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ACTUATION TIMING STRATEGIES FOR A PORTABLE POWERED ANKLE FOOT ORTHOSIS

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ABSTRACT

Ankle-foot orthoses (AFOs) are used to assist persons with lower-limb neuromuscular impairments. We have developed the portable powered AFO (PPAFO). This device uses a bidirectional pneumatic actuator powered by a CO_2 bottle to provide dorsiflexor and plantarflexor torque assistance. The PPAFO operates tether-free, allowing for use outside of the laboratory. This system has been tested on one impaired and multiple healthy subjects. Timing of the assistance provided by the PPAFO has been determined by: 1) direct event detection using sensor feedback with threshold triggers, and 2) state estimation in which gait events are estimated using a cross-correlation based algorithm. Direct event detection, while simple to implement, can be unreliable for subjects with certain gait impairments. State estimation, while more complicated to implement, provides access to state information that cannot be directly measured by the AFO, which allows for greater flexibility in assistance timing. Current hardware limitations and future work are also discussed.

NOMENCLATURE

AFO	Ankle-foot orthosis.
PPAFO	Portable powered ankle-foot orthosis.
DE	Direct event detection.
CC	Cross-correlation state estimator.

INTRODUCTION

THE NEED FOR POWERED ORTHOSES

Walking is a fundamental part of everyday life for most individuals, and greatly contributes to overall quality of life. Gait itself is a cyclic task, with one cycle defined as the initial ground contact of the foot to the following contact by the same foot. During a gait cycle the ankle joint performs four main functional tasks: deceleration of the foot during loading response, support for stability during early to mid-stance, propulsion during late stance, and motion control of the foot during swing [1-3]. The ability of the ankle joint to perform these functional tasks can be impaired by injury or neuromuscular pathologies. Sizeable populations exist in the United States alone with these types of impairments: stroke (8M), spinal cord injuries (1.3M), multiple sclerosis (1M), cerebral palsy (412K), and polio (272K) [4, 5].

Pathological gait can manifest in a variety of ways, but two common lower-leg symptoms are weakness in the plantarflexors (calf muscles) or in the dorsiflexors (shin muscles). Weakness in the dorsiflexor muscles affects both the loading response and swing phases of gait. This affect could present as an audible foot slap during weight acceptance or foot-drop during swing. Weak plantarflexor muscles, on the other hand, primarily affect limb stability and propulsion [1].

Ankle-foot orthoses (AFOs) are prescribed to help correct these types of lower limb muscle impairments [6].

AFOs can be broadly grouped into three categories: passive, semi-active, and active devices. The three main types of orthoses have benefits and limitations. For example, passive AFOs provide assistance by preventing unwanted foot motion with direct resistance and are simple and commercially available. However, by their very nature, passive AFOs may inhibit desirable motion of the shank and foot, and do not provide assistive torque. Semi-active orthoses control the impendence at the ankle joint during gait for motion control, but do not provide propulsive torque (putting energy into the system). Active orthoses use feedback from electronics and sensors to control ankle joint motion and provide assistive torque [7, 8]. Unfortunately, the size and weight of the actuators and tethers, required by active systems for power or control, limits the use of active AFOs as daily-wear devices. These limitations are the primary reasons why active AFOs are used in rehabilitation or for diagnostic purposes [9].

To address these limitations the portable powered ankle foot orthosis (PPAFO) was developed. The PPAFO utilizes pneumatic power, is lightweight, compact, and tether free, which means the device can be used outside of the lab. The performance of the controller used with the PPAFO depends critically on the ability to detect gait events based on measurements from onboard sensors (e.g., accelerometers, potentiometers, and force sensors), and to use those events to determine proper assistance for the user. In this work, two event detection methods, direct event detection and cross-correlation based state estimation are introduced, and experimental results from healthy and impaired subject trials are used to compare the performance of the two techniques.

A PORTABLE POWERED ANKLE FOOT ORTHOSIS

There are particular benefits and limitations of the different methods of actuation used with the state-of-the-art powered AFOs. Electric motors come in a range of powers, are easy to control, and are clean running, but are generally high velocity, low torque actuators. Human motion, on the other hand, is typically low in velocity, but requires high torque. As a result, electric motors require a transmission for use in applications like powered orthotics. The transmissions must be co-located by the motor, and are typically at least as heavy as the motor.

Fluid power systems, on the other hand, have actuators with high force to weight and force to volume ratios, and do not require a transmission to drive the system. The pressurized fluid can be transported to the actuator through flexible tubing, which allows for flexibility in component placement. Further, unlike some high-ratio mechanical transmissions, fluid powered systems are back- drivable. Of the two types of fluid powered systems, pneumatic and hydraulic, pneumatic systems are cleaner, and the drag through the lines and orifices is smaller.

System Hardware

The pneumatic PPAFO system uses an off-the-shelf portable, compressed CO_2 bottle and regulator (JacPac J-6901-91, 9oz capacity; Pipeline Inc., Waterloo, ON, Canada) to power a dual-vane bidirectional rotary actuator at the ankle joint (Fig. 1). The rotary, dual-vane actuator (CRB2BW40-90D-DIM00653; SMC Corp of America, Noblesville, IN, USA) is rated for a maximum pressure of 150 psig. The bottle regulator controls the supply pressure for the PPAFO system. Plantarflexor pressure comes directly from the bottle, while dorsiflexor pressure is further reduced by an additional regulator (LRMA-QS-4; Festo Corp-US, Hauppauge, NY). Excessive dorsiflexor torque is not necessary to support the weight of the foot during early stance and swing, and can actually result in subject discomfort if the pressure is not reduced. Two solenoid valves (VOVG 5V; Festo Corp-US, Hauppauge, NY) control actuation timing, based on feedback from two resistive sensors (406, 2” square Interlink Electronics Inc., Camarillo, CA, USA) placed at the heel and toe of the foot plate and a potentiometer (RV4NAYSD502A 5kΩ; Honeywell-Clarostat, Morristown, NJ, USA) located at the ankle joint, which senses angular position. Onboard electronics (eZ430-F2013 microcontroller; Texas Instruments, Dallas, TX, USA) and the CO_2 bottle worn at the waist allow the PPAFO to provide untethered powered assistance.

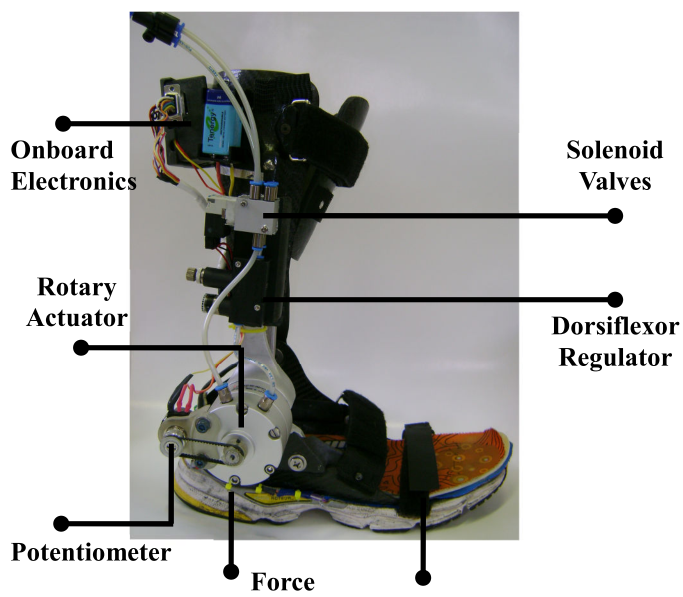


FIGURE 1. THE FIRST GENERATION PORTABLE POWERED ANKLE FOOT ORTHOSIS (PPAFO). THE ACTUATOR IS DRIVEN BY A BOTTLE OF COMPRESSED CO_2 (NOT PICTURED), WHICH IS WORN ON THE SUBJECT’S WAIST.

The output torque of the PPAFO is dependent on the pressures delivered to the actuator. When the plantarflexor valve opens, the pressure delivered to the actuator is at 110 psig (758 kPa, gauge), while the dorsiflexor valve is typically regulated to 30 psig (207 kPa, gauge). The maximum torque output of the actuator at each of these pressures is approximately 12 and 3 Nm, respectively. In healthy walkers, maximum plantarflexor torque is on the scale of 100 to 150 Nm (for a 75 kg person), while maximum dorsiflexor torque is on the scale of 10 Nm. While we could provide full 10 Nm dorsiflexor assistance at the ankle, the level of assistance is heuristically tuned to maximize subject comfort.

Functional Tasks of the PPAFO

For the healthy walkers, the gait cycle was divided into four distinct regions, bounded by specific gait events determined by the body configuration of the walker (Fig. 2). The timings of these gait events are typically expressed as occurring at certain points in the gait cycle (i.e., % gait cycle). The first of these regions, loading response, lasts from heel strike to foot flat (a-b). The second region covers early to mid-stance, and starts at foot flat and continues until heel-off (b-c). The third region, encompassing terminal stance and pre-swing, begins at heel-off and lasts until toe-off (c-d). The final region, swing, begins at toe-off and continues until heel strike (d-e).

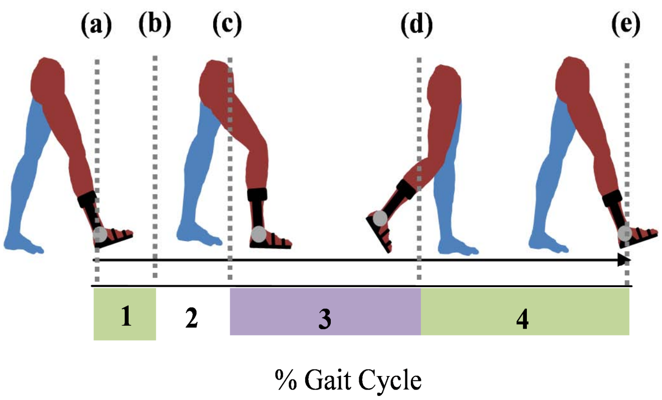


FIGURE 2. THE FOUR FUNCTIONAL REGIONS OF THE PPAFO: (1) LOADING RESPONSE (HEEL STRIKE TO FOOT FLAT, A-B), (2) EARLY TO MID-STANCE (FOOT FLAT UNTIL HEEL OFF, B-C), (3) LATE STANCE (HEEL OFF UNTIL TOE OFF, C-D), AND (4) SWING (TOE OFF UNTIL HEEL STRIKE, D-E). DORSIFLEXOR ASSISTANCE IS APPLIED IN (1) AND (4), PLANTARFLEXOR ASSISTANCE IS APPLIED IN (3).

The PPAFO applies torque during three of these regions: (1) loading response, (3) late stance to pre-swing and (4) swing phase. During (1), loading response, dorsiflexor (toes up) torque is applied to help with the controlled deceleration of the forefoot. During (2), no torque is applied at the ankle to allow for free range of motion during early to mid-stance. During (3), a propulsive plantarflexor (toes down) torque is applied until the toes leave the ground. When the foot is in (4) swing, the PPAFO provides dorsiflexor torque assistance to allow for toe clearance. Because the torque output of the current PPAFO design is relatively low, we have applied an all-or-none strategy for the level of assistance.

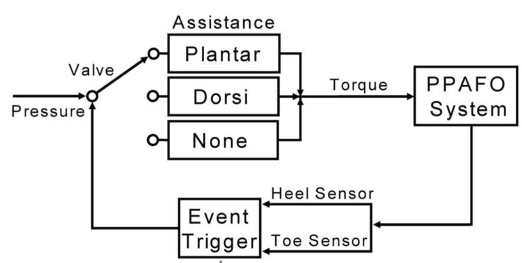
While the PPAFO can provide torque during each of these regions of gait, not all individuals require the same type of assistance. The control of the PPAFO can be changed to supply torque only when the subject requires assistance. Thus, the device can be used for subjects who have only plantarflexor or dorsiflexor weakness, in addition to subjects who have weakness in both muscle groups. Additionally, the level of dorsiflexor assistance can be further tuned with the pressure regulator to meet individual needs and prevent discomfort.

EVENT DETECTION STRATEGIES APPLIED TO THE PPAFO

As we have described above, the identification of the gait events used to determine the boundaries of the functional assistance provided by the PPAFO is key for proper assistance. In this section, we will address the problem of event detection by presenting two methods that can be used to determine the timing of PPAFO assistance (direct event detection, cross-correlation based state estimation).

Direct Event Detection

This method is a simple and easy-to-implement scheme used widely on active AFOs [10-17]. It determines gait events based on sensor feedback with threshold triggers. That is, when the sensor readings surpass a set threshold, the controller sends out a signal to open or close a valve (Fig. 3).



Gait Event	Heel Sensor		Toe Sensor		Torque Assist
	on	off	on	off	
Heel Strike	x			x	Dorsi
Foot Flat	x		x		None
Heel Off		x	x		Plantar
Toe Off		x		x	Dorsi

FIGURE 3. PPAFO FORCE SENSOR DATA ARE USED TO IDENTIFY GAIT EVENTS. THE INITIATION OF THE EVENT DETERMINES VALVE CONFIGURATION, WHICH DICTATES TYPE AND DIRECTION OF FORCE ASSISTANCE.

For the PPAFO, dorsiflexor pressure is applied when the heel sensor reading surpasses a set threshold, and is turned off when the toe sensor threshold is exceeded. When the heel sensor reading drops below the threshold value, plantarflexor torque is applied until the toe lifts from the ground. Dorsiflexor torque is applied when neither sensor is on the ground. Thresholds are heuristically adjusted for each subject to maximize user comfort. This control approach relies on the capability of the user to load and unload the force sensors at appropriate times in the gait cycle. This method fails if any of the expected sensor data are missing, and can only detect events that can be measured directly. Therefore, the use of this technique with the PPAFO may not be suitable for individuals who do not display typical loading patterns for heel-toe gait.

Cross-Correlation Based State Estimation

We posit that it should be possible to estimate the timing of gait events using limited sensor data. The following approach estimates gait events using a cross-correlation (CC) based algorithm which uses measurements from the three PPAFO sensors$y \in i^3$. This technique relies on the assumptions that gait is cyclic, and that body configuration, and thus gait events, can be approximated by a state variable [2, 18]. The state variable, i.e., percent gait cycle, is denoted by $\lambda \in [0,100)$. A given gait event is then associated with certain values of this state variable, e.g., loading response ends when λ = 15% gait cycle.

Our goal is to compute an estimate $\hat{\lambda}(t)$ of the state $\lambda(t)$ at the current time t based on the sensor measurements from the PPAFO $\{\mathbf{y}(s) \mid s \in [0,t]\}$ up to this time. The CC estimator relies on a precomputed model $\bar{\mathbf{y}}(\lambda)$ to determine what sensor measurements to expect at a given state λ. Each model is subject-specific, and does not change once created. This model is determined through a regression analysis of training data (λ, y), where λ is the percent gait cycle. The training data consists of multiple gait cycles of experimentally collected kinematic and kinetic data.

Individual sensor training data were divided into gait cycles using vertical ground reaction force data. These data were then normalized to 0-100% gait cycle, with each 1% indicating one state.

A locally weighted regression (LWR) analysis [19] was then used to determine the functional relationship between the percent gait cycle λ and the actual measurement from each sensor **y**. This analysis produces a regression model for each sensor over one gait cycle $\bar{y}(\lambda)$. The three sensor regression models are pre-computed for each subject and form a regression model matrix $\bar{\mathbf{y}}(\lambda)$. The results from applying this regression analysis to multiple gait cycles of a healthy subject can be seen in Fig. 4.

To estimate the gait state while wearing the PPAFO $\hat{\lambda}$, cross-correlation was used to compare the regression model matrix $\bar{\mathbf{y}}$ with sensor readings **y**.

At each data sampling time t_j, the gait state can be found from following equation:

$$\lambda_j = 100\left(\frac{t_j - (t-T)}{T}\right) \qquad (1)$$

where t is the current time.

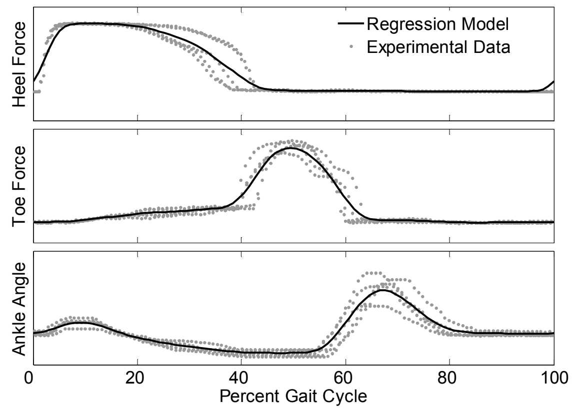


FIGURE 4. SAMPLE REGRESSION MODEL SHOWING FOUR CYCLES OF SENSOR READINGS FROM A HEALTHY SUBJECT DURING LEVEL WALKING AT A CONSTANT, SELF-SELECTED SPEED.

During walking, the CC estimator slid a window of sensor data for the past gait period T along the corresponding regression model at a resolution of each percent gait, and found the point where the mean-square-error was minimized. In other words, the estimator identified where the model best correlated to the actual data, and gave an estimate of the state. Given the regression model $\bar{\mathbf{y}}$ and the average period (defined as the time from heel strike to the consequent heel strike), T, we can apply the CC approach to estimate the state, λ, at each time, t.

This approach can be described in the following expressions. In the time history of data, each data point can be assigned a state index I_j according to Eq.2, such that each I_j will be an integer index between 0 and 100.

$$I_j = \text{round}(\lambda_j) \qquad (2)$$

The measurements are denoted by $y(\ j\ ) = y_j$. The regression model is wrapped around periodic boarders by setting $\bar{y}[i] = [i\ \pm 100\ ]$ for all i. The cost function is described in Eq.3:

$$\sum_{j=1}^{m}\left(\bar{y}[I_j + k] - y[j]\right)^T \left(\bar{y}[I_j + k] - y[j]\right). \qquad (3)$$

The integer $k \in \{0,1,\ldots,99\}$ that minimizes Eq. 3 is the state estimate.

EXPERIMENTAL EVALUATION OF THE PPAFO

To evaluate the functionality and the performance of the different actuation timing strategies, the PPAFO was evaluated by healthy and impaired subjects. All subjects walked with the PPAFO on an instrumented treadmill while both kinematic and kinetic data were collected. For the motion data, thirty-two reflective markers were attached along the body, including torso, thighs, shanks, feet, and PPAFO. Data from the healthy subjects were collected at the University of Illinois. Ground reaction force data for each foot were collected on a split-belt treadmill with embedded force plate sampling at 1500Hz (Bertec, Columbus, OH, USA). Data from the impaired subject were collected at Georgia Institute of Technology. The kinetic data were collected on a custom force-sensing instrumented split-belt treadmill sampled at 1080Hz [20]. 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.

Subject Information

Healthy Subjects. The five healthy male subjects (28 ± 4 years; height 186 ± 5cm, mass 72 ± 8kg) had no gait impairments and no history of significant trauma to the lower extremities or joints.

Impaired Subject. The impaired male subject (51 years; height 175 cm, mass 86 kg) has a diagnosis of *cauda equine syndrome* (CES) caused by a spinal disc rupture. This gait deficit caused his inability to generate plantarflexor torque on either side. The subject walks without the use of walking aids (i.e. cane or walker), but usually wears passive AFOs bilaterally. For testing, he wore his own pre-fabricated carbon composite AFO (Blue RockerTM, Allard, NJ, USA) on his left leg while walking with the PPAFO on his right leg because only the right ankle AFO was available for this experiment.

Determining the Self-Selected Walking Speed

A self-selected walking speed for each subject was determined prior to testing. For the each of the healthy subjects, comfortable walking speed was determined by averaging three self-selected walking speeds chosen while wearing the PPAFO with no actuation. Average walking speed for the five healthy subjects was 1.18 ± 0.11 m/s. The impaired subject's comfortable walking speed was determined while walking in a pair of running shoes on the treadmill with no assistive devices. This condition was chosen because it was the impaired subject's most difficult scenario. The impaired subject's comfortable walking speed was 0.7 m/s.

Training Data for the Estimation Model

The training data was collected during a 30s trial, with the subject walking on the treadmill with the un-actuated PPAFO. Heel, toe, and angle sensor data were collected and used to create the regression model needed for the estimator control.

Experimental Testing Procedure

For each event detection scheme, four experimental trials were performed to evaluate performance. These trials were conducted at two walking speeds and with and without assistive torque from the PPAFO. During the actuated healthy trials, a direct event detection scheme was used to trigger the assistance. Plantarflexor (toes down) torque was applied if both the toe and heel sensors were loaded, and a dorsiflexor (toes up) torque was applied if both sensors were unloaded—otherwise, no torque was applied. For the impaired subject, plantarflexor torque was applied if either of the sensors was loaded, and a dorsiflexor (toes up) torque was applied if both sensors were unloaded. CC could also have been used to trigger the assistive torque in these experiments. However, by choosing to view applied torque as a disturbance, we can examine the estimator independently of the control timing policy used. For each test, the subjects were given time to reach a steady state walking speed on the treadmill before data collection started. Thirty seconds of data were recorded for each of the trials.

1. Normal Speed-No Actuation. This test compared the PPAFO detection schemes under nominal un-actuated conditions. This test condition was considered to be the baseline test case. Each subject walked at his self-selected walking speed (normal speed) with no assistance provided by the PPAFO.

2. Normal Speed-Actuation. During these trials, the actuator was pressurized at 110psig (758kPa, gauge) plantarflexor, and 30psig (207kPa, gauge) dorsiflexor assistance.

3. Slow Speed-No Actuation. This trial examined the effects of slow walking (75% of the comfortable walking speed) with no PPAFO actuation.

4. Slow Speed- Actuation. The treadmill was set to 75% of the subject's self-selected speed, along with actuation. The actuation was applied in the same manner as that in trial (2) above. This test condition adjusts two conditions (speed and actuation) relative to the baseline case.

Estimation Comparison Metrics

Event times were chosen to serve as the metrics for comparison between the cross-correlation (CC) based state estimator and the direct event (DE) detector during the identification of specific events because DE detects events without state estimation. For the comparison, reference gait events were identified from ground reaction force data. Errors (in ms) between the reference event and the events identified by CC and DE were then calculated.

The gait events selected to evaluate the performance of the state estimator were right heel strike, left toe off, left heel strike, and right toe-off. The DE can only detect right side events, because the PPAFO was only on the right leg. Because the impaired subject was unable to reliably achieve the toe sensor threshold, the gait events for this subject were defined as right heel strike and right heel off.

RESULTS

For the healthy subjects, both the DE and CC methods were able to detect the specified events with reasonable accuracy during the baseline trial (1). During the remaining walking trials (2-4), the DE scheme was up to 63% more accurate than CC (Tab. 1) in terms of RMS error.

TABLE 1. EVENT DETECTION ERROR RESULTS FROM DIRECT EVENT (DE) AND CROSS-CORRELATION STATE ESTIMATOR (CC) FOR HEALTHY SUBJECTS.

5 Healthy Subjects					
Actuation	Speed	Method	RMS Error (ms)	Avg. Error (ms)	Worst (ms)
0 psi	Normal	DE	5.8 ± 2.8	2.3 ± 2.0	13.2 ± 6.8
		CC	14.8 ± 4.8	-1.7 ± 2.9	35.5 ± 19.1
110 psi	Normal	DE	31.3 ± 13.2	13.3 ± 11.9	77.2 ± 66.4
		CC	45.9 ± 15.0	17.0 ± 36.7	113.8 ± 54.8
0 psi	Slow	DE	14.2 ± 5.1	-0.9 ± 5.2	25.3 ± 28.6
		CC	49.4 ± 23.9	-29.0 ± 25.2	37.7 ± 26.1
110 psi	Slow	DE	37.9 ± 12.6	5.9 ± 15.5	59.8 ± 22.4
		CC	84.0 ± 53.2	-51.5 ± 52.9	47.7 ± 24.6

Additionally, because DE does not require training data it was simpler to implement on the PPAFO than CC. As a consequence of DE's inability to detect all four gait events, the results in Table 1 present the performance of CC during the detection of four events (heel strike and toe off for both feet) and DE during two events (heel strike and toe off for right foot). The CC demonstrated similar errors for all of the four gait events, which makes it fair to compare the 4-event result from CC to the 2-event result from DE.

For the impaired subject, the accuracy of the event detection was reduced for both CC and DE, as compared to the healthy subjects, Table 2. During the impaired subject's trials the DE RMS error was up to 52% smaller than that of the CC estimate. However, DE had performance limitations during the impaired trials. Because of the impaired subject's weakened plantarflexor muscles, he was unable to push his toes towards the ground. To ambulate, he adopted a heel walking strategy. As a result, he maintained heel contact throughout the stance phase and had minimal pressure on the toe sensor. During our initial trials with this subject, we were unable to reliably detect right toe-off with DE for the actuation trials. We were therefore forced to modify the events used to trigger the assistance during trials 2 and 4. This highlights a shortcoming of the DE approach that can be addressed by CC.

TABLE 2. EVENT DETECTION ERROR RESULTS FROM DIRECT EVENT (DE) AND CROSS-CORRELATION STATE ESTIMATOR (CC) FOR THE IMPAIRED SUBJECT.

Impaired Subject					
Actuation	Speed	Method	RMS Error (ms)	Avg. Error (ms)	Worst (ms)
0 psi	Normal	DE	39.1	3.0	78.2
		CC	36.8	-3.6	76.7
110 psi	Normal	DE	51.3	-25.8	19.2
		CC	87.6	-64.2	23.4
0 psi	Slow	DE	47.9	-8.96	78.0
		CC	74.8	-47.2	62.3
110 psi	Slow	DE	50.2	-19.4	67.6
		CC	105.5	-81.1	21.5

DISCUSSION

We have presented a portable powered orthotic device (PPAFO) that can be used to provide assistive torque at the ankle joint. In this paper, we also presented two methods that can be used to determine the timing of assistive torque during gait. This device could prove beneficial to a large population of individuals with lower leg muscle impairments, as it is capable of not only preventing undesirable motion of the foot, but is also able to provide propulsive torque. This system has been tested on both healthy and impaired subjects. The remainder of this section will discuss the performance, robustness, applications to control, and limitations of the two timing schemes.

The relative performance of the event detection methods was evaluated by the accuracy of the actuation timing. With the current system hardware, namely the solenoid valves, it is not feasible to modulate the level of assistance in a given actuation cycle. The pressure delivered to the actuator is 115 psig (793 kPa, gauge), 30 psig (207 kPa, gauge), or 0 psig. Future work will evaluate how modulating the level of assistance can allow for meeting different functional tasks of the PPAFO.

During healthy, unperturbed normal gait, both the CC and DE methods were able to detect specified events with reasonable accuracy. The DE method had more accurate results than the CC method during the remaining trials, and was also easier to implement on the PPAFO, as it did not require a training period. The CC can be improved by parameterizing the training model with more gait periods, and better results are expected. The strength of the CC algorithm appeared in its ability to detect gait events on the non-instrumented side of the body.

Although DE was found to be effective in predicting right side gait events in healthy subjects, this method was initially unable to detect gait events on the impaired subject. Certain impaired walking patterns make event detection via direct sensor measurements difficult, causing the direct event estimator to perform poorly. The gait events of interest had to be modified before DE could effectively detect when to apply assistance. In contrast, the cross-correlation method was able to estimate gait events even with limited sensor data, and as such was more robust to gait impairments.

Many powered AFOs rely on gait events to determine control objectives [10-17], which means that reliable gait event detection is required for system control. Exceptions are the orthotic systems which use surface EMG to directly control actuation [21]. This approach eliminates the need for detection of gait events, but relies heavily on the signal reliability and availability. Current work has demonstrated that the cross-correlation estimator is able to accurately and robustly determine gait events using data from PPAFO sensors.

There are currently two limitations to the proposed state estimation approach: the necessity of the preliminary calibration process, and that our method was only examined during level treadmill walking. While the preliminary training process is required to build the models, it was time consuming and could serve as an impediment for use in a clinical setting. There were also inaccuracies in the estimate due to mismatched calibration and testing conditions. These could both be addressed by continuously updating the regression model during walking. This would allow the system to adapt to changing environments, reduce the amount of calibration required to build the models, and improve controller robustness since the models would continually be constructed. The second limitation was that we only examined system performance under steady-state, level walking in the gait lab. To successfully implement the cross-correlation estimation technique outside of the lab, it would be necessary to address other modes, such as over ground walking, ramp walking, and stair ascent/descent. A possible approach to this would be to develop individual mode models and apply a methodology to identify and switch between modes.

CONCLUSIONS

In this paper, we presented a novel, untethered, powered, ankle-foot orthosis design that controls and assists ankle motion using plantarflexor and dorsiflexor torque at the ankle joint. This device is called a Portable Powered Ankle-Foot Orthosis (PPAFO). It utilizes pneumatic power to drive a rotary actuator that provides active ankle torque assistance during gait. The system is self-contained and uses a portable pneumatic power source (provided by compressed CO_2) and embedded electronics to control the actuation of the foot. Timing of the actuation was determined using two methods: direct event detection (DE) and cross-correlation based state estimation (CC). Testing on both healthy controls and an impaired subject found that while the DE approach was simple to implement, it can be unreliable for subjects with certain gait impairments. CC state estimation, while more complicated to implement, provides access to state information that cannot be directly measured by the PPAFO, and is thus more robust to abnormal gait patterns. Further work to improve the CC estimation technique and control of the PPAFO with this technique in different locomotion modes is necessary.

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