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Short communication

Articulated ankle-foot-orthosis improves inter-limb propulsion symmetry during walking adaptability task post-stroke

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ABSTRACT

Background: Community ambulation involves complex walking adaptability tasks such as stepping over obstacles or taking long steps, which require adequate propulsion generation by the trailing leg. Individuals post-stroke often have an increased reliance on their trailing nonparetic leg and favor leading with their paretic leg, which can limit mobility. Ankle-foot-orthoses are prescribed to address common deficits post-stroke such as foot drop and ankle instability. However, it is not clear if walking with an ankle-foot-orthosis improves inter-limb propulsion symmetry during adaptability tasks. This study sought to examine this hypothesis.
Methods: Individuals post-stroke ($n = 9$) that were previously prescribed a custom fabricated plantarflexion-stop articulated ankle-foot-orthosis participated. Participants performed steady-state walking and adaptability tasks overground with and without their orthosis. The adaptability tasks included obstacle crossing and long-step tasks, leading with both their paretic and nonparetic leg. Inter-limb propulsion symmetry was calculated using trailing limb ground-reaction-forces.
Findings: During the obstacle crossing task, ankle-foot-orthosis use resulted in a significant improvement in inter-limb propulsion symmetry. The orthosis also improved ankle dorsiflexion during stance, reduced knee hyper-extension, increased gastrocnemius muscle activity, and increased peak paretic leg ankle plantarflexor moment. In contrast, there were no differences in propulsion symmetry during steady-state walking and taking a long-step when using the orthosis.
Interpretation: Plantarflexion-stop articulated ankle-foot-orthoses can improve propulsion symmetry during obstacle crossing tasks in individuals post-stroke, promoting paretic leg use and reduced reliance on the non-paretic leg.

1. Introduction

Community ambulation, an important rehabilitation goal post-stroke, is a major challenge partly due to poor walking adaptability. Walking adaptability, the ability to modify walking to meet environmental demands (Balasubramanian et al., 2014), requires adequate generation of forward propulsion from both legs. Walking adaptability tasks include obstacle crossing and taking long-steps, some of which require greater forward propulsion generation compared to steady-state walking (Clark et al., 2016). Propulsion symmetry post-stroke (Roelker et al., 2019) is strongly associated with hemiparetic severity (Bowden et al., 2006) and even targeted as a main outcome of therapy (Hsiao et al., 2016). Propulsion symmetry is characterized by reduced propulsion by the paretic leg and increased propulsion by the nonparetic leg (Bowden et al., 2006). The increased reliance on the nonparetic leg can further challenge performance of complex walking adaptability tasks that require high output from both legs (Vistamehr et al., 2017). Ankle-foot-orthoses (AFOs) are prescribed as a common intervention to assist with foot clearance during swing and ankle stability during stance. However, systematic review articles reveal that prior studies of AFOs are limited to assessing either clinical outcomes or biomechanical measures of gait during steady-state walking (Choo and Chang, 2021; Daryabor et al., 2018; Shahabi et al., 2020; Tyson et al., 2013). van Swigchem et al. (2014) reported reduced gait adaptability in post-stroke AFO users

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that were classified as community ambulators. This finding was evidenced by lower success rates of obstacle avoidance and deficits in restoring steady gait after obstacle crossing compared to the healthy age-matched adults. However, this study did not include biomechanical comparisons between the AFO conditions during obstacle crossing. Further, to our knowledge, there is no evidence on how using an AFO would influence propulsion symmetry, specifically during walking adaptability tasks.

The purpose of this study was to assess the influence of a commonly prescribed polypropylene articulated AFO on inter-limb propulsion symmetry during walking adaptability tasks in individuals post-stroke. Since articulated AFOs are intended to optimize ankle range of motion during stance (e.g., Tyson and Thornton, 2001), we hypothesized that walking with an articulated AFO would improve propulsion symmetry compared to walking without the AFO.

2. Methods

2.1. Participants

Nine individuals with chronic stroke participated in this study. Table 1 lists the demographic information and participant characteristics. Participants were able to walk at least 10 m independently or with supervision with and without their AFO. All participants used their own custom-made polypropylene articulated AFOs. The AFOs had a plantarflexion stop and dorsiflexion free ankle hinge and enclosed the malleoli for mediolateral stability, with the foot plates encompassing the full length of the toes (Fig. 1). The study protocol was approved by an Institutional Review Board and all participants provided written, informed consent prior to study participation.

Fig. 1. Polypropylene articulated ankle-foot orthosis with plantarflexion-stop and free dorsiflexion.

2.2. Experimental data collection

Participants walked overground on a 10-m walkway and completed a minimum of three trials of each task: walking at self-selected comfortable speed, taking a long-step, and crossing an obstacle (Fig. 2). The obstacle dimensions were 20 cm (height), 46 cm (length) and 7 cm (width), and the obstacle was placed on the force plate in the middle of the walkway at the trial onset. Prior to the obstacle and long-step trials, participants were instructed to lead with their preferred leg for 3 trials and then lead with the opposite leg for 3 trials. Participants walked at their self-selected speed to the obstacle, cleared the obstacle and continued walking to the end of the walkway. Similarly, for the long-step task, participants walked to the force plates and upon stepping on the force plate, completed a self-selected, comfortable longest step, and then continued walking to the end of the walkway. All walking trials were completed with and without their AFO. The order of tasks and AFO conditions were randomized. Reflective markers were placed on anatomical landmarks according to a modified Helen Hayes full-body marker set. A 12-camera motion capture system (VICON, Denver, USA) was used to record three-dimensional body-segment kinematics at 100 Hz. During all tasks, three-dimensional ground-reaction-force (*GRF*) data were collected at 1200 Hz using four force plates (AMTI, Inc., Watertown, USA) embedded in the ground. Surface electromyogram (EMG) (Delsys, Inc., Boston, USA) data were collected from soleus and medial gastrocnemius at 2000 Hz.

2.3. Data processing

Data was analyzed in Visual3D (C-Motion, Inc., Germantown, USA). The kinematic and *GRF* data were low-pass filtered using a fourth-order Butterworth filter with cutoff frequency of 7 Hz and 20 Hz, respectively. *GRFs* were normalized by body weight. Forward propulsion was

Table 1
Participant characteristics: gender, age, paretic side, height, weight, time since stroke and AFO prescription. Clinical tests without their AFO are listed to further characterize the participants: 10-Meter Walk Test (10 MWT), Timed Up & Go (TUG), Four Square Step Test (FSST), Functional Reach Test (FRT). A higher speed for 10MWT, a shorter time for TUG and FSST and a longer distance in FRT implies better performance. The paretic leg ankle range of motion (ROM) during the stance phase of steady-state walking without their AFO is listed below.

Subject #	Gender	Age (yrs)	Paretic Side	Height (m)	Weight (Kg)	Months Since Stroke	Months Since AFO Prescription	10 MWT (m/s)	TUG (s)	FSST (s)	FRT (cm)	Ankle ROM (deg)
1	F	49	L	1.58	77.1	19	19	1.02	11	16	24	28
2	M	65	R	1.86	90.3	36	29	0.22	46	100	18	26
3	M	46	R	1.80	82.6	57	55	0.14	61	50	24	15
4	F	38	R	1.67	76.2	15	8	0.67	24	36	11	27
5	F	55	R	1.81	80.3	36	34	1.23	11	13	21	18
6	M	69	L	1.83	88.9	35	33	1.29	11	15	33	20
7	M	68	L	1.76	87.5	53	41	0.83	19	50	17	18
8	M	43	R	1.83	79.4	44	42	0.73	20	30	29	19
9	M	50	L	1.75	83.9	84	83	0.23	49	82	22	9
Median		**50.0**		**1.8**	**82.6**	**36.0**	**34.0**	**0.7**	**20.2**	**35.9**	**21.6**	**19.3**
IQR		**22.0**		**0.1**	**9.9**	**28.0**	**25.0**	**0.9**	**36.5**	**50.7**	**8.9**	**9.6**

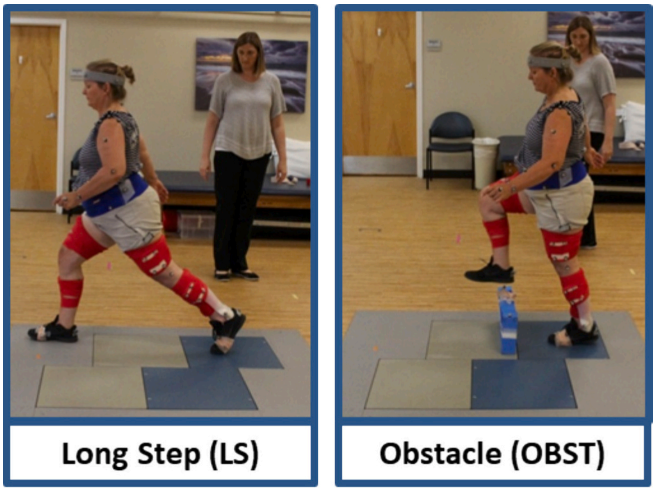

Fig. 2. Walking adaptability tasks: taking a long-step (LS) and obstacle crossing (OBST).

quantified using propulsive impulses, calculated as the net time integral of the anterior-posterior *(A/P) GRFs* during late stance for the paretic and nonparetic legs. Late stance was defined as the time period between the mid-stance gait event and ipsilateral limb toe-off. For each step, the mid-stance gait event was identified manually as the time point where the pelvis nearly aligned with the stance limb center-of-pressure and the total *GRF* vector was directed upward (i.e., the *A/P GRF* was zero). Individuals with severe hemiparesis may generate both propulsive and braking (i.e., negative) *A/P GRFs* during late stance (e.g., Peterson et al., 2010). Therefore, quantifying forward propulsion as the net *A/P GRF* impulses during late stance provides a more comprehensive assessment in individuals with varied hemiparesis severity. Inter-limb propulsion symmetry was calculated as the propulsive impulses generated by the paretic leg divided by the sum of propulsive impulses generated by both legs. For the obstacle crossing and long-step tasks, propulsive impulses from the trailing paretic and nonparetic legs were used to calculate propulsion symmetry. EMG data were high-pass filtered (30 Hz) using a fourth-order Butterworth filter, demeaned, rectified and low-pass filtered (10 Hz) using a fourth-order Butterworth filter. EMG amplitudes in each task were normalized by the mean peak amplitude of the same muscle during the self-selected walking task without the AFO. Muscle activation in each task was quantified by averaging the normalized EMG amplitudes of each muscle between the mid-stance gait event and ipsilateral limb toe-off. Also, to further characterize each task, spatiotemporal gait characteristics were calculated using three-dimensional kinematics. In addition, peak ankle plantarflexion moment during late stance as well as the peak ankle, knee and hip joint angles in the sagittal plane were calculated to further help explain the biomechanical results.

2.4. Statistical analysis

Nonparametric two-tailed Wilcoxon Signed-Rank Tests were used to compare propulsion symmetry between the AFO and no AFO (NoAFO) conditions within each task. For tasks in which significant differences were observed, nonparametric two-tailed Wilcoxon Signed-Rank Tests were used to compare other biomechanical quantities (e.g., peak ankle plantarflexor moment, peak lower extremity joint angles, and ankle muscle EMG activity) to help explain any underlying mechanisms.

3. Results

Significant differences in inter-limb propulsion symmetry were observed only during the obstacle crossing task (Fig. 3a). The median propulsion values for all the tasks are listed in Supplemental Table A.1. During the obstacle crossing task, propulsion symmetry significantly improved by 15% ($p = 0.025$; Cohen's d = 0.48) with AFO use (Fig. 3a). There were no statistically significant changes in propulsion generated by the paretic or nonparetic legs alone (Fig. 3b-c). However, AFO use

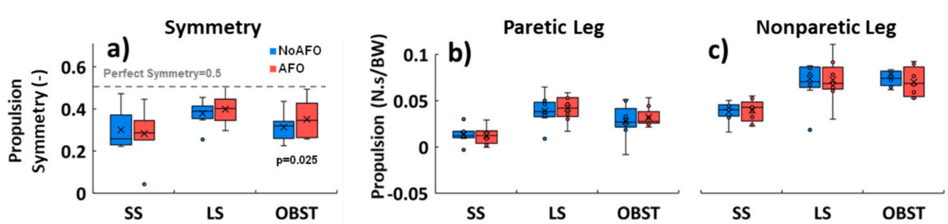

Fig. 3. Inter-limb propulsion symmetry (a); propulsive impulses from the paretic leg (b) and nonparetic leg (c) during steady-state walking at comfortable speed (SS), taking a long-step (LS), and crossing an obstacle (OBST) with (red) and without (blue) an AFO. Significant differences (p < 0.05) are shown.

Table 2
Median (IQR) Spatiotemporal data are listed for steady-state walking at comfortable speed (SS), taking a long-step (LS), and obstacle crossing (OBST). The LS and OBST tasks included trials leading with the paretic and nonparetic legs. Walking speed during the LS and OBST tasks corresponds to the stride containing the long step and obstacle crossing tasks, respectively. Step width and step length were normalized by leg length and all temporal data was normalized by the gait cycle. SLS depicts single-leg-stance. The SLS and stance durations correspond to the trailing leg support phase of the gait cycle (e.g., Fig. 2). Paretic step ratio is a measure of step length symmetry (a value of 0.5 means perfect symmetry). Significant changes between the AFO and NoAFO conditions are defined as ($p < 0.05$).

	SS			LS			OBST		
	NoAFO	AFO	*p*	NoAFO	AFO	*p*	NoAFO	AFO	*p*
Walking Speed (m/s)	0.73 (0.77)	0.77 (0.82)	**0.015**	0.77 (0.74)	0.69 (0.79)	0.859	0.57 (0.55)	0.51 (0.48)	0.767
Step Width (–)	0.25 (0.07)	0.24 (0.09)	0.086	0.26 (0.09)	0.23 (0.08)	0.051	0.23 (0.07)	0.21 (0.07)	**0.011**
Paretic Step Length (–)	0.61 (0.35)	0.64 (0.32)	0.139	0.94 (0.23)	0.93 (0.32)	0.767	0.74 (0.13)	0.72 (0.15)	0.401
Nonparetic Step Length (–)	0.48 (0.36)	0.57 (0.36)	**0.011**	0.88 (0.27)	0.87 (0.28)	0.767	0.62 (0.17)	0.63 (0.12)	0.441
Paretic SLS (% GC)	0.28 (0.17)	0.31 (0.16)	**0.028**	0.34 (0.13)	0.35 (0.15)	0.767	0.31 (0.19)	0.32 (0.13)	0.051
Nonparetic SLS (% GC)	0.40 (0.15)	0.39 (0.11)	0.440	0.42 (0.09)	0.44 (0.06)	0.678	0.46 (0.12)	0.47 (0.13)	0.208
Paretic Stance (% GC)	0.61 (0.15)	0.62 (0.10)	0.441	0.63 (0.10)	0.63 (0.12)	0.953	0.58 (0.10)	0.58 (0.09)	0.260
Nonparetic Stance (% GC)	0.73 (0.16)	0.69 (0.16)	**0.038**	0.75 (0.16)	0.71 (0.17)	0.441	0.72 (0.10)	0.71 (0.11)	0.674
Paretic Step Ratio (–)	0.54 (0.03)	0.53 (0.06)	0.110	0.49 (0.05)	0.50 (0.07)	0.990	0.54 (0.05)	0.53 (0.07)	0.093

induced the following changes in the paretic leg during stance: increased peak ankle dorsiflexion angle (NoAFO: 4.5 ± 9.4°; AFO: 6.9 ± 9.6°; p = 0.028; Cohen's d = 0.25), decreased peak knee hyperextension angle (NoAFO: −7.8 ± 10.5°; AFO: −0.77 ± 9.6°; p = 0.008; Cohen's d = 0.71), increased medial gastrocnemius activity by 22% (p = 0.038), and increased peak ankle plantarflexion moment (NoAFO: 0.83 ± 0.33 Nm/kg; AFO: 1.04 ± 0.30 Nm/kg; p = 0.011; Cohen's d = 0.64). Also, the AFO induced the following changes in the paretic leg during swing: increased peak ankle dorsiflexion angle (NoAFO: 0.20 ± 11.3°; AFO: 6.8 ± 9.2°; p = 0.017; Cohen's d = 0.64) and decreased peak hip flexion angle (NoAFO: 75.8 ± 22.4°; AFO: 70.0 ± 12.6°; p = 0.050; Cohen's d = 0.26). There were no significant differences between the AFO conditions during any of the other tasks. Lastly, the spatiotemporal gait characteristics during each walking task were listed in Table 2 to further characterize the biomechanical requirements of each task.

4. Discussion

Walking with an articulated AFO promoted improved inter-limb propulsion symmetry during the obstacle crossing task (Fig. 3a), partially supporting our hypothesis. During the paretic leg stance, the AFO corrected the excessive ankle plantarflexion and knee hyperextension during stance (Higginson et al., 2006; Mulroy et al., 2010) and induced increased medial gastrocnemius muscle activity and peak ankle plantarflexion moment in the paretic leg. The gastrocnemius can also contribute to knee flexion during late stance (Perry and Burnfield, 1992), which would also help facilitate clearing the obstacle during swing.

As expected, during the nonparetic leg stance (paretic leg swing phase), AFO use increased paretic leg ankle dorsiflexion and decreased hip flexion. The higher paretic leg hip flexion without AFO use likely served as a compensatory mechanism to facilitate foot clearance (Little et al., 2014). There were no other significant changes in the stance or swing leg kinematics during the nonparetic leg stance.

The influence of AFO on walking mechanics is likely task-dependent. While AFO use improved propulsion symmetry during the obstacle crossing, it did not influence propulsion symmetry during the long-step task. This finding is likely due to differences in the biomechanical requirements of these tasks. Obstacle crossing requires taking a high step during the single-leg-stance phase, which imposes a dynamic balance challenge (Vistamehr et al., 2017) while generating propulsion in a more upright stance leg position (e.g., Fig. 2). Therefore, any AFO induced corrections in the paretic leg kinematics during stance (e.g., excessive ankle plantarflexion and knee hyperextension) or swing (e.g., inadequate ankle dorsiflexion and excessive hip flexion) may have contributed to improved inter-limb propulsion symmetry during the obstacle crossing. In contrast, the longer step length during the long-stepping task (Table 2) required higher propulsion generation from both paretic and nonparetic legs (Fig. 3b-c), which was consistent with prior work (Clark et al., 2016). Independent of the AFO condition, the paretic and nonparetic long steps were similar in length, resulting in a symmetrical paretic step ratio (Table 2) as well as inter-limb propulsion symmetry that approached perfect symmetry (Fig. 3a). The long-stepping task requirements revealed that despite the small propulsion values from the paretic leg during the steady-state walking at their comfortable speed, individuals post-stroke had the capacity to generate higher propulsion from their paretic leg without any orthotic intervention (Fig. 3b). Therefore, consistent with findings from Clark et al. (2016), incorporating the long-step tasks in rehabilitation programs has clinical implications for promoting paretic leg propulsion output.

4.1. Study limitations

This study included a small sample of individuals with the capacity to walk with and without their own custom-made articulated AFO. Thus, these outcomes have limited generalizability to individuals post-stroke who have different walking capabilities and use other AFO types. Despite the limited sample, the outcomes were robust and provided insight into the influence of AFO use on propulsion metrics as well as the importance of incorporating walking adaptability tasks into gait assessments and interventions.

5. Conclusion

The use of plantarflexion-stop articulated AFOs improved inter-limb propulsion symmetry during the obstacle crossing task, promoting paretic leg use and reducing reliance on the nonparetic leg.

CRediT authorship contribution statement

Arian Vistamehr: Writing – review & editing, Writing – original draft, Visualization, Validation, Supervision, Software, Resources, Project administration, Methodology, Investigation, Funding acquisition, Formal analysis, Data curation, Conceptualization. **Richard R. Neptune:** Writing – review & editing, Methodology, Conceptualization. **Christy L. Conroy:** Writing – review & editing, Methodology, Investigation, Conceptualization. **Paul A. Freeborn:** Writing – review & editing, Methodology, Data curation, Investigation. **Gina M. Brunetti:** Writing – review & editing, Methodology, Investigation. **Emily J. Fox:** Writing – review & editing, Supervision, Resources, Project administration, Methodology, Funding acquisition, Conceptualization.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.clinbiomech.2024.106268.

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