Journal of Electromyography and Kinesiology 46 (2019) 63–69

Contents lists available at ScienceDirect

Journal of Electromyography and Kinesiology

journal homepage: www.elsevier.com/locate/jelekin

Effects of altering plantar flexion resistance of an ankle-foot orthosis on muscle force and kinematics during gait training

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ARTICLE INFO

Keywords:
Musculoskeletal model
Joint angle
Spatiotemporal parameters

ABSTRACT

Ankle-foot orthosis (AFO) can improve gait in stroke patients. Addition of plantar flexion resistance (PFR) can improve the first foot rocker function. However, the effect of changing the PFR on the ankle muscle force during gait training is unclear. This study aimed to determine the effect of changing the PFR of an AFO on spatiotemporal parameters (speed, bilateral step length, and cadence), peak angle of ankle plantar flexion and knee flexion, and muscle force (tibialis anterior [TA], medial head of the gastrocnemius [MGAS], and soleus) during early stance using a musculoskeletal model. Ten healthy adult men walked under five conditions: a no-AFO condition and PFR conditions 1–4. Spatiotemporal parameters and peak joint angles during the early stance phase were measured from experimental data, with muscle force estimated from simulations of a musculoskeletal model. Increasing the PFR of the AFO decreased TA muscle force and increased MGAS muscle force but had no influence on spatiotemporal parameters and joint angles. Adjustment of the PFR modifies the muscle force around the ankle, which can maximize the effect of AFO during gait training.

1. Introduction

Regaining independent gait function is often an important goal in stroke rehabilitation. Post-stroke gait function is typically limited by decreased gait speed and step length, as well as asymmetry of gait pattern (Wall and Turnbull, 1986; Kim and Eng, 2004), which usually includes a foot drop and the absence of a heel strike on the hemiplegic lower limb. According to previous studies, the use of an ankle-foot orthosis (AFO) on the affected limb can improve balance, energy cost, gait speed, and overall gait biomechanics (Hesse et al., 1999; Tyson and Kent, 2013; Tyson et al., 2013). In particular, AFOs with a plantar flexion resistance (PFR) function can generate a moment to resist the rapid ankle plantar flexion from initial contact to the loading response (early stance). Therefore, this resistive moment improves the first foot rocker function and weight acceptance response of the lower limb, which positively influences gait speed (Nolan and Yarossi, 2011; Yamamoto et al., 2009). The magnitude of the PFR of an AFO should be customized to each stroke patient's body parameters and gait performance. Yamamoto et al. (2013) reported that the forward component of ground reaction force and shank vertical angle were affected by the PFR. Kobayashi et al. (2015) reported that changing the magnitude of the PFR influenced the knee and ankle joint angles and moments of the hemiplegic lower limb. In addition, an AFO with a PFR function reduces genu recurvatum in stroke patients (Kobayashi et al., 2016). The PFR of an AFO also influences muscle activity; lowering of gastrocnemius electromyographic activity during the loading response was reported with the use of an AFO with plantar flexion stop (Ohata et al., 2011). However, it is difficult to clearly differentiate the purely biomechanical effects of the PFR in stroke patients owing to the wide variability in physical effects of stroke on gait function (Guillebastre et al., 2009). Moreover, a previous systematic review indicated that future studies should evaluate the mechanical characteristics of AFO (i.e. resistance or material used) and the magnitude of the resistive moment should be tuned to each patient's condition (Daryabor et al., 2018). Therefore, studies in healthy participants are needed to initially investigate the true effect of the magnitude of the PFR.

A musculoskeletal model can be used to estimate the effects of an AFO on muscle force, along with other discrete parameters. A few studies have reported the effect of AFO resistance on muscle or joint function during gait training. Crabtree and Higginson (2009) reported that a rigid AFO decreased activity in the tibialis anterior (TA) compared to that under a no-AFO condition; however, the effect of this

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https://doi.org/10.1016/j.jelekin.2019.03.003
Received 28 June 2018; Received in revised form 6 March 2019; Accepted 6 March 2019
1050-6411/

change in muscle activation pattern on gait pattern was not evaluated. Choi et al. (2016) reported that using an AFO with footwear increased the length of the paretic gastrocnemius during the swing phase of gait. Arch et al. (2016) reported the effectiveness of a passive-dynamic AFO in substituting function of the soleus (SOL) muscle. However, whether the magnitude of the PFR in an AFO influences the ankle muscle force and modifies spatiotemporal parameters is unclear.

Therefore, the purpose of this study was to investigate the effect of altering the PFR of an AFO on spatiotemporal parameters, gait kinematics, and the ankle muscle force (TA, medial head of the gastrocnemius [MGAS], and SOL) during early stance using a musculoskeletal model. We hypothesized that TA muscle force would decrease with an increasing PFR. Additionally, we hypothesized that the MGAS and the SOL muscle forces would increase with an increasing PFR.

2. Methods

2.1. Participants and bracing conditions

Ten healthy men participated in this study (mean age, 20.4 ± 1.26 years; height, 1.67 ± 0.03 m; weight, 59.3 ± 5.4 kg). Only participants with no previous lower limb surgeries or no pain during gait training were included. All procedures were approved by the ethics committee of the Faculty of Health and Welfare at Prefectural University of Hiroshima and were consistent with the Declaration of Helsinki. Written informed consent was obtained from all participants. An AFO with a PFR function (Gait Solution Design, Kawamura Gishi, Japan) was used in this study (Fig. 1). This AFO comprised a footplate and a cuff and generated a PFR moment without a dorsiflexion-resistive moment. The PFR could be changed continuously, from a setting of 1 (very flexible) to 4 (rigid) (Yamamoto et al., 2005). In our study, the PFR was set to be generated when the ankle was in less than 5° of dorsiflexion.

2.2. Measurement protocol

All participants walked under five conditions: a no-AFO condition and four different PFR AFO conditions (PFR1, PFR2, PFR3, and PFR4). The PFR of each condition was tuned on the basis of the setting of the AFO: the most flexible under PFR1 and the most rigid under PFR4. After sufficient practice, all participants walked on a 10-m walkway at a self-selected, comfortable speed during each condition. Although fixing the gait speed would reduce data variability by restricting the chosen gait strategy, this study used self-selected speed because previous studies reported that AFO affects gait speed (Guillebastre et al., 2009; Tyson and Kent, 2013) and thus may be related to the kinematics of using AFO gait. As a result, participants were not asked to maintain a consistent gait speed. The no-AFO condition was measured first, followed by the four PFR AFO conditions. All participants wore the AFO on their right foot, and the order of PFR conditions was randomized. Additionally, all participants wore the same type of shoes, with the size matched on both feet.

Fig. 1. An ankle-foot orthosis (AFO) with plantar flexion resistance (PFR) function was used in this study.

The three-dimensional coordinates of the markers and ground reaction forces were measured during gait training 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, 1000 Hz). For motion capture, the Plug-in gait marker set was used, with markers affixed over the following anatomical landmarks: the seventh cervical vertebra, sternoclavicular notch, xiphoid process, right scapular inferior angle, and the tenth thoracic vertebra. Additionally, the markers were affixed bilaterally over 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, calcanei, head of the second and fifth metatarsals, and tip of the second toes. In PFR AFO conditions, the right medial and lateral malleoli markers were placed directly on the AFO to allow their visibility.

2.3. Musculoskeletal model and AFO model

The motion data were used as inputs for simulation of a musculoskeletal model and an AFO model. OpenSim, an open-source musculoskeletal modeling and simulation software platform (Delp et al., 2007), was used for all simulations (Fig. 2). The musculoskeletal model comprised 92 muscle-tendon actuators, with 23 degrees of freedom. This model uses the modified Hill-type model (Thelen, 2003).

We connected the AFO model of the ToyLanding model (OpenSim Documentation, 2018) with the musculoskeletal model. 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

Fig. 2. Musculoskeletal model and ankle-foot orthosis (AFO) model.

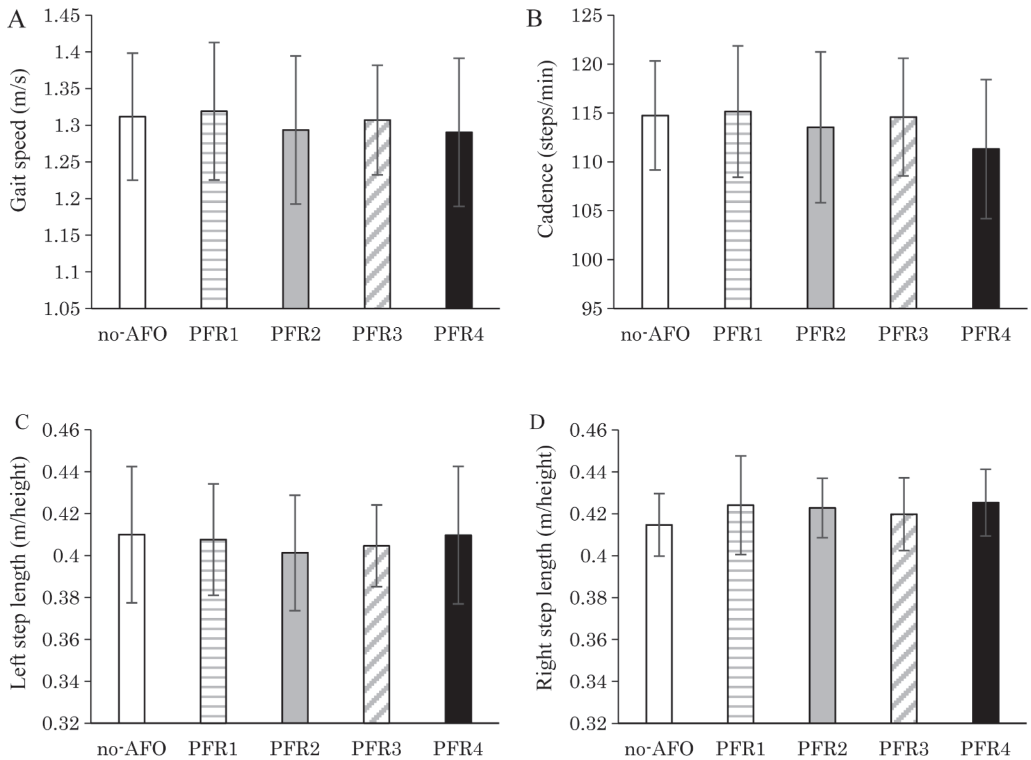

Fig. 3. Spatiotemporal parameters for each condition: gait speed (A), cadence (B), left step length (C), and right step length (D). No significant differences were detected.

dorsiflexion and plantar flexion movements at the ankle joint. Both the resistance and the ankle angle at which resistance was exerted could be adjusted in the AFO, from each PFR condition. In our study, the PFR was set to be generated when the ankle was in less than 5° of dorsiflexion. The PFR input to the AFO model used an experimental value that was measured using a torque sensor in each PFR condition. PFR values in each condition were as follows: PFR1, 0.56 Nm/°; PFR2, 0.76 Nm/°; PFR3, 0.95 Nm/°; and PFR4, 1.47 Nm/°.

The musculoskeletal model was scaled to match the anthropometry of each participant, using "scale tool" in OpenSim. In the model, the dimensions of each body segment were scaled on the basis of the relative distances between pairs of markers that were obtained from the motion-capture system and the corresponding virtual marker locations during a static trial in the model (Hamner et al., 2010; Thelen et al., 2003). After scaling, the joint angles were calculated using inverse kinematics, with the optimal solution minimizing the error between experimental marker trajectories and virtual markers on the scaled model. The residual reduction algorithm was used to minimize the effects of modeling and marker data processing errors, and to improve the dynamic consistency of the model with the measured ground reaction force data. Last, computed muscle control was used to calculate the ankle muscle force over the stance phase of a gait cycle.

Computed muscle control is a simulation that drives the kinematic trajectory of a musculoskeletal model toward a set of desired kinematics (Thelen et al., 2003; Thelen and Anderson, 2003).

$$\ddot{q}_{des}(t+T) = \ddot{q}_{exp}(t+T) + k_v[\dot{q}_{exp}(t) - \dot{q}(t)] + k_p[q_{exp}(t) - q(t)] \quad (1)$$

A set of desired accelerations, $\ddot{q}_{des}$, are first computed, which drive the generalized coordinates and speeds of the model (q and $\dot{q}$) toward the experimental kinematics (q_{exp} and $\dot{q}_{exp}$) every T (0.01 s). Feedback gains, kv and kp, weigh the current velocity and position errors, respectively (Thelen and Anderson, 2003).

$$J = \sum_{i=1}^{m} V_i[a_i(t+T)]^2 \quad (2)$$

Eq. (2) is the objective function of the optimized calculation that minimizes the required muscle activity and force. V_i is the volume of muscle (i), and a_i ($t + T$) is the activation of i at $t + T$ corresponding to the desired muscle force (Thelen and Anderson, 2003).

After computed muscle control, a correlation between the estimated ankle muscle activity and experimental muscle activity was also confirmed using electromyography (EMG) in each condition (Yamamoto et al., 2017). Estimated muscle force data were used because the Pearson correlation coefficient was similar to or more than that reported in previous study (Wibawa et al., 2016).

2.4. Data analysis

The effects of changing the PFR of the AFO spatiotemporal parameters and lower limb position were measured directly from the motion capture data during the early stance phase. The following variables were evaluated: gait speed, bilateral step length and cadence, and peak angles of ankle plantar flexion and knee flexion. Step length was normalized by height for between-participant comparisons. Peak TA, MGAS, and the SOL muscle forces were estimated from the simulation of the model. We selected the early stance phase because the PFR affects the function of the first foot rocker through assisting ankle movement (Nolan and Yarossi, 2011; Yamamoto et al., 2009). The ankle muscle force was normalized to body weight for between-participant comparisons. All data were averaged over three steps for each condition, with step length defined from the heel contact to toe off. Spatiotemporal parameters from the experimental data and muscle forces were compared among the five walking conditions using a one-way repeated measures analysis of variance and multiple comparisons (Tukey's test). Normal distribution of data was verified using the Shapiro-Wilk test; hence, parametric tests were performed. Statistical analyses were conducted using SPSS statistics version 20 (SPSS, Chicago, IL, USA). Data are presented as mean ± standard deviation. Statistical significance was set at $p < 0.05$.

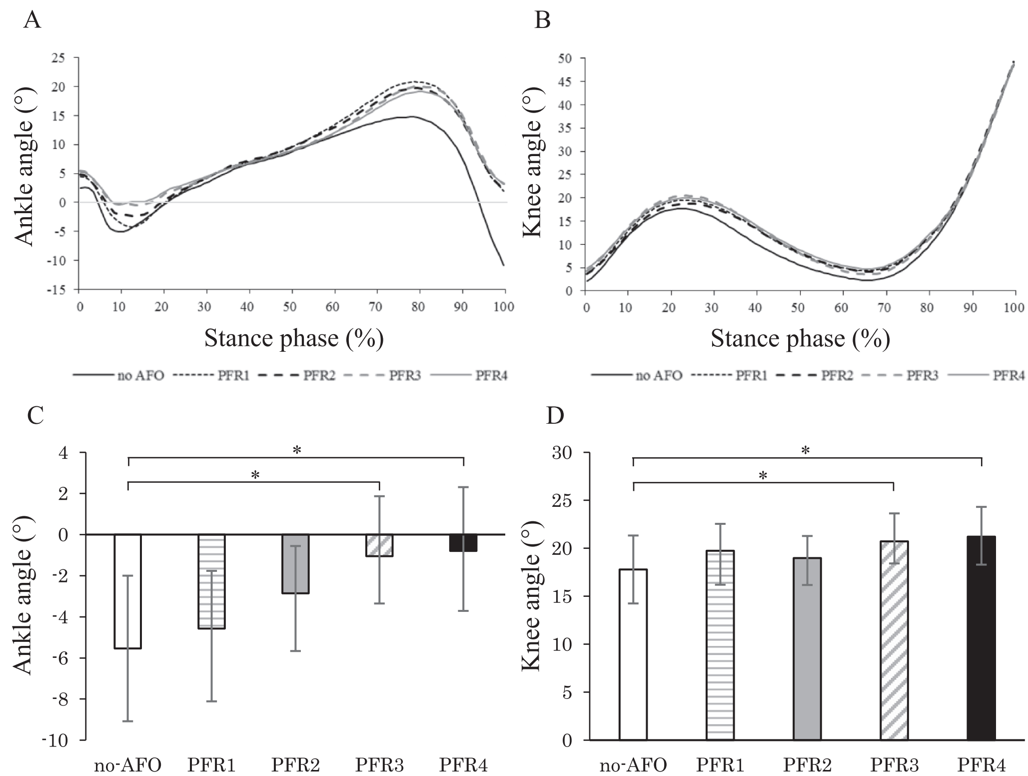

Fig. 4. Ankle (A) and knee (B) joint angles during the stance phase. Peak ankle plantar flexion angle (C) and peak knee flexion angle (D) in the early stance phase. Dorsiflexion and flexion are defined as positive. *significant at $p < 0.05$.

3. Results

We used a combination of motion analysis and modeling to quantify the effect of altering the PFR of an AFO on gait kinematics in healthy adults. The results for each measured outcome variable are shown below.

Fig. 3 shows the spatiotemporal parameters during each condition from the experimental data. No significant differences were identified for any parameters among the conditions.

Fig. 4 shows the knee and ankle angles during the stance phase and the peak angle in early stance for each condition from the experimental data. The peak ankle plantar flexion angle varied significantly across the five conditions ($p = 0.003$). The peak values for each condition were as follows: no-AFO, 5.54° ± 3.54°; PFR1, 4.57° ± 2.81°; PFR2, 2.86° ± 2.30°; PFR3, 1.05° ± 2.92°; and PFR4, 0.80° ± 3.10°. The multiple-comparison analysis identified a significant decrease in peak ankle plantar flexion angle for the PFR3 and PFR4 conditions compared with the no-AFO condition ($p = 0.018$ and $p = 0.01$, respectively). In addition, peak knee flexion angle varied significantly across the five conditions ($p = 0.005$) as follows: no-AFO, 17.79° ± 3.99°; PFR1, 19.74° ± 3.55°; PFR2, 18.98° ± 4.35°; PFR3, 20.72° ± 3.77°; and PFR4, 21.21° ± 3.58°. The multiple-comparison analysis similarly identified an increase in peak knee flexion angle for the PFR3 and PFR4 conditions compared with the no-AFO condition ($p = 0.022$ and $p = 0.005$, respectively).

Before the analyzed muscle force result, we investigated the comparison between estimated muscle activity and experimental muscle activity using EMG on TA and MGAS activity in each condition (Yamamoto et al., 2017). The Pearson correlation coefficient (r) was used to compare these data based on previous study (Wibawa et al., 2016). The r of TA muscle activity is as follows: no-AFO, 0.67; PFR1, 0.63; PFR2, 0.56; PFR3, 0.60, PFR4, 0.83 ($p < 0.001$, respectively). The r of MGAS muscle activity is as follows: no-AFO, 0.72; PFR1, 0.65; PFR2, 0.63; PFR3, 0.60, PFR4, 0.81 ($p < 0.001$, respectively). We used the estimated muscle force data because they were similar to or more than that reported in previous study (Wibawa et al., 2016).

Fig. 5 shows the ankle muscle force during the stance phase and peak muscle force in early stance for each condition from the modeling data. Peak TA muscle force during the early stance phase varied significantly across the five conditions ($p < 0.001$) as follows: no-AFO, 12.35 ± 1.05 N/kg; PFR1, 11.37 ± 0.80 N/kg; PFR2, 10.90 ± 1.72 N/kg; PFR3, 10.67 ± 1.12 N/kg; and PFR4, 9.97 ± 1.53 N/kg. The multiple-comparison analysis indicated a significant decrease in peak TA muscle force for the PFR2, PFR3, and PFR4 conditions compared with the no-AFO condition ($p = 0.028$, $p = 0.008$, and $p < 0.001$, respectively). Furthermore, the peak TA muscle force was significantly lower in the PFR4 condition than that in the PFR1 condition ($p = 0.035$). The peak MGAS muscle force also varied across the five conditions ($p < 0.001$) as follows: no-AFO, 5.70 ± 0.70 N/kg; PFR1, 6.55 ± 1.27 N/kg; PFR2, 7.02 ± 1.00 N/kg; PFR3, 7.72 ± 1.87 N/kg; and PFR4, 8.38 ± 1.77 N/kg. The multiple-comparison analysis indicated a significant increase in the peak MGAS muscle force for the PFR2, PFR3, and PFR4 conditions compared with the no-AFO condition ($p = 0.021$, $p < 0.001$, and $p < 0.001$, respectively). Furthermore, the peak MGAS muscle force was also significantly higher in the PFR4 condition than that in the PFR1 and PFR2 conditions ($p < 0.001$ and $p = 0.018$, respectively). The peak SOL muscle force also varied significantly across the five conditions ($p = 0.020$) as follows: no-AFO, 9.06 ± 3.23 N/kg; PFR1, 12.35 ± 3.41 N/kg; PFR2, 10.30 ± 3.19 N/kg; PFR3, 11.86 ± 1.62 N/kg; and PFR4, 12.44 ± 2.89 N/kg. The multiple-comparison analysis indicated a significant increase in the peak SOL muscle force during the PFR4 condition compared with the no-AFO condition ($p = 0.040$).

4. Discussion

The purpose of this study was to investigate the effect of changing

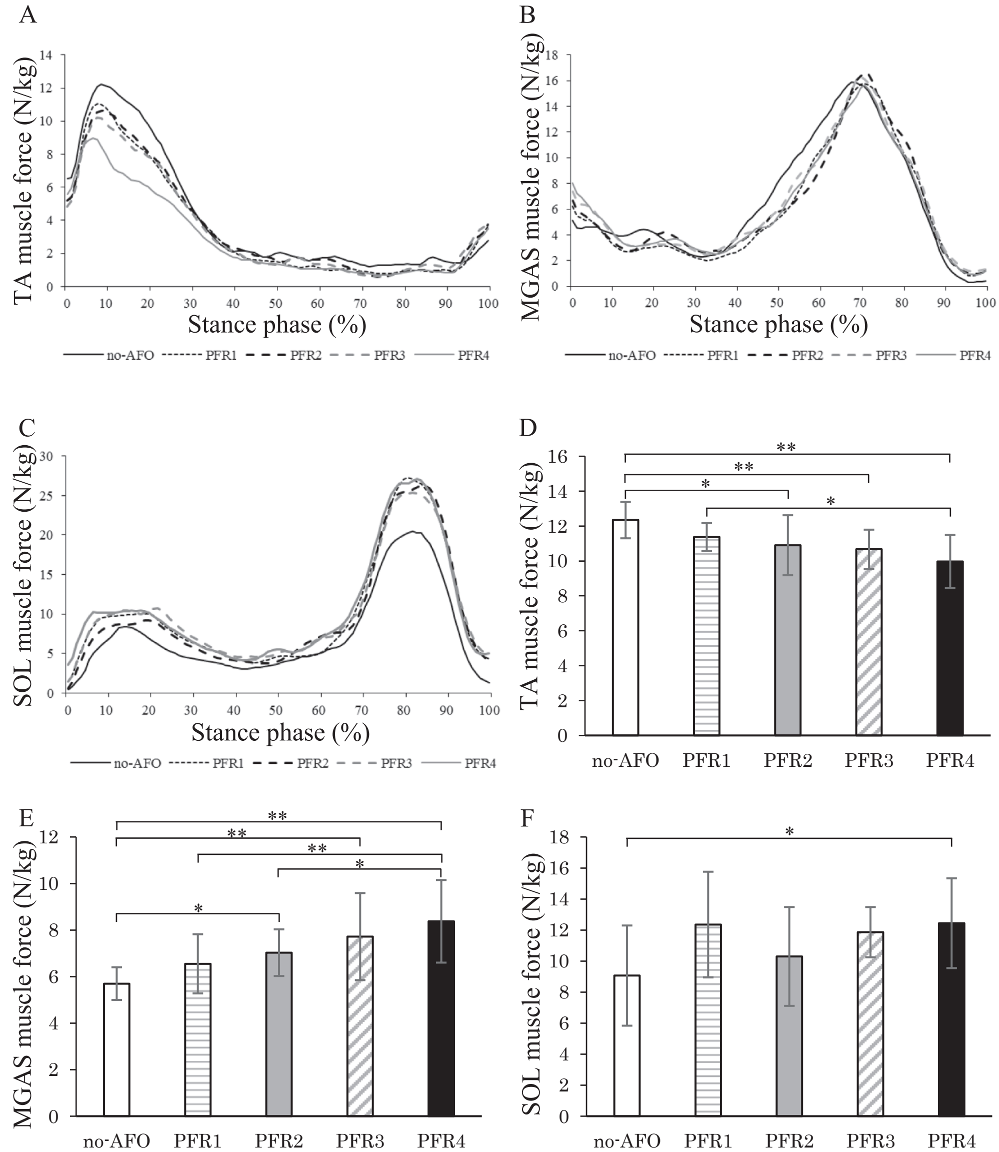

Fig. 5. The ankle muscle force during the stance phase: (A) tibialis anterior (TA), (B) medial head of the gastrocnemius (MGAS), and (C) soleus (SOL). The peak ankle muscle force in the early stance phase: (D) TA, (E) MGAS, and (F) SOL. *significant at $p < 0.05$; **significant at $p < 0.01$.

the PFR of an AFO on the force of selected muscles of the ankle joint during gait training using a musculoskeletal model. Although the estimated muscle activity data were similar to or more than that reported in previous studies, the present study did not use muscle activity but muscle force for closing experimental kinetics and kinematics data. Because muscle activity may be affected by the difference between the musculoskeletal model's maximum muscle force and each participant's maximum muscle force, we used muscle force data based on kinetics and kinematics data. Results of the spatiotemporal parameters and joint angle did not show a significant difference between the PFR conditions; however, altering the PFR significantly affected the ankle muscle force in healthy adults. The limitations of our study should be considered before application of our findings in practice. Foremost, we did not examine the effects of changing the PFR on forces of the hip and knee muscles, and we only considered joint positions and not velocities. Additionally, we evaluated the effects of changing the PFR of an AFO on the gait of healthy individuals only; however, our findings are important for the assessment of gait and adjustment of the PFR of an AFO in the rehabilitation of stroke patients.

No significant differences in spatiotemporal parameters (namely gait speed, cadence, and step length for both legs) were observed under all conditions. AFO has been shown to increase gait speed and step length in stroke patients (Tyson and Kent, 2013). In our study, participants were healthy adults; therefore, their gait ability and body function were high with little room for improvement with an AFO. In addition, the phase of ankle plantar flexion during a normal gait cycle is short. A previous study also reported that gait speed and step length in healthy adults were not significantly different in various dynamic-type AFO conditions (Guillebastre et al., 2009).

Moreover, the AFO used in our study generated a PFR when the ankle was in a position of less than 5° of dorsiflexion, providing a sufficient PFR to prevent plantar flexion (Fig. 4). The resultant decrease in peak angles of ankle plantar flexion and knee flexion in the PFR3 and PFR4 conditions, when compared with the no-AFO condition, is consistent with that reported previously (Kobayashi et al., 2015).

The TA muscle force, which was estimated from the musculoskeletal model, was lower during the PFR2, PFR3, and PFR4 conditions than in the no-AFO condition. A previous study reported that an AFO with a PFR assists with eccentric contraction of dorsiflexors in the early phase of stance (Yamamoto et al., 1993). Another study reported that a PFR-type AFO decreased the TA muscle activity compared with the no-AFO condition (Mulroy et al., 2010), which is consistent with our finding of an inverse relationship between the PFR magnitude and TA muscle force in normal adults. Furthermore, the TA muscle force was significantly different between the PFR1 and PFR4 conditions in this study. Another previous study reported that the TA muscle activity during gait training is related to gait speed in stroke patients (Mulroy et al., 2003). Therefore, adjusting magnitude of the PFR, which influences activity of the TA muscle, could be customized for individual stroke patients in order to improve gait. In comparison, the MGAS and SOL muscle forces significantly increased with an increasing PFR compared to those under the no-AFO condition. This increase in activity reflects the effect of the AFO on the weight-acceptance response of the lower limb, thus preventing plantar flexion and total foot contact during the early stance. Specifically, this required our healthy adult participants to increase the contraction force of the MGAS during the PFR AFO conditions compared with the no-AFO condition. Conversely, no significant differences were observed in the peak SOL muscle force for the other PFR conditions, which is consistent with the findings of a previous study; the study reported increased gastrocnemius activation but not SOL activation during the loading response when using an AFO with plantar flexion stop (Ohata et al., 2011). In our study, compared to that under the no-AFO condition, the peak SOL muscle force was only significantly increased during the PFR4 condition, which resulted from an excessive shank tilt with ankle dorsiflexion during the early stance phase.

5. Limitations

The present study has a few limitations. First, the sample size was relatively small. Although our results indicated that adjustment of the PFR could change the muscle force around the ankle, which can maximize the effect of AFO during gait training, further investigations need to recruit a large number of participants to establish clinical credibility. Second, we only used the EMG data for comparing the experimental muscle activity and the estimated muscle activity. Future studies should not only use physical assessment such as spasticity but also an EMG-driven musculoskeletal model and an AFO model for stroke patients because stroke patients may have abnormal gait and muscle activity pattern. It is also important to investigate further by subgrouping stroke patients into responders and non-responders. Finally, we did not estimate the hip and knee muscle forces. Previous studies reported that AFO did not affect the hip kinetics and kinematics during gait training; however, future studies should investigate the effect on joint muscle forces.

6. Conclusion

The present study demonstrated that changing the PFR of an AFO significantly affected the muscle force around the ankle joint. Although the angular positions of the ankle and knee joints significantly affected the PFR3 and PFR4 conditions compared with the no-AFO condition, no significant differences were observed among the PFR conditions. Therefore, adjustment of the PFR of an AFO would be difficult to achieve based only on the measurement of joint angle position during gait analysis. Our findings have important implications for the assessment of gait and adjustment of the PFR of an AFO in the rehabilitation of stroke patients.

In summary, we provide evidence that adjustment of the PFR can affect muscle force. Specifically, as the PFR increased, the TA muscle force decreased and the MGAS muscle force increased. However, the SOL muscle force was affected by the tilt of the shank when the AFO was set to its highest resistance (PFR4). Therefore, individualized adjustment of the PFR could be used to affect the muscle force around the ankle and, indirectly, the selected parameters of gait, such as gait speed.

Conflict of interest

The authors declared that there are no conflicts of interest.

Acknowledgements

This work was supported by the France Bed Medical Home Care Research Subsidy Public Interest Incorporated Foundations (2016).

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