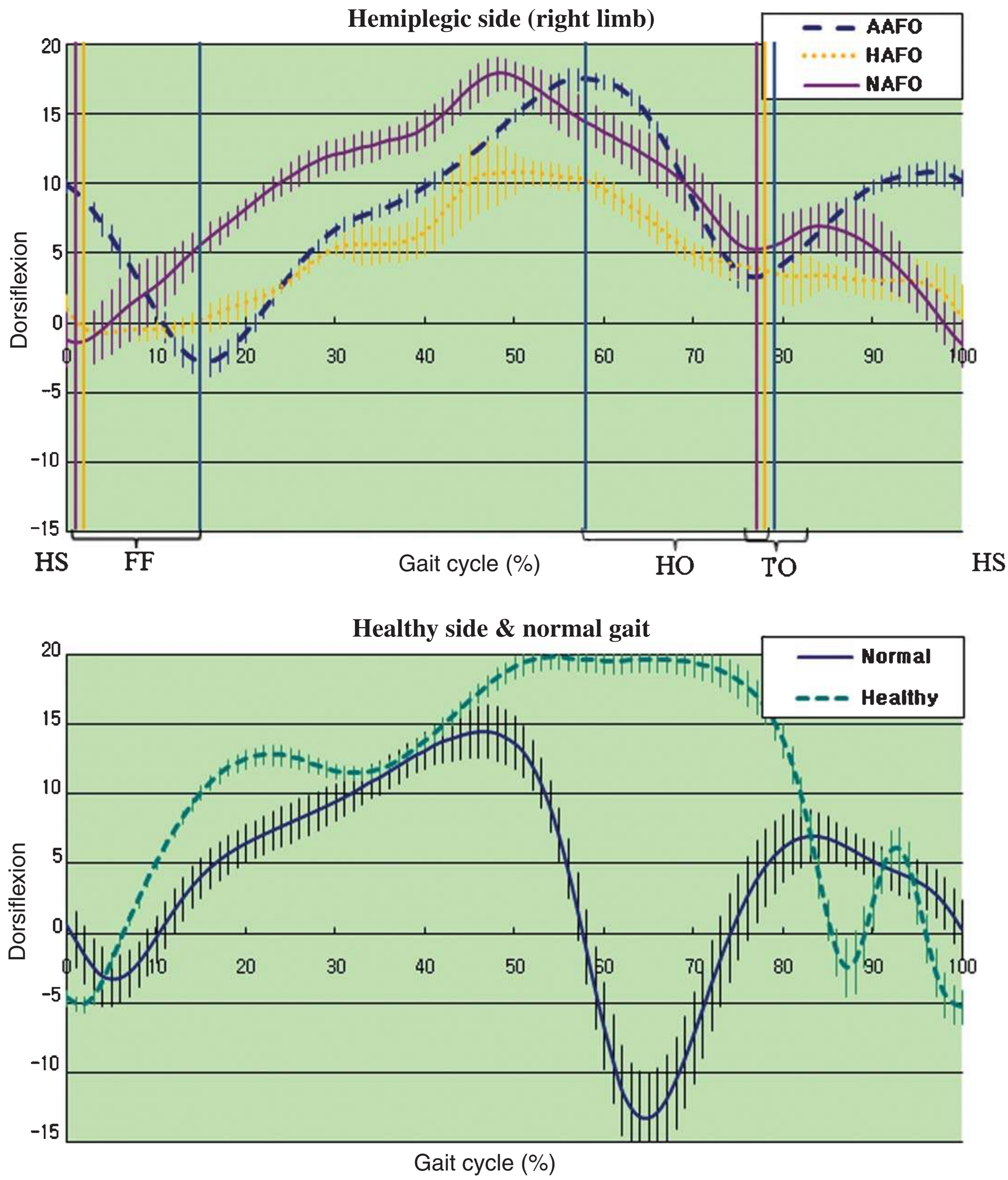

Fig. 9. Ankle joint angles during normal, NAFO, HAFO, and AAFO gaits.

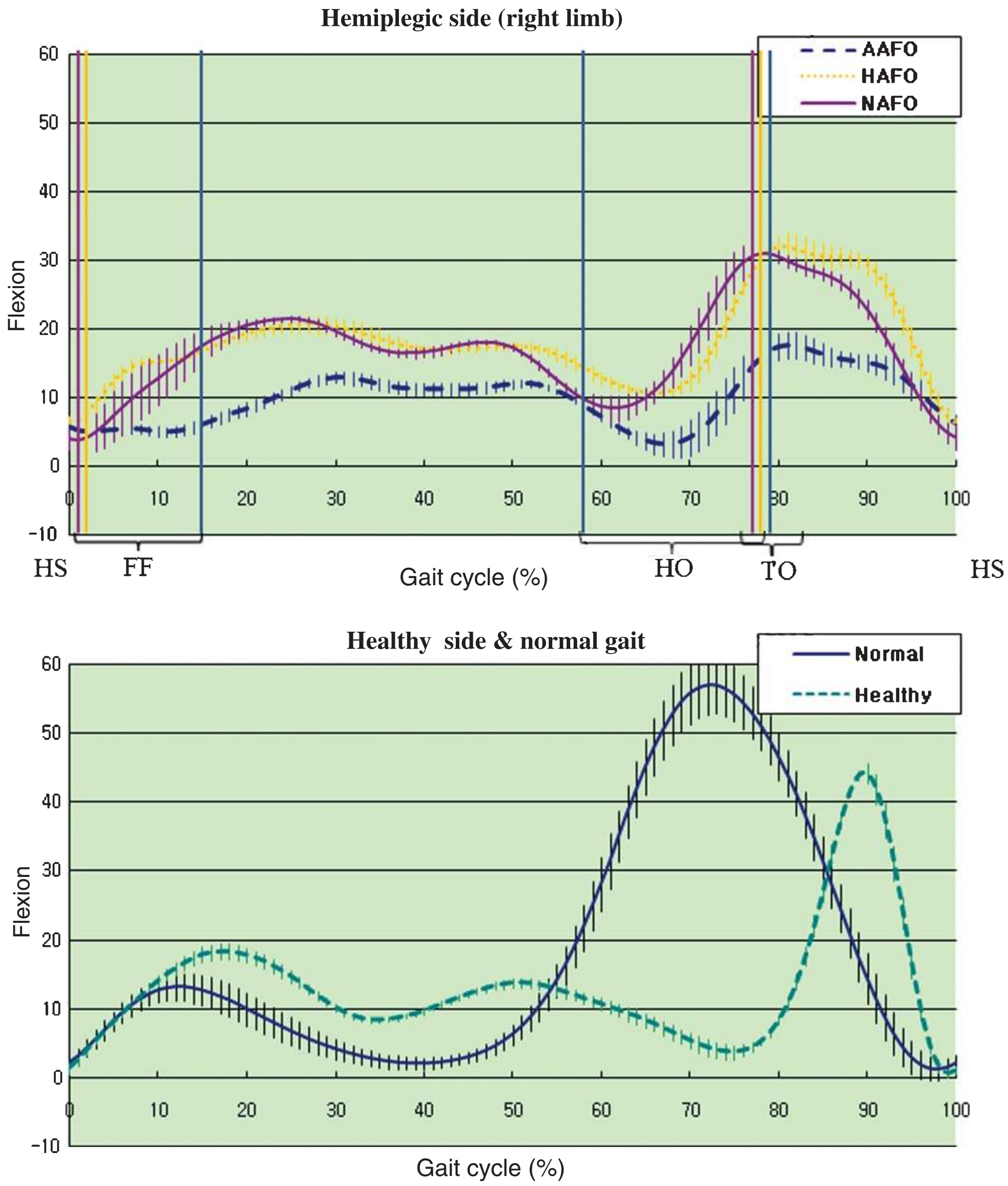

Fig. 10. Knee joint angles during normal, NAFO, HAFO, and AAFO gaits.

Figure 10 outlines the joint angles of the knee during normal, NAFO, HAFO, and AAFO gait. The hemiplegic (right) side demonstrated less knee angle movement than the normal side in each patient. HAFO and NAFO gaits demonstrated a more flexed knee joint during the standing phase than the AAFO gait.

4. Discussion

To evaluate our newly developed AAFO, we first evaluated whether the AAFO provided a ROM similar to that of normal walking. The entire ROM of the AAFO was approximately 33.4°, which is roughly equivalent to that of normal walking [15]. We then compared joint angles and temporal-spatial parameters during AAFO gait with the same measurements during NAFO and HAFO gaits. In NAFO gait, on the hemiplegic side, ankle joint angles indicated toe-drag with insufficient dorsiflexion during the swing phase. In addition, flexed knees were also noted during early stance, preventing contralateral limb advancement. In HAFO gait, the subjects demonstrated difficulties in ankle plantarflexion during the first double limb support period, and therefore the loading response was extended. These factors led to decreased walking speed when compared with the NAFO gait. During AAFO gait, on the hemiplegic side, an adequate amount of ankle plantarflexion occurred, preventing foot slap after initial contact, while rapid plantarflexion occurred during push-off. The ankle joint was rapidly dorsiflexed to prevent toe drag during the swing phase. During AAFO gait, walking speeds during the fourth week were faster than in for NAFO (4.5%) and HAFO (35.0%) gaits. However, the first plantarflexion following initial contact exhibited a large angle movement and extended time when compared with normal gait; this was a side-effect of dorsiflexion during the swing phase to prevent toe drag. In normal gait, ankle plantarflexion occurred before initial contact to pre-positioning. However, the controller was unable to identify an appropriate timing for initial contact using the FSR sensor prior to contact. Average walking speed during AAFO gait was faster than for the other gait conditions, although walking speed during AAFO was decreased by increased loading response time.

5. Conclusions

We designed and built an AAFO that controls ankle joint movements by detecting gait events in order to prevent foot drop and toe drag in hemiplegic patients. Our results demonstrate that our newly-developed AAFO not only prevents foot drop by maintaining plantarflexion during the loading response, but also prevents toe drag by causing rapid plantarflexion during pre-swing and dorsiflexion during swing phase. Our AAFO also enhanced most temporal-spatial parameters when compared with the HAFO gait. However, the current study was conducted using only three hemiplegic subjects. Further studies with larger samples will be needed to further evaluate the utility of this newly-developed AAFO.

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