J. Phys. Ther. Sci. 30: 966–970, 2018

The Journal of Physical Therapy Science

Original Article

Estimation of knee joint reaction force based on the plantar flexion resistance of an ankle-foot orthosis during gait

Masataka Yamamoto, PT, MS[1, 2)*], Koji Shimatani, PT, PhD[3)],
Masaki Hasegawa, PT, PhD[3)], Takuya Murata, BA[2)], Yuichi Kurita, PhD[2, 4)]

[1)] *Hyogo Prefectural Awaji Medical Center: 1-1-137 Shioya, Sumoto-shi, Hyogo 656-0021, Japan*
[2)] *Graduate School of Engineering, Hiroshima University, Japan*
[3)] *Faculty of Health and Welfare, Prefectural University of Hiroshima, Japan*
[4)] *PRESTO, JST, Japan*

Abstract. [Purpose] The purpose of this study was to investigate the effect of changing the plantar flexion resistance of an ankle-foot orthosis on knee joint reaction and knee muscle forces. Furthermore, the influence of an ankle-foot orthosis with an over-plantar flexion resistance function on knee joint reaction force was verified. [Participants and Methods] Ten healthy adult males walked under the following three conditions: (1) no ankle-foot orthosis, and with ankle-foot orthoses with (2) a strong and (3) a weak plantar flexion resistance (ankle-foot orthosis conditions). The knee flexion angle, quadricep muscle force, hamstring muscle force, and knee joint reaction force during the stance phase were measured using a motion analysis system, musculoskeletal model, and ankle-foot orthosis model. [Results] The peak knee joint reaction force, knee flexion angle, and quadricep muscle force in the early stance phase significantly increased in the strong plantar flexion resistance condition in comparison with the "no ankle-foot orthosis" condition. [Conclusion] Increased knee joint reaction force with over-plantar flexion resistance suggests that over-plantar flexion resistance causes various knee problems such as knee pain and knee osteoarthritis.

Key words: Joint reaction force, Orthosis, Gait

(This article was submitted Mar. 5, 2018, and was accepted May 7, 2018)

INTRODUCTION

Regaining gait function is one of the most important goals of patients with stroke. Post-stroke gait function is typically limited by the asymmetry of the gait pattern as well as by the decreased gait speed and step length[1, 2)]. An extension thrust pattern, buckling knee pattern, and stiff knee pattern are kinds of a typical abnormal gait[3)]. Mulroy et al.[4)] reported that patients with stroke and a slow velocity gait showed an excessive knee flexion or hyperextension in the mid stance. These abnormal knee movements seem to cause knee pain and deformity. Doruk[5)] reported that knee pain at rest and knee osteoarthritis had a negative effect on the ambulation level of patients with stroke. Therefore, it is important for these patients to improve their abnormal knee movements.

The use of an ankle-foot orthosis (AFO) is one of the recommended treatments for rehabilitation of patients with stroke. According to previous studies, the use of an AFO on the affected limb can improve balance, energy cost, gait speed, and overall gait biomechanics[6–8)]. In particular, an AFO with a plantar flexion resistance (PFR) function can generate a moment to resist the rapid ankle plantar flexion from initial contact to loading responses. The PFR function improves the first foot rocker and weight acceptance response of the hemiplegic lower limb, which positively influences gait speed[9)]. In addition,

*Corresponding author. Masataka Yamamoto (E-mail: qqeb2nc9k@circus.ocn.ne.jp)

another study reported that an AFO with a PFR reduced the incidence of genu recurvatum in patients with stroke[10]. In contrast, an over-PFR function, such as plantar flexion stop function, increased the knee flexion angle in the initial contact and loading response[11]. An over-PFR function may cause abnormal movements and increase the risk of knee pain and deformity. Therefore, the PFR function is very important, and the magnitude of the PFR of AFOs should be customized to each patient's body parameters and gait performance. However, whether over-PFR influences the knee joint loading is unclear. It is difficult to differentiate clearly the biomechanical effects of an AFO with a PFR function in patients with stroke because of the wide variability in the physical effects of stroke on gait function[12]. Thus, verification of an AFO with a PFR function is also necessary for healthy participants.

Recent advances in musculoskeletal modeling have enabled researchers to generate gait simulations in efforts to estimate muscle forces and joint reaction forces (JRF). A few studies have reported on the knee JRF during gait using a musculoskeletal model. Previous studies have reported that the knee JRF during self-selected walking speeds averaged 3.9 times of the body weight (BW) for healthy females and 3.4 times of the BW for healthy males[13]. When electromyography-driven models were used, the knee JRF exceeded 4.0 times of the BW[14].

The purpose of this study was to investigate the effect of changing the PFR of an AFO on the knee JRF and knee muscle forces. Furthermore, we verified the influence of an AFO with an over-PFR function on the knee JRF.

PARTICIPANTS AND METHODS

Ten healthy adult males participated in this study (mean age, 20.4 ± 1.26 years; height, 167.2 ± 2.9 cm; weight, 59.3 ± 5.4 kg). Only participants with no previous lower limb surgeries or no pain during gait were included in the study. All procedures were approved by the ethics committee of the Prefectural University of Hiroshima Faculty of Health and Welfare and were consistent with the Declaration of Helsinki principles. Written informed consent was obtained from participants (No. 15MH036). An AFO with an oil damper (Gait Solution Design, Kawamura Gishi, Japan) was used in this study (Fig. 1). This AFO generates a plantar flexion-resistive moment with no dorsiflexion-resistive moment, and the PFR can be changed continuously[15]. We set the PFR to be generated when the ankle angle is less than 5° of dorsiflexion.

Participants walked under three conditions: a no AFO condition and two different PFR AFO conditions (weak PFR [w-PFR] and strong PFR [s-PFR]). W-PFR was the most flexible, while s-PFR was the most rigid in this AFO. After sufficient practice, participants walked on a 10-m walkway at a self-selected comfortable speed in each condition. Participants wore the AFO on their right foot. The no AFO condition was measured first; thereafter, the PFR conditions were randomized. Participants wore the same type of shoes, with the size matched on both feet.

The three-dimensional coordinates of the markers and ground reaction forces were measured during gait using a VICON MX motion analysis system (Vicon, Oxford, UK). This system included 12 infrared cameras (sampling rate, 100 Hz) and six force plates (AMTI, Watterson, MA, USA; sampling rate, 1,000 Hz). For motion capture, the markers were affixed over the following anatomical landmarks: seventh cervical vertebrae, sterno-clavicular notch, xiphoid process, right scapular inferior angle, 10th thoracic vertebrae, and, bilaterally, the anterior and posterior acromion processes, anterior-superior iliac spines, posterior-superior iliac spines, lateral thighs, medial and lateral epicondyles of the femurs, lateral shanks, medial and lateral malleoli, calcaneus, head of the second and fifth metatarsals, and tip of the second toe. In the AFO conditions, the right medial and lateral malleoli markers were placed directly on the AFO to allow them to be visible.

These measured data were used for simulation using a musculoskeletal model and an AFO model. OpenSim, an open-source musculoskeletal modeling and simulation software platform[16], was used. The musculoskeletal model comprised 92 muscle-tendon actuators, with 23 degrees of freedom, as described by Zajac. This model uses the modified Hill-type model by Thelen[17]. We connected the AFO model of the ToyLanding model[18] with the musculoskeletal model (Fig. 2). This AFO model comprised a footplate and a cuff. The footplate was rigidly attached to the foot, with the cuff rigidly attached to the tibia. The footplate and cuff were connected at two hinge points, allowing both dorsiflexion and plantar flexion at the ankle joint. Both the stiffness and the ankle angle at which stiffness is exerted can be adjusted in the AFO. The PFR input to the AFO model was used as the experimental value by the torque sensor in each PFR condition.

To estimate the muscle forces and knee JRFs, we conducted the following procedure. First, the model was scaled to each participant based on the anatomical landmarks using a "scale tool." The dimensions of each body segment in the model were scaled on the basis of the relative distances between the pairs of the markers obtained from the motion-capture system and the corresponding virtual marker locations in the model during static trial[16]. After scaling, the joint angles were calculated by inverse kinematics, which minimized errors between experimental marker trajectories and virtual markers on the scaled model. Second, the residual reduction algorithm was used to minimize the effects of modeling and marker data-processing errors and to make it more dynamically consistent with the ground reaction force data. Third, computed muscle control (CMC) was used to calculate the ankle muscle force over the stance phase of a gait cycle. CMC is a simulation that drives the kinematic trajectory of a musculoskeletal model toward a set of desired accelerations[19, 20]. Finally, the knee JRF was estimated using joint reaction analysis. This force was estimated as a point load acting on the tibial plateau using the Newton-Euler equation[21, 22]. Only the muscles crossing the knee were used in this analysis: biceps femoris long head, biceps femoris short head, rectus femoris, vastus, tensor fasciae longus, sartorius, gracilis, and gastrocnemius.

We estimated the peak muscle force (quadricep muscle force and hamstring muscle force) from 0 to 50% stance phase

Fig. 1. Ankle-foot orthosis used in this study.

Fig. 2. Musculoskeletal model and ankle-foot orthosis model.

Table 1. Measured outcome data in each condition

Conditions	No AFO	w-PFR	s-PFR
Peak knee flexion angle (°)	17.8 ± 4.0	19.7 ± 3.5	21.2 ± 3.6*
Peak quadricep muscle force (N/kg)	18.3 ± 3.8	19.7 ± 3.1	20.9 ± 2.7*
Peak hamstring muscle force (N/kg)	11.7 ± 1.5	12.1 ± 1.7	11.8 ± 1.7
First peak vertical knee JRF (BW)	4.2 ± 0.4	4.4 ± 0.4*	4.6 ± 0.4*
Second peak vertical knee JRF (BW)	4.6 ± 0.6	4.8 ± 0.6	4.8 ± 0.5
Peak anterior knee JRF	1.5 ± 0.2	1.6 ± 0.2*	1.7 ± 0.2*, †

Average ± standard deviation. Using multiple comparisons (Shaffer's Modified Sequentially Rejective Bonferroni Procedure). *Significant difference at p<0.05 in the no AFO condition, †Significant difference at p<0.05 in the w-PFR condition.
AFO: ankle-foot orthosis; w-PFR: weak plantar flexion resistance; s-PFR: strong plantar flexion resistance; JRF: joint reaction force; BW: body weight.

(early stance phase) and knee JRF (vertical and anterior directions) in each condition. In addition, the peak knee flexion angle during the early stance phase was measured from the motion capture data. The peak knee flexion angle, peak muscle force, and peak knee JRF were evaluated using a one-way repeated measurement analysis of variance and multiple comparisons (Shaffer's Modified Sequentially Rejective Bonferroni Procedure) using R version 2.8.1 (CRAN, freeware). Statistical significance was set at p<0.05.

RESULTS

The results are shown in Table 1. The peak knee flexion angle during the early stance phase varied significantly across the three conditions (p=0.007). Multiple-comparison analysis again identified an increase in the peak knee flexion angle in the s-PFR condition compared with that in the no AFO condition (p=0.016).

The peak quadricep muscle force during the early stance phase also varied significantly across the three conditions (p=0.003). Multiple-comparison analysis again identified an increase in the peak quadricep muscle force in the s-PFR condition compared with that in the no AFO condition (p=0.018). The peak hamstring muscle force during the early stance phase did not vary significantly across the three conditions (p=0.338).

Figure 3 shows the vertical and anteroposterior knee JRFs during the stance phase in the no AFO condition from the estimated data. The first peak knee JRF varied significantly across the three conditions (p=0.005). Multiple-comparison analysis again identified an increase in the first peak vertical knee JRF in the s-PFR condition compared with that in the no AFO conditions (p=0.023). Furthermore, the first peak vertical knee JRF was also significantly higher in the w-PFR condition than in the no AFO condition (p=0.044). The second peak vertical knee JRF did not vary significantly across the three conditions (p=0.689).

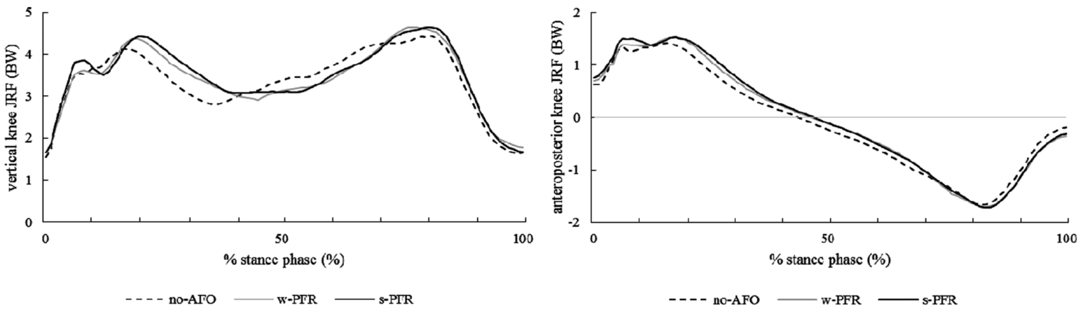

Fig. 3. Vertical and anteroposterior knee JRF during the stance phase.
AFO: ankle-foot orthosis; PFR: plantar flexion resistance; JRF: joint reaction force.

The peak anterior knee JRF varied significantly across the three conditions (p=0.012). Multiple-comparison analysis again identified an increase in the peak anterior knee JRF in the s-PFR condition compared with that in the no AFO and w-PFR conditions (p=0.044 and 0.044, respectively).

DISCUSSION

The purpose of this study was to investigate the effect of changing the PFR of an AFO on the knee JRF and knee muscle forces. Furthermore, we verified the influence of an AFO with an over-PFR function on the knee JRF.

The peak knee flexion angles during the early stance phase significantly increased in the s-PFR condition compared with that in the no AFO condition. This AFO generated a PFR when the ankle angle was less than 5° of dorsiflexion. Thus, the AFO generated an over-PFR to prevent plantar flexion, which caused a knee over-flexion with an over shank tilt. This result is consistent with previously reported effects of PFR[23]. However, there were no significant differences in the effects between the w-PFR and s-PFR conditions. The peak quadricep muscle force during the early stance phase significantly increased in the s-PFR condition compared with that in the no AFO condition. We surmised that the increase in the peak quadricep muscle force reflects the effect of the over-PFR on the weight acceptance response of the lower limb, preventing knee over-flexion. A previous study has reported that knee over-flexion in the stance phase increased the quadricep muscle force[24]. In contrast, the peak hamstring muscle force during the early stance phase was not significantly different among the conditions. Typically, the peak activity of the hamstring muscles starts from the pre-swing phase to the initial contact. Therefore, the hamstring muscles were not affected by the weight acceptance response.

In this study, the knee JRF was estimated separately in the vertical and anterior directions. The first peak knee JRF significantly increased in both AFO conditions compared with that in the no AFO condition. This increase was seemingly caused by the knee over-flexion and increase in the quadricep muscle force. A previous study has reported that the primary contributors to a compressive tibiofemoral force were the quadricep muscles in the early stance and knee flexion angle[21]. In addition, the peak anterior knee JRF during the early stance phase significantly increased in the s-PFR condition compared with that in the no AFO and w-PFR conditions. Another previous study has reported that the knee extension moment from the loading response increased using a rigid AFO[11]. We surmised that the increase in the peak anterior knee JRF during the early stance phase is attributable to the knee over-flexion by the over-PFR function. By contrast, the second peak knee JRF was not significantly different among the conditions. Generally, the PFR function of the AFO was generated by the ankle plantar flexion, such as loading response and pre-swing phase. Therefore, the PFR function did not affect the second peak knee JRF.

Our study investigated the effect of changing the PFR of an AFO on the knee JRF and knee muscle force. In addition, we verified the influence of an AFO with an over-PFR function on the knee JRF. The peak knee JRF, peak knee flexion angle, and peak quadricep muscle force in the early stance phase increased the over-PFR condition. These results suggest that an over-PFR causes various knee problems, such as knee pain and knee osteoarthritis.

Conflicts of interest

None.

REFERENCES

1) Wall JC, Turnbull GI: Gait asymmetries in residual hemiplegia. Arch Phys Med Rehabil, 1986, 67: 550–553. [Medline]
2) Kim CM, Eng JJ: Magnitude and pattern of 3D kinematic and kinetic gait profiles in persons with stroke: relationship to walking speed. Gait Posture, 2004, 20: 140–146. [Medline] [CrossRef]
3) De Quervain IA, Simon SR, Leurgans S, et al.: Gait pattern in the early recovery period after stroke. J Bone Joint Surg Am, 1996, 78: 1506–1514. [Medline] [CrossRef]

4) Mulroy S, Gronley J, Weiss W, et al.: Use of cluster analysis for gait pattern classification of patients in the early and late recovery phases following stroke. Gait Posture, 2003, 18: 114–125. [Medline] [CrossRef]
5) Doruk P: The impact of knee osteoarthritis on rehabilitation outcomes in hemiparetic stroke patients. J Back Musculoskeletal Rehabil, 2013, 26: 207–211. [Medline] [CrossRef]
6) Hesse S, Werner C, Matthias K, et al.: Non-velocity-related effects of a rigid double-stopped ankle-foot orthosis on gait and lower limb muscle activity of hemiparetic subjects with an equinovarus deformity. Stroke, 1999, 30: 1855–1861. [Medline] [CrossRef]
7) Tyson SF, Kent RM: Effects of an ankle-foot orthosis on balance and walking after stroke: a systematic review and pooled meta-analysis. Arch Phys Med Rehabil, 2013, 94: 1377–1385. [Medline] [CrossRef]
8) Tyson SF, Sadeghi-Demneh E, Nester CJ: A systematic review and meta-analysis of the effect of an ankle-foot orthosis on gait biomechanics after stroke. Clin Rehabil, 2013, 27: 879–891. [Medline] [CrossRef]
9) Nolan KJ, Yarossi M: Preservation of the first rocker is related to increases in gait speed in individuals with hemiplegia and AFO. Clin Biomech (Bristol, Avon), 2011, 26: 655–660. [Medline] [CrossRef]
10) Kobayashi T, Orendurff MS, Singer ML, et al.: Reduction of genu recurvatum through adjustment of plantarflexion resistance of an articulated ankle-foot orthosis in individuals post-stroke. Clin Biomech (Bristol, Avon), 2016, 35: 81–85. [Medline] [CrossRef]
11) Mulroy SJ, Eberly VJ, Gronely JK, et al.: Effect of AFO design on walking after stroke: impact of ankle plantar flexion contracture. Prosthet Orthot Int, 2010, 34: 277–292. [Medline] [CrossRef]
12) Guillebastre B, Calmels P, Rougier P: Effects of rigid and dynamic ankle-foot orthoses on normal gait. Foot Ankle Int, 2009, 30: 51–56. [Medline] [CrossRef]
13) Kuster MS, Wood GA, Stachowiak GW, et al.: Joint load considerations in total knee replacement. J Bone Joint Surg Br, 1997, 79: 109–113. [Medline] [CrossRef]
14) Winby CR, Lloyd DG, Besier TF, et al.: Muscle and external load contribution to knee joint contact loads during normal gait. J Biomech, 2009, 42: 2294–2300. [Medline] [CrossRef]
15) Yamamoto S, Hagiwara A, Mizobe T, et al.: Development of an ankle-foot orthosis with an oil damper. Prosthet Orthot Int, 2005, 29: 209–219. [Medline] [CrossRef]
16) Delp SL, Anderson FC, Arnold AS, et al.: OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans Biomed Eng, 2007, 54: 1940–1950. [Medline] [CrossRef]
17) Thelen DG: Adjustment of muscle mechanics model parameters to simulate dynamic contractions in older adults. J Biomech Eng, 2003, 125: 70–77. [Medline] [CrossRef]
18) OpenSim documentation: simulation-based design to prevent ankle injuries. https://simtk-confluence.stanford.edu:8443/display/OpenSim/Simulation-Based+Design+to+Prevent+Ankle+Injuries (Accessed Feb. 4, 2018)
19) Thelen DG, Anderson FC, Delp SL: Generating dynamic simulations of movement using computed muscle control. J Biomech, 2003, 36: 321–328. [Medline] [CrossRef]
20) Thelen DG, Anderson FC: Using computed muscle control to generate forward dynamic simulations of human walking from experimental data. J Biomech, 2006, 39: 1107–1115. [Medline] [CrossRef]
21) Steele KM, Demers MS, Schwartz MH, et al.: Compressive tibiofemoral force during crouch gait. Gait Posture, 2012, 35: 556–560. [Medline] [CrossRef]
22) Richards C, Higginson JS: Knee contact force in subjects with symmetrical OA grades: differences between OA severities. J Biomech, 2010, 43: 2595–2600. [Medline] [CrossRef]
23) Kobayashi T, Singer ML, Orendurff MS, et al.: The effect of changing plantarflexion resistive moment of an articulated ankle-foot orthosis on ankle and knee joint angles and moments while walking in patients post stroke. Clin Biomech (Bristol, Avon), 2015, 30: 775–780. [Medline] [CrossRef]
24) Perry J, Antonelli D, Ford W: Analysis of knee-joint forces during flexed-knee stance. J Bone Joint Surg Am, 1975, 57: 961–967. [Medline] [CrossRef]