Technical Note

Design and Evaluation of an Articulated Ankle Foot Orthosis with Plantarflexion Resistance on the Gait: a Case Series of 2 Patients with Hemiplegia

Daryabor A.[1,2], Arazpour M.[1]*, Aminian G.[1], Baniasad M.[3], Yamamoto S.[4]

[1]PhD, Department of Orthotics and Prosthetics, University of Social Welfare and Rehabilitation Sciences, Tehran, Iran
[2]PhD, Researcher in International University of Health and Welfare, Japan, Tokyo
[3]PhD, Mechanical Engineering Department, Sharif University of Technology, Tehran, Iran
[4]PhD, International University of Health and Welfare, Tokyo, Japan

*Corresponding author: M. Arazpour
Department of Orthotics and Prosthetics, PHD of orthotics and prosthetics in the University of Social Welfare and Rehabilitation Sciences, Kodakyarst, Daneshjo Blvd, Evin, Tehran 1985713834, Iran
E-mail: m.arazpour@gmail.com

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ABSTRACT

Ankle-foot orthoses (AFOs) have been described to have positive effects on the gait biomechanics in stroke patients. The plantarflexion resistance of an AFO is considered important for hemiplegic patients, but the evidence is still limited. The purpose of this case series was to design and evaluate the immediate effect of an articulated AFO on kinematics and kinetics of lower-limb joints in stroke patients. The articulated AFO with the adjustment of plantarflexion resistance was designed. The spring generates a plantarflexion resistance of the ankle joint at initial stance phase. The efficacy of orthosis was evaluated on two stroke patients in 2 conditions: without an AFO and with the AFO. Results showed the immediate improvements for walking speed, stride length and angular changes of dorsiflexion of the paretic ankle joint during a gait cycle of both subjects using the AFO compared with barefoot walking. The AFO also was able to reduce the paretic knee extension in the single-support phase of the stance and increase the vertical COM displacement during stance phase on the affected leg. In conclusion, the designed AFO affect not only the movement of the ankle joint but also the movements of the knee joint and the vertical COM height. These changes indicate improvement of the first and the second rockers and swing phase gait but not third rocker function. Further investigation is recently underway to compare its effect compared with other AFOs on the gait parameters of hemiplegic patients.

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Keywords

AFO; Orthotic devices; Stroke; Gait; Rehabilitation

Introduction

Individuals suffering from the stroke often develop abnormal joint kinematics and kinetics. Gait abnormalities in patients with a stroke history may result from impairment in muscle strength, motor coordination, constraint in joint range of motion, spasticity and/or deterioration in sensitivity [1, 2]. To take over these problems, wearing an ankle–foot orthosis (AFO) is clinically useful, and many researches have shown the positive influences of their use [3-5]. AFOs can be classified into two groups: Non-articulated AFO and articulated AFO. When compared with the two types of AFOs, studies have reported that non-articulated AFOs obstructed the natural movement of the ankle joint in stance and shortened the stride length, resulting in slower walking speed [6-8]. The articulated AFOs with mechanical stops are able to prevent drop-foot effectively by providing dorsiflexion assisting force or locking the ankle in an appropriate position, but these AFOs also constrain other normal movement of the ankle. For overcoming this problem, some researchers have introduced different motion control elements to provide more normal gait motion [9].

The plantarflexion resistive moment plays an essential role to reach heel contact in the first rocker of stance [10], and preserving the first rocker is an important function for an AFO [11]. In addition, the plantarflexion resistive moment created by an AFO may need to be adjusted without effect on the dorsiflexion resistive moment. There are some evidences considering the effect of an articulated AFO with the magnitude of the plantar flexion resistive moment on post stroke gait. Yamamoto et al. developed an articulated AFO with an oil-damper joint. This joint is a small shock absorber which uses hydraulic resistance [12]. Moreover, Kobayashi et al. investigated the influence of changing the plantarflexion resistive moment of another articulated AFO on ankle and knee joint angles and moments in patients with stroke [13]. Both of authors reported a substantial link between the kinematic/kinetic variables of the lower limb joints and the plantarflexion resistive moment of an articulated AFO [13-15].

Since eliminating every mechanical element could be an advantage for an orthosis, it appears simpler design of orthosis makes more durability, less repair and cheaper [9]. In this study, we designed an AFO with simple design in associated with the plantarflexion resistance without an oil damper. In addition, some articulated AFOs with plantarflexion resistance should be attached to the stirrup that it provides additional weight to an AFO [13, 16]. So, we evaluated effect of a newly designed ankle joint with light material attached footplate and the leg sections of AFO on stroke gait. The purpose of this study was, therefore, to design and evaluate a new articulated AFO incorporating a spring to determine its efficacy on spatiotemporal parameters, kinematics and kinetics of lower-limb joints in two stroke patients.

Material and Methods

Design considerations for an articulated AFO

Yamamoto et al. reported that the most important function of an AFO for hemiplegic patients is to provide the plantar-flexion resistive moment (synonymous with the dorsiflexion assisting moments) which are normally provided by eccentric contraction of the dorsiflexors at the initial contact of the stance phase [17]. Based on the findings of previous investigations, the characteristics for design were as follows (Figure 1):

- The AFO generates dorsiflexion assistive moment.
- The ankle joint of the AFO moves freely up to 30 degrees in dorsiflexion.
- The initial ankle joint angle of the AFO should be adjustable at 0 degree.
- Magnitude of the plantarflexion resistive moment for different body weights can be easily adjusted by a screw.
- The AFO should generate a resistive moment against plantarflexion.
- The plantarflexion range should be more than 10 degrees from the initial ankle joint angle.

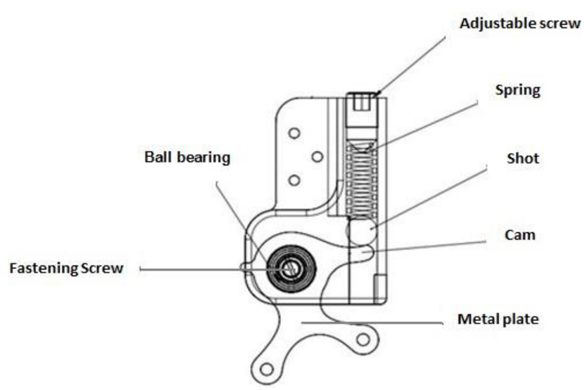

Figure 1: A schematic diagram of the ankle joint of new AFO.

For the initial calculation of the spring constant, we assumed that spring force must at least counterbalance the foot weight in swing phase. To calculate how much force should be applied to the joint of the spring, the foot mass and centre of mass was estimated using anthropometric data [18]. The lever arm (LR) of the spring to the anatomical axis of ankle joint was initially guessed based on the conceptual design, and as a result, F_{spring} was calculated as formula 1.

$$(1): \quad F_{Spring} = \frac{M_{foot}\, g\, LR_{foot}}{LR_{Spring}}$$

$$(2): \quad F_{Spring} = K\ \Delta x$$

Based on the force needed at each phase of gait cycle, the cam was designed and Δx was computed. Then, the spring constant was calculated using formula 2. To provide the spring with the specific coefficient, free and solid length, and long life, we decided to purchase the most similar commercial spring to avoid error and fatigue failure due to unstandardized hand-made fabrication process. Finally, the spring (B 10-044, Nouva Ret s.r.l., Italy) was selected based on the relevant calculation and detail design restrictions. In addition, we used aluminium T7075 which is light and strong, with strength comparable to many steels, and has good fatigue strength and average machinability [19].

The weight of the AFO joint is 100g, and therefore, the total weight of the AFO with this joint is almost 400g.

Participants

Two patients with hemiplegia participated in this phase of the study while they were walking either the AFO or the barefoot. Table 1 shows the characteristics of the subjects who volunteered for this study. Inclusion criteria for participants were at least six months post stroke, ability to walk independently without assistive devices, maximum grade 2 in ankle plantar flexion muscles spasticity according to Modified Ashworth Scale (MAS). Exclusion criteria included people with hammer toes deformity, hip and knee contracture, and patients with cardiorespiratory disorders and communication problems.

Study design and procedure

The study's design was a case series, and both patients gave informed consent to participate. The custom-made AFO was provided by an expert orthotist for every subject. To prepare the AFO, the patients' affected limb was cast while sitting on a chair. Then, the orthosis kept the ankle and foot at 90 degrees to the lower leg (neutral plantigrade position). The footplate and the leg sections of AFO were separately were made on positive moulded. These two segments were joined together by the designed ankle joint in lateral side of the ankle joint, and the medial side had a single axis hinge joint.

Gait was measured using a 3D motion analysis system (10 VICON cameras with 120 HZ frequency and two force plates with 1200 HZ frequency). Two force plates were arranged in two rows. The retro-reflective markers were attached to the landmarks of the patients the way full body modelling with Plug-in Gait. First, gait without an AFO (barefoot) was measured at the subject's selected speed, and

Table 1: Subjects demographic information

Subject	Gender	Affected side	Weight (kg)	Height (cm)	Age (year)	Time since onset (year)	Modified Ashworth Scale
A	Female	right	55	162	37	11	2
B	Female	right	62	164	50	10	2

measurements were repeated so that at least five steps of affected limb on the force plates were obtained. Then, the subjects started use of the new AFO. The gait with the AFO was measured after one hour practice to adapt to the AFO. The subjects stepped onto the right-hand plates with the right limb and onto the left-hand plates with the left limb. Before measuring gait with the AFO, the magnitude of the resistive moment of the AFO was tuned by the expert orthotics to each individual's condition according to the therapist's observation and the patient's opinion.

Data processing

The link segment model was defined, and the inverse dynamic model was employed to calculate the joint kinematics and kinetics [20] (Table 2). For each walking trial, the initial strikes of each leg were determined, based on the the foot markers and confirmed by the forceplate data using a Vicon Nexus 2.6 software. Outcome measures were: Spatiotemporal parameters, peak value of the joint angles of the ankle, knee, hip, and pelvis in a gait cycle, peak value of the internal ankle moment, negative and positive peak power around the ankle joint, and the height of the 2 peaks of the vertical displacement of the centre of mass (COM) in a gait cycle. A positive joint angle value shows dorsiflexion and flexion. The power was normalized by body mass. The vertical The COM displacement is relation to the energy cost of gait and was normalized to body height.

Outcome measures during gait with and without the AFO for each patient were compared.

Results

The gaits without an AFO (barefoot), and when wearing the new AFO were measured. The initial ankle joint angle of the AFO was adjusted at 0 of dorsiflexion. Table 2 and Figures 2 and 3 show data for the gait parameters measured for both subjects A and B.

Case Description

Case A

When subject A wore the AFO, there was an increase in walking speed and paretic stride length by 9% and 7%, respectively. When walking with the AFO, the ankle dorsiflexion in the phases of the initial contact, loading response, single support, pre-swing and swing increased 15.28%, 9.05%, 4.1%, 15.8%, and 11.64% respectively, when compared with the condition without an AFO.

The patient had some knee hyperextensions in initial contact and the single support phase of stance during barefoot walking. Wearing the AFO reduced amount of the knee hyperextension in two phases compared with barefoot walking (initial contact: 8.3 degrees% and single support: 2.93 degrees%, respectively). The pelvic obliquity in swing did not change with the AFO (difference: 1 degree). When the patient walked with the AFO, the first peak of COM trajectory also was increased by 8.52 mm% compared with when the patient did not use the AFO. As for the parameter related to third rocker function, however, peak ankle power generation in terminal stance was decreased 0.88% in the condition of walking with AFO compared with barefoot walking.

The patient's impression was "I would rather this orthosis compared with barefoot walking because it enables me to move my ankle joint smoothly with it".

Case B

When she wore the AFO, there was an increase in walking speed, cadence, stride length, single support time, and a decrease in double Support time (Table 2). Kinematic data for subject B showed a plantarflexion reduction in a gait cycle, an increase in knee and hip maximum flexion at initial contact, and a decrease in knee maximum flexion during late stance while wearing the new AFO. In addition, an increase in the first peak of COM trajectory was demonstrated under the AFO condition than without the AFO (10.2 mm%). Moreover, a difference of 1.62 degrees% in

Table 2: Results of spatiotemporal, kinematic and kinetic data

Parameter description	Subject A		Subject B	
Spatiotemporal	**WO**	**With AFO**	**WO**	**With AFO**
Walking speed (m/s)	0.67 (0.12)	0.76 (0.076)	0.47 (0.02)	0.51(0.02)
Cadence (step/min)	88.57 (7.87)	89.48 (5.73)	63.08 (4.23)	68.36 (6.07)
Stride time (s)	1.36 (0.12)	1.48 (0.28)	1.91 (0.13)	1.76 (0.16)
Step time(s)	0.74 (0.065)	0.71 (0.041)	1.11 (0.05)	0.99 (0.11)
Foot. Off (% gait cycle)	61.28 (3.906)	43.93 (2.97)	64.47 (0.73)	62.69 (1.98)
Single. Support time (s)	0.50 (0.077)	0.47 (0.036)	0.44 (0.04)	0.48 (0.02)
Double. Support time (s)	0.33(0.050)	0.32 (0.023)	0.78 (0.07)	0.62 (0.11)
Paretic Stride length (m)	0.94(0.070)	1.01 (0.040)	0.85 (0.01)	0.91 (0.04)
Non-Paretic Stride length (m)	1.01 (0.081)	1.07 (0.036)	0.89 (0.015)	0.95 (0.01)
Paretic step length (m)	0.52(0.034)	0.52 (0.068)	0.47 (0.01)	0.48 (0.04)
Non-Paretic step length (m)	0.54 (0.051)	0.53 (0.025)	0.43 (0.01)	0.43 (0.005)
COM				
P1 of COM (mm)	830.10 (4.17)	838.62 (1.30)	811.08 (5.55)	821.28 (5.88)
np1 of COM	0.512407 (0.00257)	0.51766 (0.02469)	0.49456 (0.00338)	0.50078 (0.00358)
P2 of COM (mm)	844.70 (3.99)	842.83 (3.23)	826.20 (6.39)	812.15 (5.81)
np2 of COM	844.70 (3.99)	0.51392 (0.00196)	0.50378 (0.00389)	0.49521 (0.00354)
Ankle				
Angle at initial contact (^{0})	-11.41 (2.79)	3.87 (1.43)	-21.44 (6.57)	0.36 (8.67)
Peak PF angle in loading response (^{0})	-17.58 (3.45)	-8.53 (0.68)	-24.79 (6.44)	-14.84 (2.82)
Peak DF in stance(^{0})	8.58 (1.67)	12.68 (0.89)	6.62 (1.46)	19.15 (2.37)
Peak PF in pre-swing(^{0})	-12.28 (5.05)	3.52 (2.14)	-7.52 (3.05)	10.20 (1.60)
Peak DF in swing(^{0})	-2.16 (2.53)	9.48 (0.72)	3.95 (2.91)	14.90 (0.83)
Peak DF moment in loading response (N.mm/kg)	87.47	-90.92	-58.94	-104.15
Peak PF moment in terminal stance (N.mm/kg)	1499.12	1271.63	1005.48	1036.29
Max ankle power absorption (W/kg)	-1.06 (0.25)	-1.15 (0.08)	-0.52 (0.02)	-0.84 (0.09)
Max ankle power generation (W/kg)	1.43 (0.37)	0.55 (0.09)	0.7 (0.01)	0.26 (0.07)
Knee				
Angle at initial contact(^{0})	-4.92 (2.64)	3.38 (3.01)	9.34 (2.66)	15.21 (6.79)
Peak extension in stance(^{0})	-11.05 (0.46)	-8.12(0.50)	-1.85 (0.47)	-0.52 (0.41)
Peak flexion in swing(^{0})	41.47(2.05)	45.89 (2.25)	62.25 (3.04)	69.55 (3.28)
Hip				
Flexion at initial contact(^{0})	30.07 (1.47)	33.72 (2.21)	36.82 (4.34)	47.26 (5.61)
Peak extension in stance(^{0})	-5.60 (1.98)	-5.22 (1.09)	1.66 (0.56)	8.07 (0.43)
Swing flexion(^{0})	37.67 (1.39)	38.83 (2.39)	43.17 (2.27)	51.40 (0.99)
Pelvis			1.66	
Obliquity in swing (^{0})	-7.17 (2.03)	-8.56(1.18)	-4.55 (0.82)	-2.93 (0.98)

WO: without the AFO, PF: plantar flexion, DF: dorsiflexion, negative number indicates plantar flexion and extension angles; positive number indicates dorsiflexion and flexion angle, COM = centre of mass; P1 of COM and P2 of COM = height of 1st peak (affected leg in stance) and 2nd peak (nonaffected leg in stance) of the COM vertical trajectory in a gait cycle; np1 of COM and np2 of COM = normalized 1st peak (affected leg in stance) and 2nd peak (nonaffected leg in stance) of the COM vertical trajectory in a gait cycle.

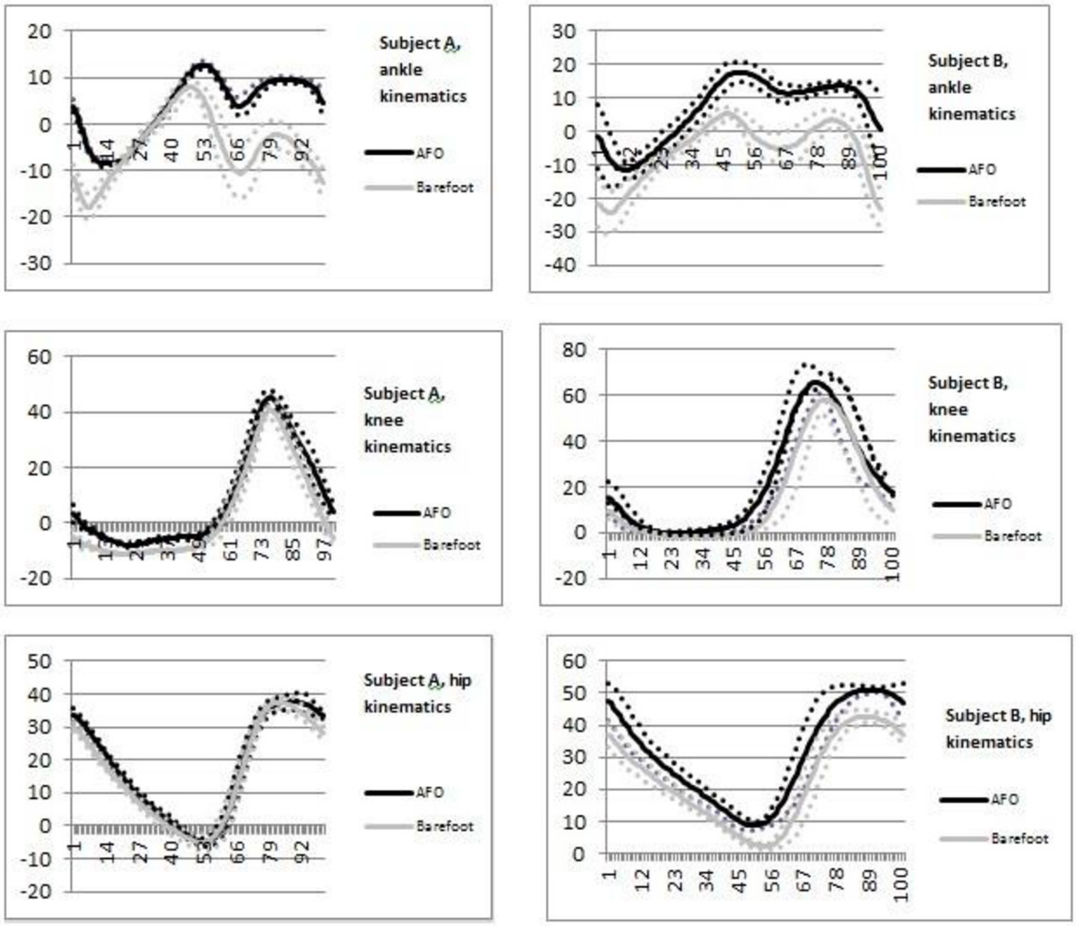

Figure 2: The sagittal kinematic data at the ankle, knee, and hip joints of subjects A and B. Positive Values flexion/dorsiflexion, negative values: extension/plantar flexion. The barefoot curves are depicted with solid grey line, The AFO curves are depicted with solid black line, and standard deviations for every condition are depicted with dotted line.

the swing pelvic obliquity was found with the AFO walking versus the barefoot walking. As for the parameters related to third rocker function, peak ankle power in terminal stance decreased in the condition with AFO compared with barefoot like the patient A. Patient B's impression about the AFO was the same as that of patient A.

Discussion

The present results showed that the AFO properties affect not only movement of the ankle joint but also the movements of the knee and hip joints. In both of subjects, foot-slap, knee flexion and hip flexion improved during initial stance phase when they wore the AFO. In addition, the knee hyperextension during the stance phase reduced when walking with AFO than walking without AFO. It can be said the AFO with plantarflexion assists to make up for insufficient activity of the dorsiflexors, making first rocker function possible during loading response of the paretic limb. In addition, AFOs can affect indirectly knee joint angle. The plantarflexion resistive moment of the AFO allowed the ankle joint to be retained in a more dorsiflexed position. This position created probably the knee to kept more flexed position and anterior to the ground reaction force, resulting in a decreased peak knee ex-

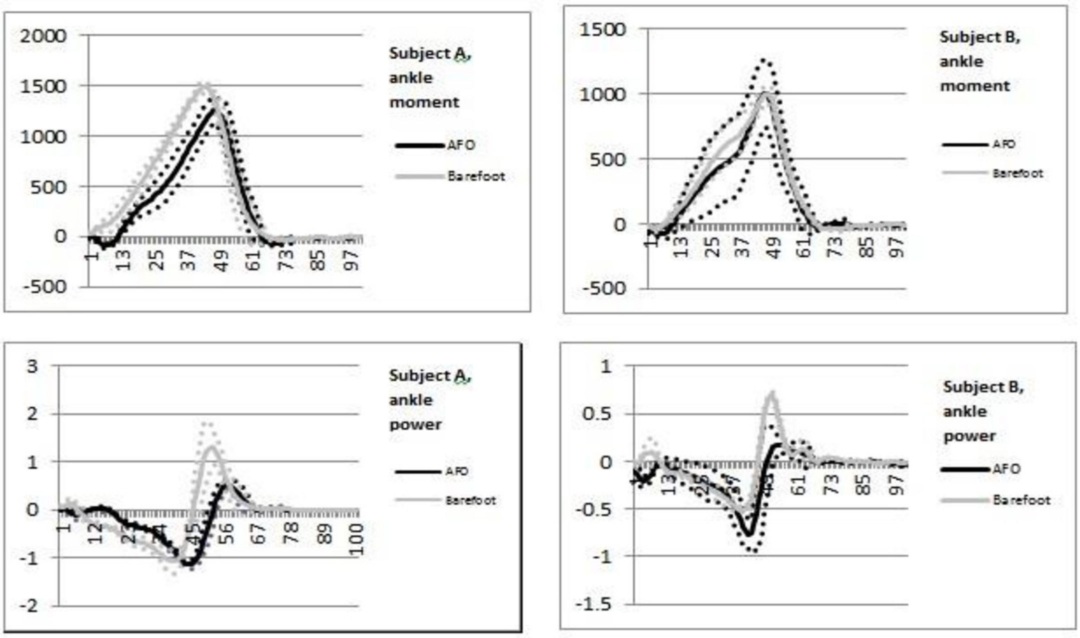

Figure 3: The sagittal Kinetic data at the ankle joint of subject A and B. Positive Values: plantar flexion, negative values: dorsiflexion. The barefoot curves are depicted with solid grey line, The AFO curves are depicted with solid black line, and standard deviations for every condition are depicted with dotted line.

tension angle during single support phase. These findings would reinforce the results of the previous studies comparing gait with and without an AFO having plantarflexion resistance [13, 15, 21].

Both walking velocity and stride length were increased for both subjects when they walked with the AFO relative to walking without the AFO. This improvement probably was resulted from more comfortable walking with the AFO than barefoot walking according to the patients' impressions of the AFOs. In one of our subjects, cadence, double support time, and single support time also improved using the AFO compared with barefoot walking. However, no differences were observed between two conditions in other subject for these parameters. It is possible, and more likely, that there was not enough training for using that. Therefore, it is our objective to include more time for patient's training in a bigger sample size in the future study.

In this study, use of the AFO increased the 1st peak (paretic leg in stance) of the body's COM vertical displacement in both subjects during gait on the affected leg. It is thought the vertical COM displacement is related to energy cost of gait [22]. In addition, the peak height of the COM trajectory for the paretic leg when measured during stance phase is generally lower than that measured for the non-paretic leg in the same phase in individuals with stroke hemiplegia [23]. So, it has been suggested that optimizing the vertical displacement of the body's COM could result in a more energy efficient gait [24]. On the other hand, stroke patients usually use hip hiking by increase of hip abduction on the non-paretic side as elevating the paretic pelvis in swing phase. This strategy may result in greater vertical displacement of the COM on the nonparetic leg [25, 26]. In this study, the use of the AFO also reduced 2st peak (nonparetic leg in stance) of the COM vertical displacement.

However, the reduction amounts of the peak pelvic obliquity in swing phase were just 1.61 degrees and 1.62 degrees. To understand the better influence of an AFO on the COM vertical displacement, however, enough instruction in long-term use of daily life should be given patient.

The new AFO generates the resistive moment to plantar flexion in the third rocker to prevent excessive plantar flexion. In both of subjects, we observed a reduction in ankle power resulted from the resistive moment to plantar flexion, the reduction in the ankle ROM and no improvement in the plantar flexion moment in terminal stance induced by the AFO. Therefore, the potential for generating power at the ankle has been limited by the AFO. In line with our findings, none of previous studies found an augmentation of the ankle power with mechanical AFOs in the terminal stance compared with walking without orthosis [27-30]. In this study, however, we evaluated the immediate effect of the AFO on gait biomechanics. On the other hand, There is some evidence that push-off improvement is an important function of the rocker bar modification in footwear [31, 32]. It is our objective to include this aspect in our future study.

There were a number of limitations to this study. The number of patients who participated in this study was small to gain statistical significance because this was a case report just evaluating the effect of the AFO on stroke gait. Moreover, the efficacy of the AFO with this mechanism must be proved by its long-term use in daily life in associated with gait training by physical therapist. On the other hand, larger studies are required to evaluate fully the temporal-spatial, kinematic, and kinetics parameters of gait as well as muscular activity between the new AFO and the conventional AFOs.

Conclusion

When the patient walked with the AFO, the angular changes of dorsiflexion on the paretic ankle joint during gait cycle were improved. The AFO also was able to control knee extension and increase the vertical displacement of the COM during stance phase on the affected leg. These changes indicate improvement of the first and the second rockers and swing phase gait but not third rocker function.

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Conflict of Interest

None

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