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Preservation of the first rocker is related to increases in gait speed in individuals with hemiplegia and AFO

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ABSTRACT

Background: Changes in impulse during the first rocker (braking force) and third rocker (propulsion force) may affect changes in gait speed after orthotic intervention. The purpose of this investigation was to objectively measure changes in impulse during double support and correlate those findings to changes in gait speed with and without ankle foot orthosis in individuals with hemiplegia.

Methods: Fifteen adults with stroke-related hemiplegia walked with and without ankle foot orthosis while foot pressure data was collected bilaterally. Outcome measures included: gait cycle time (s), mean force (N), and impulse (Ns) in the wholefoot, hindfoot, forefoot, and toe box during initial double support and terminal double support.

Findings: Time significantly decreased during the entire gait cycle, initial double support, and terminal double support, with the ankle foot orthosis. During initial double support, affected limb impulse significantly decreased with the ankle foot orthosis in the wholefoot ($P=0.016$), and hindfoot ($P=0.006$), and hindfoot impulse % change and gait speed % change were significantly correlated ($P=0.007$). During terminal double support, affected limb impulse was not significantly different in the wholefoot or forefoot and these changes were not significantly correlated to gait speed.

Interpretation: Previous research found that orthotics increase gait speed in individuals with hemiplegia. This research suggests that the increase in speed is not due to increased propulsive forces at the end of terminal double support, but due to decreased braking forces during initial double support. Therefore, the orthosis preserved the first ankle rocker and provided a more efficient weight acceptance which positively affected gait speed.

1. Introduction

Recovery of walking ability is a key functional objective for patients and clinicians in stroke rehabilitation and gait speed is a commonly used outcome measure (Bohannon et al., 1988; Schmid et al., 2007). Ankle foot orthotics (AFOs) are often prescribed to individuals with hemiplegia to assist with ambulation (Jutai et al., 2007). Gait speed has been used as a primary indicator of orthotic effectiveness, and improved functional ambulation (Tilson et al., 2010). Several studies have provided evidence to support the predictive validity of gait speed and more recently gait speed has been shown to be positively correlated with level of disability, function, and quality of life in individuals with stroke (Goldie et al., 1996; Schmid et al., 2007). Previous research has evaluated the compensatory benefits of using an AFO on temporal and spatial patterns of walking but has failed to correlate these changes to gait mechanisms during weight transfer (Nair et al., 2010). Although gait speed reflects overall gait performance, it is limited in identifying the mechanisms behind orthoses mediated changes in gait and overall post-stroke functional recovery (Olney et al., 1994).

Research has suggested that the greatest effects of impairment during hemiplegic gait occur during double support, when weight is transferred from one limb to another (Giuliani, 1990). Motor deficits in ankle-foot function resulting from hemiplegia can delay weight transfer onto the affected limb causing an increase in double support time which can decrease walking speed (Nolan and Yarossi, 2011; Perry and Burnfield, 2010). In healthy individuals, weight transfer during initial double support (IDS) is assisted by the heel rocker (first rocker) and during terminal double support (TDS) by the forefoot rocker (third rocker). These gait mechanisms preserve progression during weight transfer and weight acceptance throughout double support (Perry and Burnfield, 2010). After orthotic intervention changes in impulse during the first rocker (braking force) and third rocker (propulsion force) may directly affect changes in gait speed (Bowden et al., 2006).

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Disruptions in weight transfer have been previously described as anterior–posterior perturbations of the center of pressure during the initial double support in subjects with hemiplegia. Use of an AFO has been shown to reduce this perturbation of the center of pressure (Fatone and Hansen, 2007). Existing literature evaluating the center of pressure indicates that applying an AFO in individuals with stroke can improve the gait pattern at the expense of ankle range of motion and power generation during push-off (Fatone and Hansen, 2007; Perry and Burnfield, 2010).

The forces associated with plantar loading during gait after orthotic intervention have been measured in previous research utilizing force platforms (Nair et al., 2010). Although force platforms can provide valuable information regarding the ground reaction force and the center of pressure, they provide little information on how the plantar surface of the foot is loaded with respect to the supporting surface (Orlin and McPoil, 2000). Force plates do not provide regional analysis of loading in specific planter areas such as the heel or forefoot and therefore information about the forces associated with specific plantar regions cannot be quantified. The use of force plates is also limited by the number of gait cycles that can be collected. The use of pedobarography allows the analysis of multiple gait cycles and comprehensive information on how the foot is loaded.

Limited research has been conducted to understand the specific mechanisms leading to improved gait speed after orthotic intervention. The purpose of this investigation was to objectively measure changes in impulse during double support and correlate those findings to changes in gait speed with and without AFO in individuals with hemiplegia.

2. Methods

2.1. Participants

Individuals with hemiplegia secondary to stroke with symptoms lasting more than 6 months were recruited for participation. All participants were previously prescribed an AFO for functional ambulation by their treating physician. Other inclusion criteria for the stroke group included: 1) uninvolved lower limb had no history of injury or pathology; 2) must be able to walk independently or with supervision for 25 ft, both with and without AFO; and 3) must wear an AFO when walking at least 50% of the time. Individuals with significant orthopedic, neuromuscular, or neurological pathologies or history that would interfere with walking or limit the range of motion of the legs were excluded from the study.

2.2. Procedures

Gait speed was measured using the average velocity during a 2 Minute Walk Test (2MWT), a performance based functional ambulation test. Participants completed one 2MWT in each condition, with and without AFO, with the order being randomly assigned. All participants received standardized instruction to walk as far as possible, safely, at their self-selected comfortable pace for 2 min. Participants were allowed to stop and rest if necessary but time kept running. The distance walked over 2 min was recorded to the nearest inch using a measuring wheel. Total distance walked over 2 min was converted to the metric system and used to calculate gait speed.

Wireless pedobarography data was collected bilaterally using the pedar®-x (Novel GmbH, Munich, Germany). The pedar®-x is an objective, quantifiable pressure distribution measuring system for monitoring magnitude and timing of plantar loading (Ramanathan et al., 2010). The system consists of a portable data collection device attached to the subject's hip, and two thin elastic sensor insoles that are inserted directly below the plantar surface of the foot. The insole sensor technology allows for bilateral analysis of multiple steps without the need to target foot placements during gait as is needed with the use of force platforms. Using the pedar®-x, force is calculated by multiplying the recorded pressure by the sensor area resulting in a force "normal" to each sensor in the matrix (Kernozek et al., 1996). Calculation of force when using this technique may underrate the force as compared to the forces recorded from a force plate. The accuracy of the forces using the pedar®-x has previously been determined to be reliable for research and clinical applications (Barnett et al., 2001; Kernozek et al., 1996; Ramanathan et al., 2010).

Data for all participants was collected at 100 Hz sampling rate during 10 walking trials at a self-selected pace on an 8 m walkway (5 with AFO and 5 without AFO). Participants wore neutral walking shoes (New Balance Athletic Shoe, Boston, MA, USA) style NB 575VW, last SL2, for all walking trials and no comparisons were made to a barefoot condition. Shoes were standardized to minimize shear friction and plantar pressure attenuation due to low pass filtering effects. Members of the study team provided supervision and non-contact guarding during all walking trials for safety. All procedures performed in this investigation were approved by the Human Subject Review Board and informed consent was obtained prior to study participation.

2.3. Data analysis

Demographic information including age, gender, time since stroke, location and type of stroke was collected and verified with medical records. Data from all assessments are represented as mean (standard deviation).

Pedobarography data was divided into nine plantar regions (masks) bilaterally using Multimask Evaluation (Novel Electronics, Inc., Munich, Germany). All data were exported as masked time process files and imported into Matlab (The Mathworks, Inc., Natick, MA, USA) for custom analysis. Masks were then combined to form anatomical regions including the wholefoot, heel, hindfoot (heel and arch region), toe box (toes and hallux) and forefoot (metatarsal heads and toe box), Fig. 1. Per convention the first two steps were omitted from analysis for all walking tests.

Time series force data was used to determine gait cycle events. Footstrike was defined as the time point at which any sensor measured a force that was greater than zero and toe off was defined as the first time point subsequent to footstrike when all forces plantar to the foot were equal to zero. A gait cycle was defined unilaterally as the time period between footstrikes on the same leg. The stance phase designates the entire period during which the foot is on the ground. The stance phase was subdivided into 3 subphases: initial double support (IDS); single support (SS); and terminal double support (TDS). IDS is the first period following footstrike where both feet are in contact with the ground. Single limb support begins when the opposite foot is lifted off the ground and ends at contact of the opposite foot with the ground. TDS begins with contact of the opposite limb with the ground and ends when the original stance limb is lifted for swing (Perry and Burnfield, 2010). Impulse was calculated in custom Matlab programs by integrating the local forces under specific anatomical regions during IDS and TDS using a trapezoidal integration method with a time interval of 10 ms.

2.4. Outcome measures

Primary outcome measures included gait speed (m/s), temporal characteristics (seconds) of the gait cycle phases (gait cycle, stance, IDS, SS, and TDS) as well as mean force (bodyweights), and impulse (bodyweight · seconds) in the heel, hindfoot, toe box and forefoot during IDS and TDS. The hindfoot impulse during IDS was used to quantify the total braking force associated with weight acceptance subsequent to initial foot contact. Likewise, the forefoot impulse during TDS was used to quantify total propulsive force associated with weight transference in preparation for swing. Changes in hindfoot

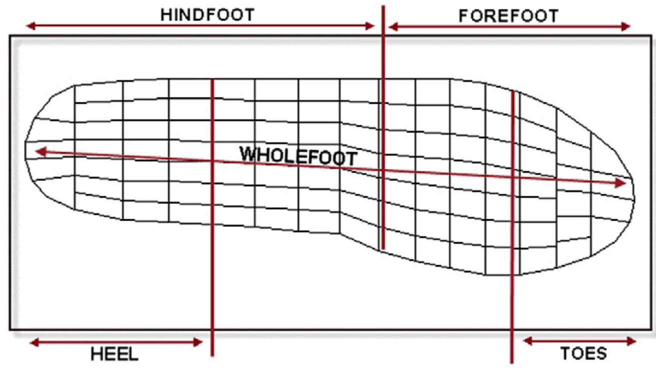

Fig. 1. Anatomical plantar regions for data analysis. A relative mask was applied to all data using Multimask Evaluation (Novel Electronics, Inc., Munich, Germany) throughout the gait cycle to create anatomical regions representing the wholefoot, hindfoot, heel, forefoot, and toes.

impulse during IDS and forefoot impulse during TDS were selected to compare to changes in functional measures of gait speed.

2.5. Statistical analysis

Demographic data were analyzed using descriptive statistics in PASW Statistics 18.0 (SPSS Inc., Chicago, Illinois, USA). Paired sample t-tests were used to determine if there were significant differences in gait cycle timing, mean force and impulse in individuals with stroke between conditions, with and without AFO ($P \leq 0.05$). Comparisons of mean force and impulse were made bilaterally during IDS and TDS for the affected and unaffected limbs to examine the intralimb (within a limb) differences during walking trials with and without AFO in individuals with hemiplegia secondary to stroke. Percent change in outcome and functional ambulation variables were calculated as:

$$\frac{With\ \ AFO - Without\ \ AFO}{Without\ \ AFO} \times 100.$$

Pearson correlation coefficients were calculated to relate changes in impulse and gait speed with and without AFO.

3. Results

Twenty participants with stroke agreed to participate in the study. Pedobarography data was not successfully collected for three participants in the without AFO condition because participants could not complete the walking trials without AFO and data from two participants, one with a hinged and another with a carbon fiber AFO were excluded from analysis due to any potential biomechanical advantage in brace design. Data from fifteen participants 45.9(35.2) months post stroke (11 male and 4 female) were available for analysis (age 51.6(12.5) years; height 172.6(12.2) cm; weight 85.8(21.7) kg). Nine participants were affected on the right side and six were affected on the left. All participants used their own custom molded rigid plastic AFO on their affected limb during ambulation in the with AFO condition. All AFOs were positioned below the knee just beneath the neck of the fibula and secured across the anterior aspect of the proximal tibia with a hook and loop strap. Distally, the AFO was trimmed posterior to the ankle malleoli and included a footplate that extended along the length of the plantar surface of the foot to at least the metatarsal heads. The AFO was used for ambulation inside the shoe and shoe type (style and last) was standardized.

Gait speed was evaluated using paired sample t-tests during the 2MWT. With the AFO, there was a significant increase in gait speed, t(14) = 4.22, $P = 0.0008$. Gait cycle timing on the affected limb significantly decreased with the AFO during the entire gait cycle t(14) = −2.35, $P = 0.034$, IDS t(14) = −2.42, $P = 0.030$, and TDS t(14) = −2.172, $P = 0.048$, Table 1. Time spent in SS on the affected limb significantly increased with the AFO t(14) = 2.191, $P = 0.046$. Time spent in stance on the affected limb decreased but the change was not significant t(14) = 1.873, $P = 0.082$. Gait cycle timing on the unaffected limb significantly decreased when the AFO was added to the paretic limb during the entire gait cycle t(14) = −2.317, $P = 0.036$, IDS t(14) = −2.200, $P = 0.045$, stance t(14) = −2.798, $P = 0.014$, and TDS t(14) = −2.435, $P = 0.029$. Time spent in SS on the unaffected limb decreased with the AFO but was not significant t(14) = −1.994, $P = 0.066$.

Table 1
Temporal outcome variables.

	Affected limb		Healthy limb	
	With AFO	Without AFO	With AFO	Without AFO
Gait cycle (s)	1.41 (0.24)[a]	1.55 (0.27)[a]	1.41 (0.23)[b]	1.54 (0.27)[b]
Stance (s)	0.81 (0.15)	0.89 (0.23)	1.00 (0.18)[b]	1.16 (0.28)[b]
IDS (s)	0.18 (0.04)[a]	0.24 (0.12)[a]	0.22 (0.05)[b]	0.28 (0.13)[b]
SS (s)	0.41 (0.08)[a]	0.38 (0.05)[a]	0.60 (0.12)	0.64 (0.09)
TDS (s)	0.22 (0.05)[a]	0.28 (0.13)[a]	0.18 (0.04)[b]	0.24 (0.12)[b]

[a] $P \leq 0.05$ with vs. without AFO for the affected limb.
[b] $P \leq 0.05$ with vs. without AFO for the unaffected limb.

Mean force during IDS on the affected limb significantly decreased in the wholefoot t(14) = −3.359, $P = 0.0029$, hindfoot t(14) = −2.42, $P = 0.0069$, and heel t(14) = −4.297, $P = 0.0007$, Table 2. During IDS, impulse on the affected limb also significantly decreased in the wholefoot t(14) = −2.741, $P = 0.016$, hindfoot t(14) = −3.247, $P = 0.006$, and heel t(14) = −4.094, $P = 0.001$. During IDS, hindfoot impulse % change and gait speed % change demonstrated a significant positive correlation r(14) = 0.66, $P = 0.008$, Fig. 3. On the unaffected limb there were no significant changes in mean force or impulse during IDS in any of the masked anatomical regions or the wholefoot.

Mean force during TDS on the affected limb significantly increased in the wholefoot t(14) = 4.929, $P = 0.0002$ and remained relatively unchanged in the forefoot t(14) = 1.123, $P = 0.28$ and toe box t(14) 1.042, $P = 0.31$, Table 3. During TDS, impulse on the affected limb was not significantly different in the wholefoot t(14) = −0.955, $P = 0.32$, forefoot t(14) = −0.928, $P = 0.37$ or toe box t(14) = −0.807, $P = 0.43$. During TDS, forefoot impulse % change and gait speed % change were not significantly correlated, r(14) = 0.17, $P = 0.535$, Fig. 4. On the unaffected

Table 2
Initial double support (IDS) outcome variables.

	Affected limb		Healthy limb	
	With AFO	Without AFO	With AFO	Without AFO
Whole foot impulse (N/s)	0.05 (0.02)[a]	0.09 (0.06)[a]	0.10 (0.03)	0.13 (0.08)
Hindfoot impulse (N/s)	0.04 (0.01)[a]	0.06 (0.03)[a]	0.09 (0.02)	0.11 (0.06)
Heel impulse (N/s)	0.02 (0.01)[a]	0.04 (0.02)[a]	0.07 (0.02)	0.08 (0.04)
Whole foot mean force (N)	0.311 (0.062)[a]	0.365 (0.080)[a]	0.465 (0.113)	0.471 (0.110)
Hindfoot mean force (N)	0.216 (0.046)[a]	0.271 (0.072)[a]	0.420 (0.106)	0.416 (0.101)
Heel mean force (N)	0.123 (0.045)[a]	0.201 (0.071)[a]	0.329 (0.101)	0.313 (0.112)

[a] $P \leq 0.05$ with vs. without AFO for the affected limb.

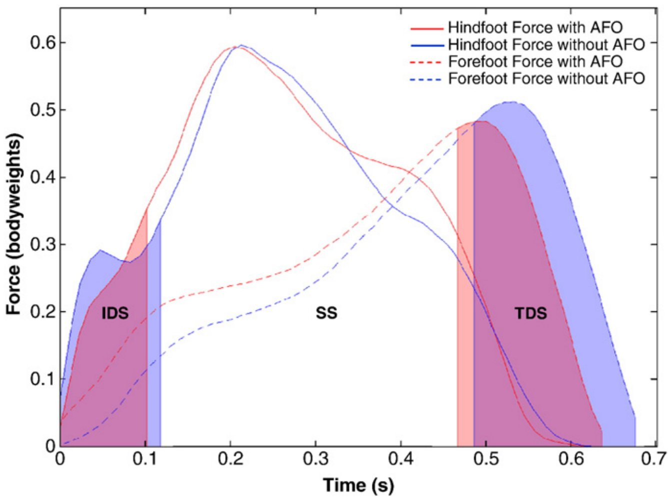

Fig. 2. Affected limb hindfoot and forefoot force during stance provided for one representative participant with and without ankle foot orthosis: IDS — initial double support; SS — single support; and TDS — terminal double support.

limb there were no significant changes in mean force during TDS in any of the masked anatomical regions or the wholefoot. During TDS, impulse on the unaffected limb significantly decreased in the wholefoot $t(14) = 2.306$, $P = 0.037$, and remained relatively unchanged in the forefoot $t(14) = -2.088$, $P = 0.055$ and the toe box $t(14) = -2.010$, $P = 0.64$.

Table 3
Terminal double support (TDS) outcome variables.

	Affected limb		Healthy limb	
	With AFO	Without AFO	With AFO	Without AFO
Whole foot impulse (N/s)	0.10 (0.02)	0.11 (0.04)	0.08 (0.02)[b]	0.11 (0.05)[b]
Hindfoot impulse (N/s)	0.06 (0.02)	0.07 (0.03)	0.07 (0.02)	0.09 (0.03)
Heel impulse (N/s)	0.02 (0.01)	0.02 (0.02)	0.03 (0.01)	0.03 (0.01)
Whole foot mean force (N)	0.466 (0.079)[a]	0.412 (0.062)[a]	0.478 (0.097)	0.481 (0.071)
Hindfoot mean force (N)	0.296 (0.114)	0.277 (0.099)	0.418 (0.109)	0.409 (0.096)
Heel mean force (N)	0.102 (0.060)	0.095 (0.060)	0.171 (0.073)	0.166 (0.072)

[a] $P \leq 0.05$ with vs. without AFO for the affected limb.
[b] $P \leq 0.05$ with vs. without AFO for the unaffected limb.

4. Discussion

4.1. Walking speed

Gait speed increased with the AFO during a functional ambulation test. Previous research has shown that gait speed increases when the AFO is added to the paretic limb in individuals with hemiplegia (Nolan et al., 2009; Tyson and Rogerson, 2009). The average gait speed attained throughout all walking trials was 0.67(0.22) m/s with AFO and 0.57(0.24) m/s without AFO. Healthy self-selected walking speed is 1.48 m/s for men and 1.23 m/s for women (Blessey et al., 1976). According to Perry et al. (1995) individuals with a gait speed between 0.4 and 0.8 m/s would have limited community ambulation equivalent to moderate gait impairments. Although the addition of the AFO increased walking speed not all individuals improved to the level of full community ambulation, gait speed > 0.8 m/s (Perry et al., 1995).

4.2. Temporal changes

Temporal outcome measures revealed a bilateral decrease in gait cycle time when the AFO was added to the paretic limb. Stance time remained constant as double support decreased on the affected limb leading to an increase in SS on the affected side with AFO. Decreased time in IDS and TDS during gait has been previously shown to increase walking speed and is an indicator of increased stability (Khodadadeh et al., 1987; Roth et al., 1997).

Individuals with hemipelgia spent more time on the unaffected limb regardless of condition. When the AFO was added participants increased the time spent in SS on the affected limb but it was still less than the time spent on the unaffected limb. This is likely due to the time required to advance the affected limb during the swing phase and a decrease in forefoot IDS impulse (propulsive power) at toe off on the affected side regardless of AFO utilization (Goldie et al., 1996). The ability to maintain gait speed is dependent on the ability to shift body weight to the affected limb. Stroke rehabilitation goals focus on improving load transfer onto the affected limb to facilitate more healthy movement patterns (Goldie et al., 1996; Sackley, 1990).

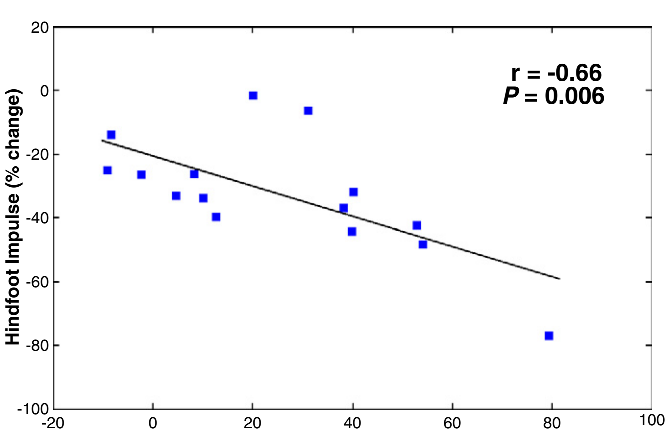

Fig. 3. Correlation of gait speed and hindfoot impulse during IDS.

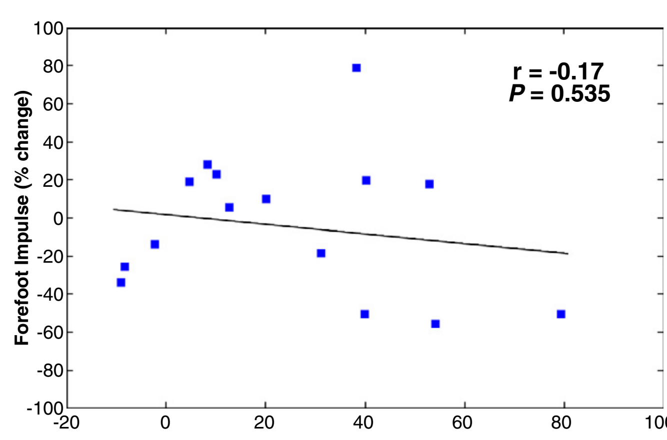

Fig. 4. Correlation of gait speed and forefoot impulse during TDS.

Although gait speed increased with the AFO, gait cycle time still revealed an asymmetrical gait pattern regardless of condition, with or without AFO. This compensatory gait deviation may cause greater difficulty maintaining balance during ambulation.

4.3. Impulse during double support

This investigation demonstrated the use of hindfoot impulse during IDS to quantify the total braking force associated with weight acceptance subsequent to initial foot contact in order to examine the effect of an AFO on preservation of the first rocker.

With the AFO a decrease in affected limb hindfoot impulse during IDS resulted from significant decreases in both the time spent in IDS and the hindfoot mean force.

In healthy subjects, the initial rapid force generated by the falling limb at foot strike is counteracted by eccentric plantarflexion of the pretibial muscles to control foot drop causing a small perturbation in loading force (Perry and Burnfield, 2010). In individuals with hemiplegia greater ankle instability on the affected limb causes an increase in the magnitude and duration of force perturbation following foot strike which disrupts forward progression during the first rocker, Fig. 2. During IDS, properly designed AFOs should assist eccentric contraction of the dorsiflexors during the loading response of the paretic limb. This resists the abrupt plantarflexion of the ankle joint from foot contact to foot flat, making the heel rocker function possible (Perry and Burnfield, 2010). The heel rocker redirects a portion of the downward force during loading response to the pretibial muscles as ankle motion is restrained (Perry and Burnfield, 2010). The dorsiflexion support and ankle stability afforded by the AFO compensated for weak or paralyzed pretibial muscles during initial contact reducing the downward force and smoothing force perturbations during the loading response. This provided a smoother weight transfer which translates kinetically to a decrease in braking force, evidenced by a significant decrease in hindfoot impulse during IDS (Desloovere et al., 2006; Perry and Burnfield, 2010). The reduction of the force perturbation with the addition of the AFO found in this study may be related to the reduction in the anterior–posterior center of pressure perturbation found in previous research (Fatone and Hansen, 2007).

Previous research involving AFOs and healthy subjects found a reduction of the braking force at initial contact accompanied by an increase in double support time, concluding the decrease in braking force resisted forward progression (Nair et al., 2010). The current investigation found a similar decrease in the braking force on the affected limb but the decrease in force was accompanied with a decrease in IDS and TDS time. The decrease in hindfoot impulse (braking force) on the affected limb with the AFO was correlated with an increase in gait speed indicating a more efficient forward progression leading to preservation of the first rocker (Desloovere et al., 2006; Perry and Burnfield, 2010).

Existing literature indicates that improvements in gait speed through the addition of an AFO to the affected limb come at the expense of reducing ankle range of motion and power generation at push-off impairing the third rocker (Fatone and Hansen, 2007; Nair et al., 2010). The current investigation agrees with previous findings that walking with an AFO in individuals with hemiplegia impairs the third rocker (Desloovere et al., 2006). There was no change in impulse during TDS in the forefoot or toe box on the affected limb. This indicated no change in propulsion during TDS when the AFO was added to the paretic limb, however, changes in the propulsive forces were variable. Eight subjects experienced no change or a slight increase in propulsive force and the seven remaining participants experienced a decrease in propulsive force when the AFO was added. As a result, changes in forefoot impulse during TDS did not correlate to changes in walking speed.

A potential limitation of the current study is that the pedobarography technology utilized in this investigation measured the normal forces plantar to the foot and not directional forces such as those measured by a force plate. Although this may limit the comparison of these results to force platform data, the ability to measure loading under specific plantar regions over multiple uninterrupted gait cycles is essential in AFO research. Future research synchronizing pedobarography and force platform data will provide a more comprehensive representation of orthotic intervention during hemiplegic gait.

An additional possible limitation is that gait speed was measured during functional ambulation tests and was not collected simultaneously with pedobarography data. Clinical measures of functional ambulation have previously been related to community ambulation (Tilson et al., 2010).

Another potential limitation of this study is that only traditional plastic AFOs were used in this investigation. It is possible that different types of AFOs will provide various biomechanical advantages and disadvantages and additional types of AFOs should be evaluated in future research. In addition, the AFO–footwear combination was not tuned to optimize gait prior to data collection. The researchers did not attempt to adjust the AFO and completed all testing procedures with the device as prescribed and fitted by the participant's physician and orthotist.

5. Conclusions

It is generally accepted clinically that AFOs improve functional ambulation in individuals with hemiplegia. The current research has shown that orthotics increase gait speed but the increase in gait speed may not be functional for community ambulation. The AFO provides increased dorsiflexion at footstrike creating a decreased impulse (braking force) in the hindfoot during IDS, thereby preserving the first ankle rocker and positively affected gait speed. No effect on propulsive forces during TDS was found with the addition of the AFO. Therefore, the results of this investigation show no effect of the AFO on the third rocker. Generation of plantarflexor propulsion during TDS is vital for forward body progression (Nair et al.) and producing gait speeds functional for community ambulation, which may only be accomplished with a brace capable of preserving the third rocker. Future research is required to fully understand the mechanisms underlying the increases in gait speed associated with orthotic intervention in adults with hemiplegia.

Disclaimers

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Declaration of interest

The authors report no conflicts of interest. The authors alone are responsible for the content and writing of the paper.

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