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

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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

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 length1-3). An extension thrust pattern, buckling knee pattern, and stiff knee pattern are kinds of a typical abnormal gait3). 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. Doruk5) 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 biomechanics6-8). 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 speed9). In addition,
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 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 (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. 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 (quadriceps muscle force and hamstring muscle force) from 0 to 50% stance phase
(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 quadriceps 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 quadriceps 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 (\( p=0.338 \)).

**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 quadriceps 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.

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 condition (\( p=0.023 \)). 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 (\( p=0.689 \)).
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. However, there were no significant differences in the effects between the w-PFR and s-PFR conditions. The peak quadriceps 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 quadriceps 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 quadriceps muscle force. 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 quadriceps muscle force. A previous study has reported that the primary contributors to a compressive tibiofemoral force were the quadriceps muscles in the early stance and knee flexion angle. 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. 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 quadriceps 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.

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