



The influence of solid ankle-foot-orthoses on forward propulsion and dynamic balance in healthy adults during walking



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ABSTRACT

Background: In post-stroke hemiparetic subjects, solid polypropylene ankle-foot-orthoses are commonly prescribed to assist in foot clearance during swing while bracing the ankle during stance. Mobility demands, such as changing walking speed and direction, are accomplished by accelerating or decelerating the body and maintaining dynamic balance. Previous studies have shown that the ankle plantarflexors are primary contributors to these essential biomechanical functions. Thus, with ankle-foot-orthoses limiting ankle motion and plantarflexor output during stance, execution of these walking subtasks may be compromised. This study examined the influence of a solid polypropylene ankle-foot-orthosis on forward propulsion and dynamic balance in healthy adults.

Methods: Kinematic and kinetic data were recorded from 10 healthy adults walking with and without a unilateral ankle-foot-orthosis at steady-state slow (0.6 m/s) and moderate (1.2 m/s) speeds, and during accelerated (0–1.8 m/s at 0.06 m/s²) and decelerated (1.8–0 m/s at –0.06 m/s²) walking. Propulsion was quantified by propulsive and braking impulses (i.e., time integral of the anterior–posterior ground reaction force) while dynamic balance was quantified by the peak-to-peak range of whole-body angular momentum.

Findings: The propulsive impulses decreased in the leg with ankle-foot-orthosis compared to the contralateral leg and no ankle-foot-orthosis condition. Further, the ankle-foot-orthosis resulted in a greater range of angular momentum in both the frontal and sagittal planes, which were correlated with the reduced peak hip abduction and reduced ankle plantarflexor moments, respectively.

Interpretation: Solid ankle-foot-orthoses limit the successful execution of important mobility subtasks in healthy adults and that the prescription of ankle-foot-orthosis should be carefully considered.

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1. Introduction

Ankle-foot-orthoses (AFOs) are frequently prescribed to assist with gait impairments in post-stroke hemiparetic subjects (Tyson and Thornton, 2001). The most commonly prescribed AFO is an L-shaped design made of polypropylene (Cakar et al., 2010; Nair et al., 2010), which holds the ankle in a near neutral position to assist in foot clearance during swing while bracing the ankle during stance. A number of studies have investigated the influence of these types of AFOs on post-stroke hemiparetic gait. Although these studies have found no improvement in step length symmetry when subjects walked with and without an AFO (Abe et al., 2009; Tyson and Thornton, 2001), most

have reported an increase in walking speed and improved toe clearance when using some type of AFO (e.g., Abe et al., 2009; Bregman et al., 2010). However, a previous simulation analysis of healthy subjects showed that in order to generate adequate push-off during normal walking, significant plantarflexor strength was required to deform the AFO (Crabtree and Higginson, 2009). Thus, patients with weak plantarflexors may be hindered in generating adequate propulsion. In addition, recent studies have provided conflicting results as to whether AFOs reduce the risk of falling (Cakar et al., 2010; Guerra Padilla et al., 2011; Wang et al., 2007). Other studies have found no difference in clinical balance scores (Park et al., 2009; Wang et al., 2005) or in static or dynamic weight-bearing tasks when wearing an AFO (Simons et al., 2009). However, no study has used quantitative measures to assess the influence of AFOs on dynamic balance during walking.

Improving mobility (e.g., ability to change walking speed and direction) is a common rehabilitation goal. One aspect of mobility is to effectively accelerate or decelerate the body while maintaining dynamic balance. Previous studies have shown that the ankle plantarflexors are

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primary contributors to regulating forward propulsion in healthy subjects (e.g., Liu et al., 2008, Neptune et al., 2008, Peterson et al., 2011). In hemiparetic walkers, analysis of soleus and gastrocnemius electromyography data indicates that they have reduced amplitude and altered timing of plantarflexor activity compared to healthy subjects (Knutsson and Richards, 1979). In addition, simulation analyses have identified the primary impairment in post-stroke limited community walkers as decreased paretic plantarflexor contributions to forward propulsion (Peterson et al., 2010). Thus, with AFOs restricting ankle movement, the generation of forward propulsion may be further impaired.

Diminished balance control is another post-stroke complication. Greater than 50% of stroke survivors experience falls within one year post-stroke (e.g., Ashburn et al., 2008; Sackley et al., 2008). Studies have shown that whole-body angular momentum is highly regulated during normal walking to maintain dynamic balance (e.g., Herr and Popovic, 2008; Pijnappels et al., 2005). Poor regulation of angular momentum has been associated with a higher range of angular momentum and is indicative of poor dynamic balance (e.g., Pijnappels et al., 2004). This is consistent with research showing that post-stroke hemiparetic subjects with lower clinical balance scores have difficulty regulating their frontal-plane angular momentum (Nott et al., 2014). Previous simulation analyses of healthy walking have shown that the plantarflexors are major contributors to the regulation of whole-body angular momentum (Neptune and McGowan, 2011; Neptune et al., 2011). Thus, AFOs likely influence dynamic balance because they limit ankle motion and plantarflexor output.

Given the widespread use of AFOs in rehabilitation and the important role of the ankle plantarflexors in regulating propulsion and dynamic balance during walking, it is important to understand how AFOs influence the execution of these important biomechanical functions. As a first step, the purpose of this study was to assess the influence of a commonly prescribed solid polypropylene AFO on forward propulsion and dynamic balance in healthy subjects across a range of walking conditions including steady-state, accelerated and decelerated walking. Further, to help interpret any observed differences, changes in the net intersegmental joint moments and powers with and without the AFO were investigated. We hypothesize that when walking with a unilateral AFO: 1) the generation of forward propulsion from the AFO leg will decrease, and 2) the range of whole-body angular momentum will increase. Assessing the influence of AFOs on healthy subjects' walking mechanics will provide a baseline for comparison with hemiparetic subjects, which collectively can provide clinicians with quantitative rationale as to whether AFOs improve paretic leg impairments and overall walking mobility.

2. Methods

Ten healthy subjects (age: 27.3, SD = 2.8 years; mass: 72.6, SD = 10.2 kg; height: 1.75, SD = 0.1 m) walked on an instrumented treadmill in randomized trials of steady-state (0.6 m/s and 1.2 m/s), accelerated (0–1.8 m/s at 0.06 m/s²) and decelerated (1.8–0 m/s at –0.06 m/s²) walking. For each walking speed condition, subjects walked with and without a common clinically prescribed unilateral solid polypropylene AFO (Fig. 1) on a randomly assigned leg while three 30-second trials were collected. The study protocol and consent form were approved by an Institutional Review Board and all participants provided informed, written consent prior to study participation.

A 12-camera optical motion capture system (PhaseSpace Inc., San Leandro, CA, USA) was used to record 3D kinematics at 120 Hz using a modified Helen Hayes marker set. 3D ground reaction force (GRF) data were collected at 2000 Hz using an instrumented treadmill (Bertec Corp., Columbus, OH, USA). The kinematic and GRF data were smoothed using a fourth-order Savitzky–Golay (Savitzky and Golay, 1964) least-square polynomial smoothing filter and were resampled at 100 Hz before performing an inverse dynamics analysis. GRF data were



Fig. 1. Solid polypropylene ankle-foot-orthosis (AFO) used in this study.

normalized by body weight. A 13-segment whole-body model including the head, torso, pelvis, upper arms, lower arms, thighs, shanks and feet was used to determine the mass and inertial properties of the body segments, whole-body center-of-mass (CoM) position and velocity as well as the intersegmental joint moments and powers. The joint moments and powers were normalized by body-mass.

At each time step, whole-body angular momentum (H) about the CoM was calculated as:

$$\vec{H} = \sum_{i=1}^n \left[\left(\vec{r}_i^{COM} - \vec{r}_{body}^{COM} \right) \times m_i \left(\vec{v}_i^{COM} - \vec{v}_{body}^{COM} \right) + I_i \vec{\omega}_i \right]$$

where \vec{r}_i^{COM} and \vec{v}_i^{COM} are the position and velocity vectors of the i -th segment's CoM, respectively, and $\vec{\omega}_i$ is the angular velocity of the i -th segment. \vec{r}_{body}^{COM} and \vec{v}_{body}^{COM} are the position and velocity vectors of the whole-body CoM, m_i and I_i are the mass and moment of inertia of the i -th segment and n is the number of segments. H was normalized by the product of subject mass, height and $\sqrt{g \cdot l}$, where $g = 9.81 \text{ m/s}^2$ and l is the subject height. The term $\sqrt{g \cdot l}$ has units of m/s and is independent of walking speed. During the non-steady-state (i.e., accelerated and decelerated) walking conditions, four representative steps (S1–S4) were selected in each trial and each subject. For each step, the corresponding H was averaged across all trials and all subjects. The ranges of treadmill speed during the selected steps in the accelerated walking condition were approximately S1 = 0.40–0.49 m/s, S2 = 0.62–0.70 m/s, S3 = 1.20–1.26 m/s, and S4 = 1.53–1.58 m/s. The ranges of treadmill speed during the selected steps in the decelerated walking condition were similar to those during the accelerated condition but with a descending speed order. Dynamic balance was assessed using the range of H in each plane, which was calculated as the difference between the maximum and the minimum value of H over the entire gait cycle. Propulsion was assessed using the braking and propulsive GRF impulses, which were calculated using the time integral of the negative and positive anterior–posterior (A/P) GRFs, respectively. During the steady-state walking conditions the impulses were averaged across all steps and subjects. During the non-steady-state walking conditions, the impulses varied over each step due to A/P GRFs and stance time varying with walking speed (e.g.,

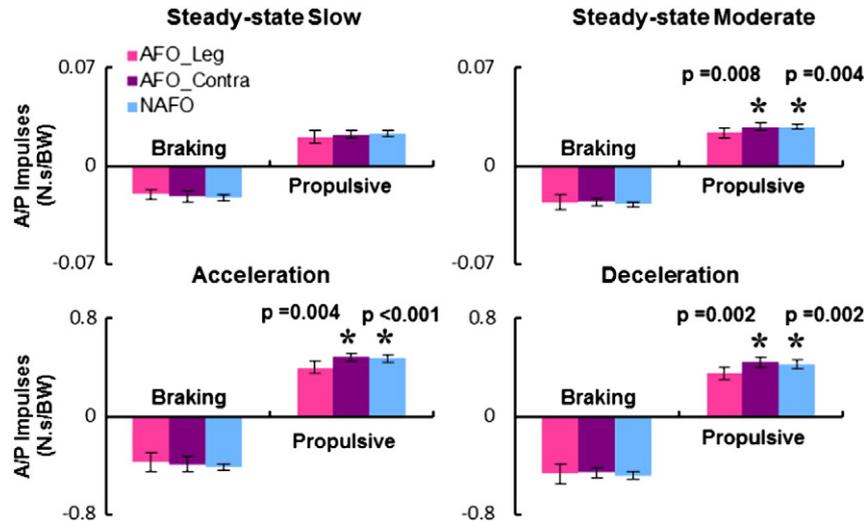


Fig. 2. Mean braking and propulsive impulses (normalized by body weight) for the AFO leg (AFO_Leg), contralateral leg (AFO_Contra) and average of both legs in the no AFO (NAFO) condition. A significant difference between the AFO leg and other conditions (contralateral leg and NAFO) is indicated with *** ($P < \frac{\alpha}{3}$, $\alpha = 0.05$).

Peterson et al., 2011). Therefore, for comparison purposes, the propulsive and braking impulses were each summed over the entire trial before they were averaged across subjects. In addition, due to step symmetry in the no AFO (NAFO) condition, the GRF impulses from both legs were averaged.

The range of *H* was compared between the AFO and NAFO conditions at each walking speed using a paired *t*-test. In the sagittal plane, the range of *H* was analyzed during the first (0–50%) and second (51–100%) halves of the gait cycle. Braking and propulsive impulses were compared using a one-factor (AFO condition) repeated measures ANOVA ($\alpha = 0.05$) with three levels (AFO leg, contralateral leg, and average of both legs during the NAFO condition). Mauchly’s test of sphericity was used to verify equality of variances. If sphericity was violated, a Greenhouse–Geisser adjustment was applied. When significant effects were found, pairwise comparisons with a Bonferroni adjustment were used to determine which values were significantly different. Peak intersegmental joint moments and powers during early (0–50%) and late (51–100%) stance were compared using an ANOVA during steady-state walking speeds to further understand observed differences between the AFO and NAFO conditions. Pearson correlation analyses were performed between the range of *H* and the peak joint moments.

3. Results

3.1. Anterior–posterior GRF impulses

During steady-state slow (0.6 m/s) walking, there were no significant differences in the A/P GRF impulses with and without wearing the AFO (Fig. 2). Similarly, wearing the AFO had no influence on the peak ankle plantarflexor moment (Table 1). However, the peak plantarflexor power generation and absorption were significantly

($P < \alpha/3$, $\alpha = 0.05$) lower in the AFO leg compared to the contralateral leg and the NAFO condition (Table 2 and Fig. 3).

During the steady-state moderate (1.2 m/s), and accelerated and decelerated walking conditions the propulsive impulse generated by the AFO leg was significantly lower than that from the contralateral leg and the NAFO condition (Fig. 2). There were no differences in the braking impulses. In addition, during steady-state moderate walking condition, the peak ankle plantarflexor moment (Table 1) and power (Table 2 and Fig. 3) were lower in the AFO leg although differences in the plantarflexor moment were not significant ($P > \alpha/3$, $\alpha = 0.05$). In addition, the peak hip extensor moment and power were significantly ($P < \alpha/3$, $\alpha = 0.05$) higher in the AFO leg (Tables 1 and 2).

3.2. Frontal-plane angular momentum

During steady-state walking, the range of frontal-plane *H* was significantly higher when wearing the AFO during slow ($P = 0.008$) and moderate ($P = 0.002$) walking speeds (Figs. 4–5). In addition, the first (early stance) and second (late stance) peak hip abduction moments were significantly ($P < 0.01$) lower in the AFO leg compared to the contralateral leg and the NAFO condition (Table 1). During steady-state slow walking, the range of *H* was negatively correlated with the peak hip abduction moment in the AFO leg in early ($r = -0.62$, $P = 0.003$) and late ($r = -0.56$, $P = 0.01$) stance. This correlation was also present during steady-state moderate walking in early ($r = -0.60$, $P = 0.004$) and late ($r = -0.59$, $P = 0.006$) stance. During accelerated walking, the range of *H* was significantly higher with the AFO during steps S2 ($P = 0.039$), S3 ($P = 0.017$) and approached significance in S4 ($P = 0.062$) (Fig. 6). During decelerated walking, the range of *H* was significantly higher with the AFO during steps S3 ($P = 0.046$), S2 ($P = 0.015$) and S1 ($P = 0.004$) (Fig. 6).

Table 1

Mean (SD) peak intersegmental joint moments normalized by body mass for the AFO leg (AFO_Leg), contralateral leg (AFO_Contra) and average of both legs during the no AFO (NAFO) condition. ($P < \frac{\alpha}{3}$, $\alpha = 0.05$).

Peak joint moment (N·m/kg)	Slow steady-state speed			Moderate steady-state speed		
	AFO_Leg	AFO_Contra	NAFO	AFO_Leg	AFO_Contra	NAFO
Hip abduction, 1st peak	0.62 (0.16) ^{ab}	0.91 (0.19)	0.82 (0.07)	0.70 (0.19) ^{ab}	1.02 (0.18) ^a	0.88 (0.11)
Hip abduction, 2nd peak	0.55 (0.13) ^a	0.77 (0.19)	0.75 (0.07)	0.45 (0.14) ^{ab}	0.70 (0.20)	0.71 (0.10)
Hip extension	0.37 (0.12)	0.34 (0.14)	0.31 (0.08)	0.80 (0.14) ^a	0.79 (0.18)	0.70 (0.08)
Hip flexion	0.56 (0.22)	0.51 (0.23)	0.50 (0.14)	0.92 (0.21)	0.97 (0.27)	0.97 (0.18)
Ankle plantarflexion	1.32 (0.19)	1.23 (0.18)	1.32 (0.15)	1.49 (0.19)	1.53 (0.09)	1.59 (0.12)

^a Indicates a significant difference with the NAFO condition.

^b Indicates a significant difference between the AFO leg and the contralateral leg.

Table 2
Mean (SD) peak joint power normalized by body mass for the AFO leg (AFO_Leg), contralateral leg (AFO_Contra) and average of both legs during the no AFO (NAFO) condition. Positive and negative values indicate power generation and absorption, respectively. ($P < \frac{\alpha}{3}$, $\alpha = 0.05$).

Peak joint power (W/kg)	Slow steady-state speed			Moderate steady-state speed		
	AFO_Leg	AFO_Contra	NAFO	AFO_Leg	AFO_Contra	NAFO
Hip adduction (+)	0.04 (0.03)	0.03 (0.02)	0.04 (0.03)	0.11 (0.10)	0.14 (0.13)	0.13 (0.10)
Hip abduction (–)	–0.11 (0.09) ^{ab}	–0.26 (0.17)	–0.21 (0.10)	–0.29 (0.20) ^{ab}	–0.63 (0.22)	–0.54 (0.18)
Hip abduction 1st pk (+)	0.14 (0.06)	0.25 (0.17) ^a	0.16 (0.12)	0.21 (0.10)	0.31 (0.14)	0.27 (0.16)
Hip abduction 2nd pk (+)	0.10 (0.05) ^a	0.19 (0.13)	0.21 (0.09)	0.14 (0.11) ^{ab}	0.40 (0.15)	0.41 (0.14)
Hip extension (+)	0.35 (0.17)	0.31 (0.17)	0.27 (0.12)	0.94 (0.20) ^a	0.85 (0.35)	0.73 (0.18)
Hip flexion (–)	–0.38 (0.21)	–0.31 (0.15)	–0.27 (0.10)	–0.79 (0.27)	–0.80 (0.23)	–0.67 (0.16)
Hip flexion (+)	0.38 (0.16)	0.42 (0.09)	0.36 (0.07)	1.04 (0.24)	1.03 (0.14)	1.03 (0.09)
Ankle plantarflexion (–)	–0.40 (0.10) ^{ab}	–0.51 (0.09)	–0.59 (0.18)	–0.75 (0.32) ^a	–0.85 (0.12) ^a	–1.21 (0.23)
Ankle plantarflexion (+)	0.50 (0.30) ^{ab}	1.10 (0.35)	1.19 (0.07)	1.22 (1.04) ^{ab}	3.41 (0.61)	3.69 (0.51)

^a Indicates a significant difference with the NAFO condition.

^b Indicates a significant difference between the AFO leg and the contralateral leg.

3.3. Transverse-plane angular momentum

There were no significant differences in the range of transverse-plane H during any of the walking conditions.

3.4. Sagittal-plane angular momentum

During steady-state slow walking, there were no significant ($P > 0.05$) differences in the sagittal-plane H (Fig. 4). However, during the moderate walking speed, the range of H in both the first ($P = 0.003$) and second ($P = 0.001$) halves of the gait cycle was significantly higher when wearing the AFO (Fig. 5). Also, the peak ankle plantarflexor moment in the AFO leg was negatively correlated with the range of H in the first ($r = -0.54$, $P = 0.01$) and second ($r = -0.44$, $P = 0.05$) halves of the gait cycle. During accelerated walking, at the fastest speed (step S4), the range of H was significantly higher during the first ($P = 0.041$) and marginally higher during the second ($P = 0.09$) halves of the gait cycle when walking with the AFO (Fig. 6). During decelerated walking, in the first half of the gait cycle, the range of H was significantly higher with the AFO at the faster speeds S4 ($P = 0.004$) and S3 ($P = 0.006$). Also, during the second half of the gait cycle, the range of H was higher with the AFO at the faster speeds S4 ($P = 0.059$) and S3 ($P = 0.024$).

4. Discussion

The purpose of this study was to assess the influence of a commonly prescribed solid polypropylene AFO on forward propulsion and dynamic balance in healthy subjects across a range of walking conditions. During the steady-state slow walking condition, the AFO had no influence on the generation of propulsive and braking impulses and peak ankle plantarflexor moment. However, ankle plantarflexor power generation

and absorption significantly decreased with the AFO (Fig. 3) due to the AFO limiting the ankle range of motion. Therefore, our first hypothesis that forward propulsion decreases with an AFO was not supported during the slow steady-state walking condition. However, during all other walking conditions (steady-state moderate, accelerated and decelerated walking), our first hypothesis was supported in that the propulsive impulses decreased in the AFO leg compared to the contralateral leg and NAFO condition (Fig. 2). In addition, during the moderate walking condition, the peak ankle moment and power also decreased in the AFO leg. These results were consistent with previous modeling work showing that significant plantarflexor strength is required to deform the AFO and generate needed propulsion (Crabtree and Higginson, 2009). Thus, the limited ankle range of motion provided by AFOs appears to diminish plantarflexor output and the generation of forward propulsion.

An interesting finding was that the peak hip extensor moment and power increased during the moderate steady-state walking condition in the AFO leg (Tables 1 and 2). This is consistent with simulation analyses showing that the biarticular hamstrings contribute to forward propulsion in early and mid-stance in healthy walkers (e.g., Neptune et al., 2004; Liu et al., 2006) and the nonparetic leg rectus femoris and biarticular hamstrings contribute to forward propulsion in post-stroke subjects to compensate for the lack of plantarflexor output (Hall et al., 2011). In addition, previous studies have shown that both transistibial and transfemoral amputees also use similar strategies at the hip level to compensate for the loss of plantarflexor output (Prinsen et al., 2011). Thus, it is most likely that the peak hip extensor moment increased in the AFO leg to compensate for the decreased plantarflexor output. In contrast to a previous analysis of accelerated and decelerated walking that showed a higher hip extensor moment was associated with a higher braking impulse (Peterson et al., 2011), we did not observe any changes in the braking impulse. However, the variability in the AFO leg braking impulse increased in all conditions while the propulsive impulse in the AFO leg decreased (Fig. 2). With the generation of forward propulsion being a key factor for modulating walking speed (e.g., Shumway-Cook and Woollocott, 2001), the present results suggest that an AFO may hinder forward propulsion to a greater extent during non-steady-state walking (Fig. 2).

The AFO also influenced dynamic balance control, as wearing the AFO resulted in a higher range of H in both the frontal and sagittal planes, which supported our second hypothesis. Whole-body angular momentum has previously been found to be highly regulated during walking (Herr and Popovic, 2008) and that successful regulation of H is needed in older adults to prevent a fall following a trip (Pijnappels et al., 2005). Thus, poor regulation of H during walking results in higher magnitudes of H , which may lead to a decreased ability to recover dynamic balance following a perturbation.

In the frontal plane, the range of H was higher with the AFO during both steady-state and non-steady-state walking conditions, although

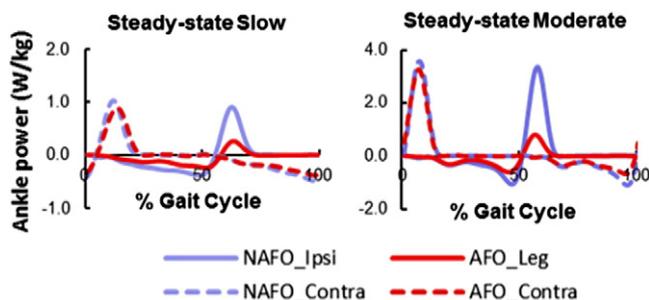


Fig. 3. Mean ankle power (normalized by body mass) for each leg with and without a unilateral AFO during steady-state slow and moderate walking conditions. The AFO condition consists of the AFO leg (AFO_Leg) and contralateral leg (AFO_Contra). The no AFO condition consists of the ipsilateral leg (NAFO_Ipsi) and contralateral leg (NAFO_Contra).

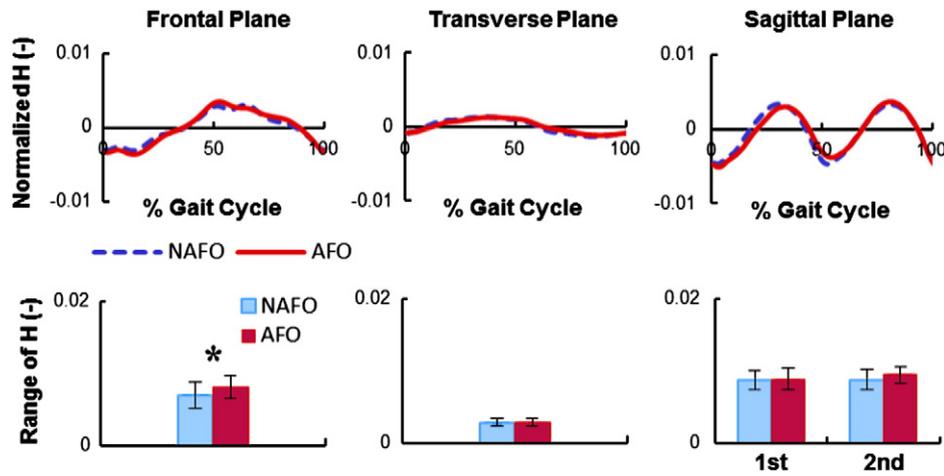


Fig. 4. Normalized, mean 3D whole-body angular momentum (H) during the steady-state slow (0.6 m/s) walking condition with (AFO) and without (NAFO) an AFO. Figures are in the AFO leg reference frame. '1st' and '2nd' indicate the first and second halves of the gait cycle, respectively. The mean (SD) range of H is shown in the bottom row. A significant difference between the AFO and NAFO conditions is indicated with '*' ($P < 0.05$).

differences were not significant at the lowest speed during the acceleration and at the highest speed during the deceleration conditions. During steady-state slow and moderate walking, the increased H corresponded to a lower peak hip abduction moment in the AFO leg during early and late stance. These findings are consistent with a previous study showing a correlation between a higher range of frontal-plane H and a lower peak hip abduction moment (Silverman et al., 2012). They also suggested increasing the hip abduction moment as a mechanism for reducing the range of frontal-plane H . Others have shown that the hip abduction moment is an important contributor to maintaining balance in the frontal plane by acting on the pelvis and moving the HAT (head, arms and trunk) CoM laterally towards the supporting foot and reducing the net external moment about the CoM (MacKinnon and Winter, 1993). In addition, simulation analyses of healthy walkers have shown that during late stance, the plantarflexors contribute to the frontal-plane H by rotating the body towards the contralateral leg, whereas the gluteus medius acts to rotate the body towards the ipsilateral leg (Neptune et al., 2011). The current study found that when wearing an AFO the peak hip abduction moment in the AFO leg decreased, which negatively affected the regulation of frontal-plane angular momentum. During the steady-state moderate walking condition, it is possible that the peak hip abduction moment decreased in the AFO leg due to a decreased ankle plantarflexion

moment (Table 1). However, during steady-state slow walking, the peak hip abduction moment still decreased in the AFO leg even though the peak plantarflexor moment was not affected by the AFO. Thus, the underlying reason for a decreased peak hip abduction moment in the AFO leg is not clear.

In the sagittal plane, the range of H was higher with an AFO during steady-state moderate walking (Fig. 5) and at the higher speeds during accelerated and decelerated walking (Fig. 6). Further, the peak ankle plantarflexor moment was lower in the AFO leg during steady-state moderate walking and negatively correlated with the range of H in both the first and second halves of the gait cycle (i.e., a lower ankle plantarflexor moment was associated with a higher range of H). Previous simulation analyses of healthy walkers have identified the ankle plantarflexors as the primary contributors to the regulation of H in the sagittal plane (Neptune and McGowan, 2011). Thus, AFOs restricting the ankle range of motion and hindering plantarflexor muscle force generation not only limit propulsion generation but also lead to a poor dynamic balance control. This has important implications for prescribing AFOs to post-stroke subjects as the influence of AFOs on propulsion and dynamic balance in those with post-stroke hemiparesis with different levels of plantarflexor impairment is unclear. Previous simulation analyses of unilateral below-knee amputees have shown that a foot-ankle prosthesis, which is functionally similar to an AFO, performs

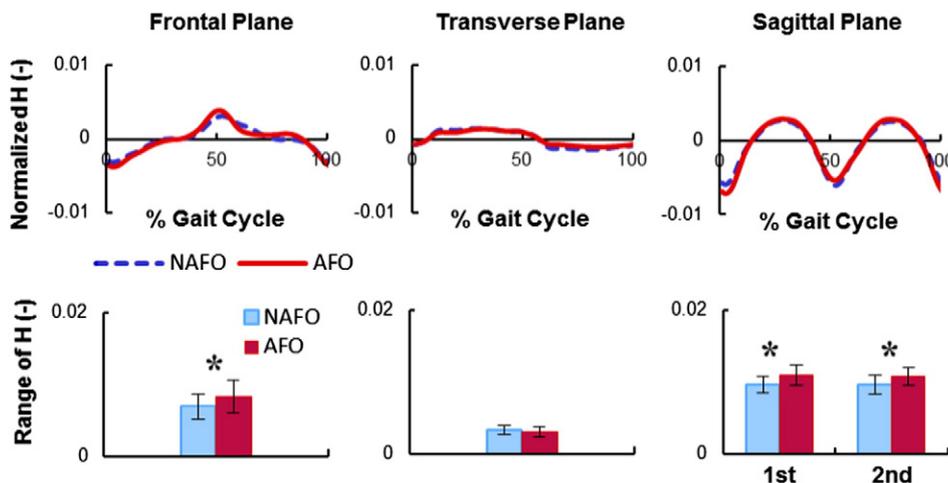


Fig. 5. Normalized, mean 3D whole-body angular momentum (H) during the steady-state moderate (1.2 m/s) walking condition with (AFO) and without (NAFO) an AFO. Figures are in the AFO leg reference frame. '1st' and '2nd' indicate the first and second halves of the gait cycle, respectively. The mean (SD) range of H is shown in the bottom row. A significant difference between the AFO and NAFO conditions is indicated with '*' ($P < 0.05$).

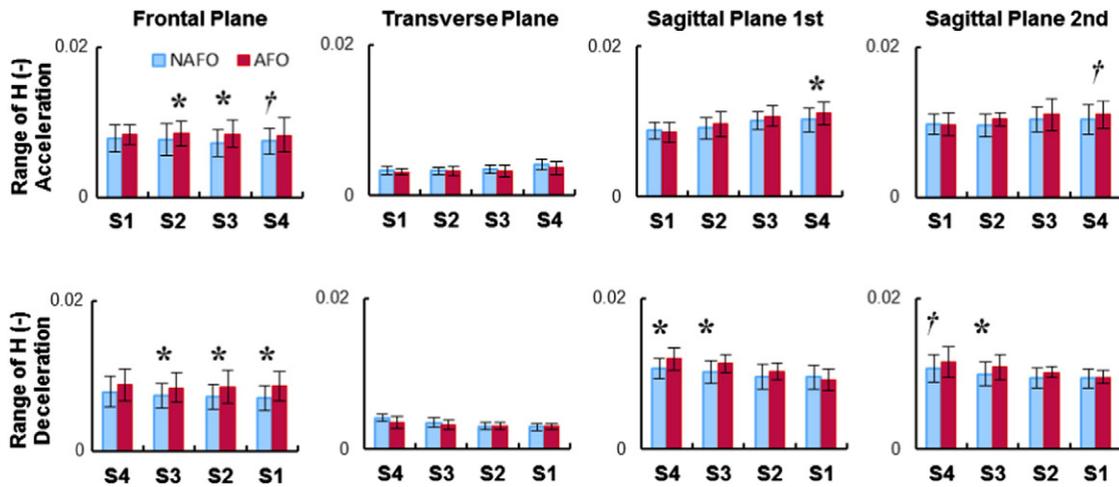


Fig. 6. Normalized, mean (SD) range of whole-body angular momentum (H) in 3D with (AFO) and without (NAFO) an AFO. Top row shows the accelerated and bottom row shows the decelerated walking condition. '1st' and '2nd' indicate the first and second halves of the gait cycle, respectively. A significant difference between the AFO and NAFO conditions is indicated with ** ($P < 0.05$). Marginal differences are shown with \dagger ($0.05 < P < 0.1$). Treadmill speed values at the selected steps are S1 = 0.40–0.49 m/s, S2 = 0.62–0.70 m/s, S3 = 1.20–1.26 m/s, and S4 = 1.53–1.58 m/s.

similarly as the uniaxial soleus muscle to provide forward propulsion (Silverman and Neptune, 2012). Thus, it is likely that those hemiparetic subjects with minimal plantarflexor output would benefit from the additional ankle stiffness provided by an AFO to generate needed stability and perhaps some levels of propulsion (e.g., the elastic energy stored in the AFO due to deformation is released and contributes to the net joint moment at the ankle). Future studies are needed to identify biomechanical markers that distinguish which subjects would benefit from an AFO from those that would not. Similarly, in cases where an AFO is needed to compensate for gait impairments such as foot drop, perhaps more advanced AFO designs that promote ankle plantarflexion (e.g., Bartonek et al., 2007; Desloovere et al., 2006) or functional electrical stimulation systems (e.g., The WalkAide System, Austin, TX, USA) that promote dorsiflexion can improve rehabilitation outcomes.

Conclusions

In summary, common clinically prescribed solid polypropylene AFOs adversely influenced the generation of forward propulsion and dynamic balance in healthy adults during steady-state and non-steady-state walking. The ankle plantarflexors normally play an important role in the generation of propulsive impulses and regulation of whole-body angular momentum. However, the AFO restricted the ankle range of motion and was found to hinder the generation of propulsion and the regulation of angular momentum in both the frontal and sagittal planes. Thus, during mobility tasks when these important functions are needed, such as when changing walking speed and direction, AFOs limit their successful execution and suggest that the prescription of AFOs should be carefully considered.

Conflict of interest

There is no conflict of interest regarding the publication of this manuscript.

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