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Experimental Evaluation of a Portable Powered Ankle-Foot Orthosis

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Abstract—Ankle-foot orthoses (AFOs) ameliorate the impact of impairments to the lower limb neuromuscular motor system that affect gait. Emerging technologies provide a vision for fully powered, untethered AFOs. The portable powered AFO (PPAFO) provides both plantarflexor and dorsiflexor torque assistance via a bi-directional pneumatic rotary actuator. The system uses a portable pneumatic power source (bottle of compressed CO_2) and embedded electronics to control foot motion during level walking. Experimental data were collected to demonstrate functionality from two subjects with bilateral impairments to the lower legs. These data demonstrated the PPAFO's ability to provide functional assistance during gait. The stringent design requirements of light weight, small size, high efficiency and low noise make the creation of daily wear assist devices challenging; but once such devices appear, they will present new opportunities for clinical treatment of gait abnormalities.

I. Introduction

THE authors have previously developed a portable powered ankle-foot orthosis (PPAFO) [1]. Preliminary results demonstrated the PPAFO's ability to provide functional plantarflexor assistance, while inferring dorsiflexor assistance through data collected from healthy subject data. In this paper, results from two subjects with gait impairments are presented to provide a more compressive demonstration of the PPAFO during dorsiflexor and plantarflexor assistance.

Functional performance of AFOs has been quantified with time and distance measures, such as walking velocity, cadence, step length, stride length, and cycle timing [2, 3]. Additionally, kinematic and kinetic data from motion capture systems and embedded sensors have been used for both direct performance comparisons and to calculate other parameters such as joint angles and moments for quantitative and qualitative device assessment (e.g., [2, 4, 5][1, 6-8]).

In this study, the PPAFO was used to provide functional assistance for a subject with a significant plantarflexor impairment that affected push-off during stance, and a subject with a significant dorsiflexor impairment that affected motion control of the foot during stance and swing. Time and distance measures along with leg joint kinematic and kinetics were used to quantify PPAFO performance.

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II. Methods

A. System Description

The PPAFO uses carbon composite shank and foot pieces articulated by a rotary actuator at the ankle joint, which is powered by a portable pneumatic power supply, Fig. 1. The device provides up to 12 Nm of torque at a pressure of 120 psig. Torque magnitude is modulated with two pressure regulators. Sensor feedback from two force sensors in the PPAFO foot piece trigger assistance in three regions determined by functional gait requirements: (1) dorsiflexor assist to prevent foot slap during loading response by controlling foot motion, (2) plantarflexor torque to provide assistance for propulsion during stance, and (3) dorsiflexor torque to prevent foot drop by maintaining toe clearance during swing. The force sensors in the PPAFO foot plate are used to detect the event boundaries of the three regions. Events are detected when sensor magnitudes exceed tuned thresholds for the heel and metatarsal sensors.

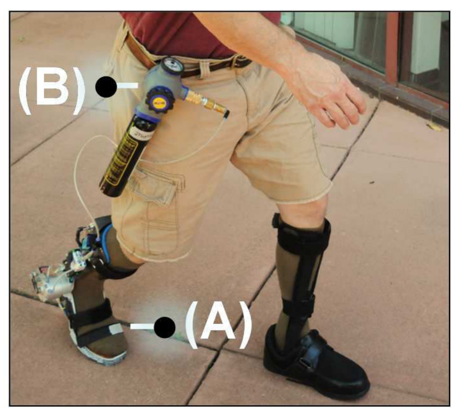

Fig. 1. The PPAFO **(A)** is shown assisting a subject with a plantarflexor impairment. A belt worn CO_2 bottle and regulator power the device **(B)**.

B. Subject Information

The plantarflexor impaired subject (male; 51yrs; height 175cm; mass 86kg) has a diagnosis of cauda equine syndrome (CES) caused by a spinal disc rupture and will be referred to as ISubPF. This bilateral impairment rendered the subject unable to generate a plantarflexor torque to push his toes down. However, the subject was able to generate and control a dorsiflexor torque to lift the toes up. ISubPF could walk without walking aids (e.g., cane or walker), but wore pre-fabricated carbon composite PPAFOs for daily use (Blue RockerTM, Allard, NJ, USA).

The dorsiflexor impaired subject (female; 37 yrs; height 157 cm; mass 62 kg) has a diagnosis of facio-scapulo-humeral type muscular dystrophy and will be referred to as ISubDF. This condition has resulted in bilateral muscle weakness to the subject's dorsiflexors, but this subject was able to plantarflex. ISubDF wore a prefabricated sleeve type ankle support in place of more traditional custom AFOs. This subject was able to walk without walking aids, but used the treadmill handrails for increased stability during testing. For both subjects, use of the PPAFO required the shoe on the right foot to be removed

All procedures were approved by the institutional review boards of the University of Illinois and Georgia Institute of Technology, and all participants gave informed consent.

C. Experimental Procedure and Data Collection

For both subjects, a comfortable treadmill walking speed was determined by averaging three self-selected speeds while wearing shoes without their daily-wear AFOs since this scenario produced the most difficult walking condition. The self-selected walking speeds were found to be 0.6 m/s and 0.2 m/s, for ISubPF and ISubDF respectively.

Kinematic and kinetic data were collected during 30 s walking trials at the self-selected speeds for three different conditions. For the kinematic data collection (Model 460; Vicon, Oxford, UK; sampled at 120 Hz), reflective markers were attached to the body, including the torso, thighs, shanks, feet, and the PPAFO. Ground reaction force (GRF) and center of pressure data (COP) were collected on a custom force-sensing instrumented split-belt treadmill sampled at 1080 Hz [9]. Data were filtered by a low-pass, fourth-order, zero-lag, Butterworth filter with cut-off frequencies of 6 Hz (treadmill) and 12 Hz (camera).

The three conditions were presented in the following order. (1) Subjects walked with running shoes on the both feet. During the shoe walking trials the subjects wore their daily-wear AFO on the left leg and no AFO on the right leg. (2) The subjects walked with the unpowered PPAFO on the right leg and a running shoe with the daily-wear AFO on their left leg. (3) The previous condition was modified by powering the PPAFO. For ISubPF, the PPAFO provided ~12 Nm (120 psig) propulsive assist during stance beginning at foot flat and continuing to heel off. For ISubDF, the PPAFO provided ~6 Nm (60 psig) assist during swing from toe off to the following heel strike.

D. Data Analysis

Kinematic and kinetic gait parameters were used to quantify the effect of PPAFO assistance on the gait of the subjects. PPAFO foot sensor and ground reaction force data were used to identify gait events and divide the data into individual cycles for each side of the body. A cycle was defined from consecutive heel strikes of the same limb. Data were normalized to 0-100% of the cycle, and aligned at the subject's average toe off for each trial.

Kinematic and kinetic data were used to calculate the following bilateral univariate parameters: ankle, knee, and hip maximum joint angle maximum ranges of motion (ROM), step length (SL), step width (SW), cycle time (CT), and stance time (ST) time. Flexion-extension joint angles were computed using the procedure proposed by [10]. A symmetry index (SI) was also calculated for the bilateral parameters [11]. This metric quantifies right (positive) or left (negative) side bias for a given gait parameter. Inverse dynamics analysis [12] was used to calculate the sagittal plane joint moments and powers for only ISubPF. Moments and powers were not calculated for ISubDF because she grasped the treadmill handrail throughout the trial leading to an unmeasured contact force during the walking trials.

III. Results

A. Results for Subject ISubPF

Right joint ROM decreased between the assisted and shoe walking trials at the ankle, but increased at the knee and hip, Table 1. The decrease in the ankle joint ROM was created by reduced dorsiflexion during late stance and early swing. Left joint ROM did not change appreciably between trials. The changing right ROM resulted in decreased symmetry index (SI) values at the ankle but large increases at the knee and hip. PPAFO assistance did not change step length on the right side, although cycle time, stance time, and step width all increased, Table 2. Step length on the left side increased during PPAFO assistance, creating a corresponding increase in cycle time and stance time, Table 2.

Right side peak GRF increased during the PPAFO assisted walking trial, Fig. 2. During the propulsive phase of gait, forward propulsive force (negative AP-GRF values) increased by 25 N (at 57% gait cycle) and vertical force (Z-GRF) increased by 112 N (at 55% gait cycle).

TABLE I Joint range of motion, mean (and standard deviation)

		Ankle ROM			Knee ROM			Hip ROM		
	Trials	Right	Left	SI	Right	Left	SI	Right	Left	SI
ISubPF	Shoes	26.3 (2.3)	18.0 (1.7)	37.3	63.7 (6.2)	58.4 (6.7)	8.7	43.3 (2.8)	42.2 (2.1)	2.7
	No Assist	21.9 (3.0)	17.7 (1.4)	21.4	65.2 (5.2)	59.9 (5.2)	8.5	49.0 (2.7)	41.7 (1.8)	16.1
	Assist	21.2 (3.0)	19.1 (1.4)	10.6	75.5 (4.4)	56.8 (10.1)	28.3	51.4 (2.4)	42.3 (3.2)	19.4
ISubDF	Shoes	27.4 (2.4)	14.4 (0.6)	62.5	60.0 (3.6)	60.8 (2.8)	-1.2	48.1 (4.1)	41.6 (2.9)	14.4
	No Assist	29.0 (2.3)	14.1 (0.7)	69.3	56.7 (6.9)	45.5 (3.8)	21.9	42.8 (5.4)	33.0 (2.5)	25.8
	Assist	18.1 (1.7)	12.7 (0.8)	35.2	49.8 (7.8)	61.5 (9.4)	-20.9	NA	NA	NA

TABLE 2 Time and distance gait parameters

	Trials	Step Length Right	Step Length Left	Step Length SI	Cycle Time Right	Cycle Time Left	Cycle Time SI	SLS Time Right	SLS Time Left	SLS Time SI	Step Width
ISubPF	Shoes	492.7 (26.7)	483.4 (28.5)	1.9	1.09 (0.03)	1.09 (0.04)	0.07	0.76 (0.06)	0.79 (0.04)	-4.26	202.6 (21.4)
ISubPF	No Assist	479.2 (27.9)	488.8 (21.4)	-2.0	1.09 (0.04)	1.09 (0.04)	-0.06	0.75 (0.05)	0.78 (0.03)	-4.91	213.5 (14.5)
ISubPF	Assist	490.0 (20.1)	512.3 (22.1)	-4.4	1.13 (0.03)	1.13 (0.03)	0.03	0.84 (0.04)	0.81 (0.04)	3.15	216.4 (16.8)
ISubDF	Shoes	364.2 (22.7)	362.9 (26.0)	0.3	2.48 (0.12)	2.49 (.17	-0.14	1.82 (0.13)	1.91 (0.15)	-4.66	175.4 (21.2)
ISubDF	No Assist	275.3 (20.7)	287.2 (25.6)	-4.3	2.21 (0.13)	2.23 (0.15)	-0.64	1.71 (0.11)	1.72 (0.12)	-0.56	193.1 (17.4)
ISubDF	Assist	276.9 (16.4)	282.0 (27.5)	-1.8	2.16 (0.09)	2.16 (0.13)	0.11	1.67 (0.06)	1.58 (0.09)	5.34	165.9 (10.2)

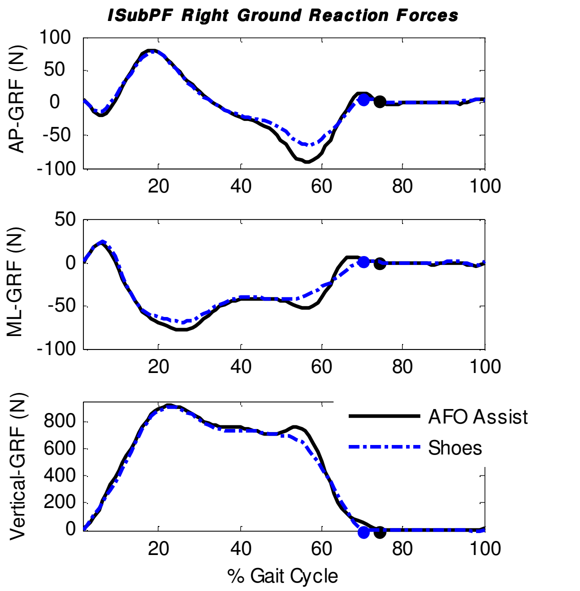

Fig. 2. ISubPF average anterior-posterior (AP), medial-lateral (ML), and vertical ground reaction force (GRF) data. Positive AP-GRF data indicates forces directed toward the anterior direction of the subject. Average toe off for each condition is indicated by a circle.

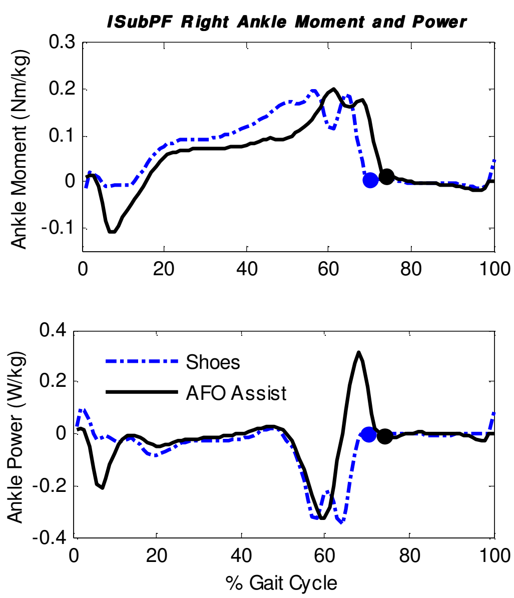

Fig. 3. ISubPF average right ankle moments (Top) and powers (Bottom) for the assisted and shoe walking trials. Positive moment values are in the plantarflexion direction. Average toe off for each condition is indicated by a circle.

Fig. 2 illustrates that stance time increased during the assisted trials. The assistance-driven increase in the forces and changes in timing contributed directly to changes observed in the moment and power generated at the assisted ankle joint, Fig. 3.

The right ankle moment for the assisted case was similar to the moment from the shoe walking data; however, the peak power at the joint increased significantly. Without assistance, the subject did not generate significant power at the end of stance for propulsion (around 70% gait cycle). With the addition of PPAFO assistance, the subject was able to generate a peak power of 0.31 W/kg at 68% gait cycle.

B. *Results for Subject ISubDF*

The ankle joint angle plots for ISubDF showed that the PPAFO assistance was able to restrict foot ROM during swing, Fig. 4 top panel. Without PPAFO assistance the joint angle dropped to ~10° below neutral during swing. However, with PPAFO assistance the joint angle was held at ~8° above neutral. The dorsiflexor assist also resulted in better positioning of the foot for initial contact at heel strike.

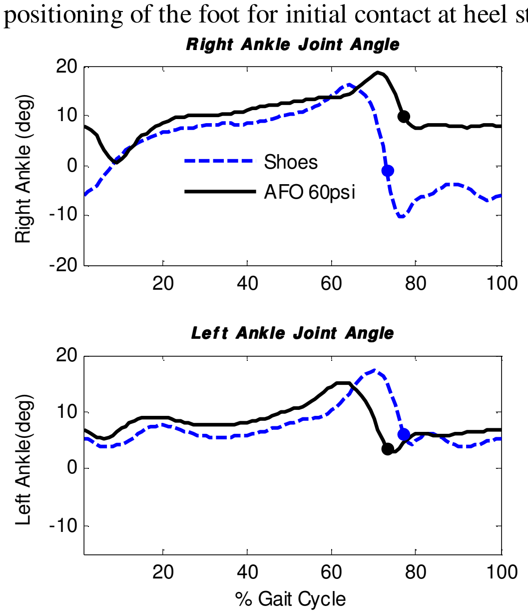

Fig. 4. Average right and left ankle angles for subject ISubDF during the shoe (top)and PPAFO assisted (bottom) walking trials.

Joint angle ROM for ISubDF on the assisted side decreased at the ankle and the knee during the assisted trial, Table 1. Joint angle ROM from the hip was not available because of missing marker data during the assisted walking trial. As with ISubPF, the reduced ROM resulted in a smaller ROM symmetry index at the ankle and increased knee ROM SI. Step length, cycle time, stance time, and step width all decreased during the assisted trial, Table 2. The SI for ST time increased by 9, but the SIs for SL and CT did not vary greatly between the shoe and PPAFO assisted trials.

IV. Discussion

The results from the walking trials in this work demonstrate that the PPAFO is capable of providing functional assistance during gait. For the subject with impaired plantarflexor function, the PPAFO assistance was observed as increased power (Fig. 3), ground reaction forces (Fig. 2), and cycle and stance times (Table 2) during the assisted walking. PPAFO assistance resulted in a peak power increase from ~0 W/kg, during shoe walking, to 0.31 W/kg with assistance. This clearly indicates the functional assistance provided by the PPAFO. While the peak power generated during the assisted walking was 36% of that generated by a healthy walker (1.55 W/kg, shoes, normal walking speed), it was a significant increase from the subject's unassisted levels. The functional benefit provided by the PPAFO was also illustrated by increased ground reaction forces in the vertical and anterior-posterior directions during the assisted trial. Without assistance, ISubPF's vertical GRF data had only a single peak present in early stance, a symptom of weak plantarflexors (Fig. 2 bottom panel, dot-dashed line).When the power assist was activated, a second peak in the vertical reaction force was present at 54% gait cycle (Fig. 2 bottom panel, solid line), and the magnitude of the propulsive force in the anterior-posterior GRF grew more negative in late stance (Fig. 2 top panel, solid line, 57% gait cycle). The second peak in the vertical GRF data was indicative of push-off propulsion during healthy gait, while increased anterior-posterior GRF indicated more force for forward propulsion. Both changes demonstrated appropriately timed functional assistance from the PPAFO.

Functional dorsiflexor assistance was demonstrated during ISubDF's assisted walking trial. When the PPAFO dorsiflexor assistance was applied, the ankle joint angle was held above neutral throughout swing (Fig. 4 top panel, solid line). PPAFO assistance prevented the foot drop present during the unassisted trial (Fig. 4 top panel, dot-dashed line) and effectively maintained toe clearance during swing. In addition to maintaining clearance, the PPAFO assistance prevented excessive dorsiflexion at heel strike (Fig. 4 top panel, solid line, 0-10% gait cycle).

Feedback from both subjects about the performance of the PPAFO was positive. ISubPF made the comment that as soon as the PPAFO assistance was applied he stopped thinking about his right leg (with the PPAFO) and instead focused on the leg with the carbon composite AFO. ISubDF was also aware of the dorsiflexor assistance as soon as it was activated. She was initially apprehensive about the size of the device. However, after walking with the assistance, ISubDF made the comment that she did not have to work as hard when the PPAFO was turned on. She went on to say that she thought that this device (even in its current form) could be a useful tool to assist impaired individuals during special tasks, such as distance walking.

V. Conclusion

Compact, lightweight, and efficient powered ankle-foot orthoses will expand rehabilitation and daily-wear assistance opportunities for individuals with impairments to the ankle joint complex. The results from this study demonstrated that the PPAFO was capable of providing untethered functional assistance for impaired walkers. Although these pilot data demonstrated that the PPAFO is capable of providing functional assistance, the quantification of the PPAFO's ability to assist gait requires further examination with an expanded subject pool. Additionally, improved performance and efficiency of the PPAFO are crucial to transitioning the device from a laboratory tool into a practical human assist device. As such, future work will be focused on improvements to system hardware and system control.

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