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Changes in angular kinematics of the paretic lower limb at different orthotic angles of plantar flexion limitation of an ankle-foot-orthosis for stroke patients

Hye Young Lee, PhD, PT[1)], Jeon Hyeong Lee, PhD, PT[2)], Kyoung Kim, PhD, PT[1)*]

[1)] *Department of Physical Therapy, College of Rehabilitation Science, Daegu University: 201 Daegudae-ro, Gyeongsan-si, Gyeongsangbuk-do 712-714, Republic of Korea*
[2)] *Department of Physical Therapy, Daegu Health College, Republic of Korea*

Abstract. [Purpose] An ankle-foot-orthosis (AFO) is an assistive brace that allows stroke patients to achieve an independent gait. Therefore, we examined whether or not the orthotic angle for plantar flexion limitation affects the kinematic parameters of the hip and knee joints on the affected side of patients with stroke. [Subjects and Methods] Fifteen patients with chronic hemiplegia were recruited for this study. Kinematic three-dimensional data was acquired, while patients walked along a walkway wearing AFOs under five different conditions of 0°, 5°, 10°, 15°, and 20° of plantar stop limitation angle in the orthotic joint. Peak angles of the hip and knee joints on the affected side were analyzed. [Results] At the peak angle of the knee joint, statistically significant differences were found only at mid-stance in the sagittal plane and the horizontal plane. However, no significant differences were observed among any of the orthotic limitation angles in the frontal plane. [Conclusion] According to the results, an orthotic limitation angle of more than 10° elicits changes in the knee joint angle at mid-stance in the sagittal and horizontal planes. This study provided basic data on postural changes of patients with stroke.

Key words: Stroke, Ankle-foot-orthosis, Gait

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INTRODUCTION

Stroke is a major cause of morbidity and mortality, and it is the third most common cause of death worldwide after heart disease and cancer[1)]. Many stroke patients suffer from locomotion deficits and postural control disability[2)]. These dysfunctions are caused by a variety of movement abnormalities, such as motor weakness caused by hemiparesis, proprioceptive deficits, and abnormal synergic patterns[3)]. In particular, motor disability in the knee and ankle joints, such as back knee and foot-drop, impair functional walking ability[4)]. In clinical settings, an ankle-foot-orthosis (AFO) is frequently prescribed to correct ankle joint alignment and facilitate normal gait patterns for hemiplegic stroke patients[5, 6)]. In addition, an AFO enhances walking function by providing stability and foot clearance during the stance and swing phases[7–10)].

AFO is typically designed for plantar flexion motion to be limited to under 90° in order to prevent foot-drop by limiting the plantar flexion range of motion of the talocrural joint[11, 12)]. Articulated AFOs with a plantar flexion stop elicit modest increases in walking speed and improve dorsiflexion in the swing and early stance phases[8, 9, 13)]. Some studies have suggested that a modification of the orthotic angle to limit plantar flexion would be an effective way of improving walking quality[8, 12, 14)]. However, according to the study of Mulroy et al., which investigated the effect of three different plastic AFOs on the gait of stroke patients in terms of plantar stop, dorsi-assist/stop, and rigid orthosis, the extent to which various angle limitations of plantar flexion affect hemiplegic gait cannot be elucidated[10)]. Accordingly, the controversy regarding the effect of modifications to the orthotic angle of AFOs continues.

Therefore, in the current study, we examined whether or not five different orthotic angles limiting plantar flexion (0°, 5°, 10°, 15°, and 20°) affect the kinematic parameters of the hip and knee joints on the affected side during level walking of patients with stroke.

SUBJECTS AND METHODS

Fifteen patients who suffered from stroke with cerebral hemorrhage or infarct were recruited for this study. All the subjects understood the purpose of this study and provided their written, informed consent prior to their participation in this experiment. This study was approved by the Insti-

*Corresponding author. Kyoung Kim (E-mail: kykim257@hanmail.net)

tutional Review Board of a Yeungnam University Hospital. The inclusion criteria were as follows: over six months since stroke onset; no severe cognitive impairment, a score of over 24 points in the Mini-Mental State Examination; the ability to walk independently for over 6 minutes without a walking assistive device; no experience of wearing a short leg brace since stroke onset, or for at least one month before this experiment; no visual or vestibular problems; a Modified Ashworth Scale assessment of < 2 for ankle plantar flexor spasticity of the affected side; and no joint contracture of the hip, knee, or ankle joints of the affected limb.

A plastic AFO that is typically used to prevent the foot-drop phenomenon exhibited by stroke patients was provided for all the subjects. First, a plaster pattern was made of the leg contours of each individual patient. An AFO made from 5 mm polypropylene was then manufactured, in which a hinge joint was embedded to limit the plantar flexion of the ankle joint to 90°. Fabric straps were attached to the dorsal region of the ankle, and to the lower and upper region of the leg to maintain proper leg and foot alignment. We drove a screw spike into posterior region of the AFO to adjust the angle of the orthotic ankle joint. According to the suggestion by Meadows et al., the angle of the orthotic joint was measured between the vertical line and the shank line of the AFO[15]. The angle of the orthotic joint was adjusted using the screw spike, so that the plantar flexion of the AFO was limited to 0°, 5°, 10°, 15°, and 20° from the original 90° of plantar flexion.

All the patients were instructed to walk comfortably along walkway that was 10 m long and 1.5 m wide, while wearing AFOs under five different conditions: 0°, 5°, 10°, 15°, and 20° of plantar stop limitation. For the actual measurements, 10 walking trials were performed for each condition, and 3 trials were selected for the final data analysis. The five different conditions were scheduled in a counterbalanced manner to prevent repetitive performance eliciting learning effects. In addition, the patients wore house shoes on the unaffected limb to adjust the height and length of both limbs when wearing the AFO. Before the actual measurements, practical trials were performed to familiarize the subjects with gait in a laboratory environment until the patients felt comfortable. Enough resting periods were provided at intervals between each condition to eliminate fatigue.

Kinematic three-dimensional data was acquired, using the VICON motion capture system (Vicon, Oxford, England) with 12 infrared cameras (MX-F40, Vicon, UK) and 25 mm reflective markers. According to the plug-in gait marker set, which is based on a kinematic segmental axis model, sixteen markers were placed on the pelvic region and segments of the lower extremities: the anterior sacral iliac spine, posterior sacral iliac spine, region of the thigh, the center of the patella, region of the lower leg, lateral malleolus, head of the second metatarsal bone, and calcaneus in bilateral lower limbs. The infrared cameras capture the position data of the individual markers and individual segment values at a rate of 120 Hz, and the VICON system reconstructs the images using a three-dimensional mechanical analysis of the individual joints, according to Euler's method[16]. All kinematic motion data were collected and analyzed using a Vicon workstation 5.2 and Polygon 3.1 software. For the main points of the gait cycle, based on Perry's suggestion, we selected initial contact (IC), mid-stance (MS), pre-swing (PS), and mid-swing (MSw)[17].

Table 1. Demographic information of the patients

Variables	Total
Gender (M/F)	15 (13/2)
Age (years)	46.5±14.4
Body weight (kg)	68.5±8.1
Height (cm)	168.3±5.7
Paretic side (right/left)	8/7
Mean duration since stroke onset (months)	26.5±12.5

Values are expressed as frequencies or mean ± SD

Statistical analysis of all kinematic data was performed using PASW 18.0 statistical package (SPSS, Chicago, IL, USA). Descriptive statistics were used to analyze demographic data: sex, age, height, weight, paretic side, and time since stroke. ANOVA was used to compare differences in angular data of the hip and knee joints among the five different conditions of orthotic joint limitation (0°, 5°, 10°, 15°, and 20°). The LSD procedure was adopted for posthoc analysis. Statistical significance was accepted for values of $p<0.05$.

RESULTS

The demographic information of all the patients is shown in Table 1. At the peak angular position of the hip joint on the affected side, no significant differences were found among the five different orthotic limitation angles (0°, 5°, 10°, 15°, and 20°) at IC, MS, PS, and MSw in the sagittal plane, frontal plane or horizontal plane. At the peak angular position of the knee joint on the affected side, statistical significance was found among the five different orthotic limitation angles (0°, 5°, 10°, 15°, and 20°) only at MS in the sagittal plane. LSD post-hoc analysis revealed that the peak knee angle of 0° orthotic limitation was different from those of 10°, 15°, and 20°. In addition, the peak knee angle at MS in the horizontal plane showed a significant difference among the five different orthotic angles (0°, 5°, 10°, 15°, and 20°). Post hoc analysis showed that the peak knee angle of only 20° orthotic limitation was significantly different from that of 0°. No significant differences were observed among the orthotic limitation angles in the frontal plane (Table 2).

DISCUSSION

The kinematic analysis results of the various orthotic limitation angles show that the flexion movement of the knee joint increased significantly at 10°, 15°, and 20° of orthotic limitation compared to 0° in mid-stance in the sagittal plane. In the horizontal plane, with 20°orthotic limitation the knee position was rotated externally compared to 0° in mid-stance.

The results of the present study are consistent with those of other studies that have determined the effect of AFOs on gait when dorsiflexion was limited[10, 18–20]. In the study of Mass, patients with cerebral palsy wore AFOs and the angle of the tibia was changed using a wedge. Their gaits were ob-

Table 2. Peak angles of the knee joint of the paretic lower limb of the five different posterior-stopped angles of AFO in the sagittal, frontal, and horizontal planes

Angle (°)			0	5	10	15	20
Knee	Sagittal plane	IC	9.6±1.9	11.9±1.8	12.3±1.8	14.4±1.6	15.4±1.7
		MS	5.5±0.8	8.0±0.7	9.3±0.7	11.5±0.7	13.6±0.8*
		PS	10.9±0.9	11.2±0.9	10.9±0.9	11.7±0.9	11.4±0.8
		MSw	53.9±2.9	56.2±2.6	55.7±2.9	56.7±2.7	55.9±2.7
	Frontal plane	IC	−0.2±0.6	0.3±0.6	0.1±0.5	0.6±0.6	0.6±0.7
		MS	−0.2±0.4	−0.3±0.4	0.0±0.4	0.1±0.5	0.1±0.6
		PS	−1.7±0.8	−1.9±0.8	−1.6±0.8	−1.6±0.8	−1.5±0.8
		MSw	11.0±4.1	10.5±4.2	10.8±4.1	10.4±4.1	10.3±4.1
	Horizontal plane	IC	2.0±1.3	3.9±1.5	4.3±1.1	6.3±1.4	6.8±1.4
		MS	−0.1±0.9	1.0±1.0	1.4±0.7	3.5±1.0	4.6±1.1*
		PS	3.4±0.9	3.7±1.1	3.1±1.0	4.4±1.1	4.1±1.0
		MSw	9.4±2.7	10.9±2.7	10.7±2.5	12.2±2.6	12.1±2.6

IC: Initial contact, MS: Mid-stance, PS: Pre-swing, MSw: Mid-swing. Values are presented as mean ± SD. *Significant difference ($p<0.05$) among five different orthotic limitation angles

served at shank angles from the vertical of 10°, 12°, and 14°. Their results showed that the angles of the maximum knee joint flexion at mid-stance were 27°, 32°, and 33°, while the angles of the mean knee joint flexion at the loading response and the end of the stance phase were 15°, 20°, and 22°, and there were significant differences between them, although no significant differences were found for the hip joints[20]. Jagadamma et al. conducted a study in which patients with cerebral palsy who experienced knee joint hyperextension in the stance phase had the angle from the vertical shank (SVA) adjusted to 5° and 10° using a wedge. They found no significant difference in the knee joints at the initial contact while significant differences of 2.6° and 3.7° were revealed at maximal extension of the knee joints during the stance phase. At maximum knee joint flexion, some change was shown at 20° and 25° but no significant difference was found[18]. Another study compared the knee angles at SVA of 0° and SVA of 14°, induced by a wedge, of adult patients with hemiplegia whose calf muscles had excessive tension. The results showed that the knee joint flexion changed from 8° to 18° at the initial contact while the maximum knee joint extension changed from −12° to 0° in the stance phase[19]. Moreover, Mulroy et al. compared the gait patterns of patients who had experienced stroke by dividing them into two groups: more than 0°, and between 10 and 15° of ankle joint dorsiflexion. Dorsiflexion was induced passively under the knee joint extension condition and the patients were asked to wear AFOs, HAFOs, and dorsiflexion-assist AFOs. In that study, subjects whose ankle dorsiflexion was more than 0° showed 8 to 12° of knee joint flexion at the initial contact, 16 to 20° of knee joint flexion at the loading response, and 6° of knee joint extension in the single limb stance phase. Subjects whose ankle dorsiflexion was between 10 and 15° showed 5 to 9° of knee joint flexion at the initial contact, 8 to 13° of knee joint flexion at the loading response, and −5 to 0° of knee joint extension in the single limb stance phase. These results show that all the orthoses increased knee joint flexion at the initial contact and loading response[10].

In summary, changes in angles of the knees were found at orthotic limitation angles over 10°, and a significant difference from 0° in the knee angle at mid-stance was found in the horizontal plane when 20° orthotic limitation was used. These results are explained by compensation movements due to the excessive change in the ankle joints, which resulted in about 5° rotation. Therefore, an adjustment of more than 10° in the orthotic limitation angle could result in changes in the knee joint angle at mid-stance, which may provide basic data for postural correction patients with hyperextension or spasticity in gait. However, long-term application could damage the knees because of the forced changes to the knee joint. In addition, an orthotic limitation angle of more than 20° should be avoided, because it may elicit changes not only in the flexion of the knee joints, but also in the rotational component.

An orthotic device is an important tool for aiding functional activities and the daily living of patients with gait disturbance due to various diseases. Many researchers have studied changes in SVA elicited by a variety of wedges and orthotic devices on flat ground. Better orthotic devices will be developed in the future using the results of these studies aiming to help patients with gait disturbance to have a more independent gait. However, this study was limited in the sense that subjects did not have enough time to adapt to the orthotic devices due to the one-off application of the plastic AFOs, and they were afraid of slipping as they had to walk without shoes, using the orthosis only, which made it more difficult for them to perform natural gait. Therefore, additional studies should be conducted addressing these limitations in the future.

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