Closed-loop control of ankle plantarflexors and dorsiflexors using an inverted pendulum apparatus: A pilot study

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***Abstract*—Considerable demand exists for a device to facilitate hands-free, stable stance in individuals with neurological disorders such as spinal cord injury (SCI) and stroke. In this regard, applying functional electrical stimulation (FES) to muscles of the lower limbs in closed loop has shown promise. In particular, it has been suggested that a PID control strategy could offer functional benefits to stability by mimicking the neurological control strategy employed in able-bodied stance. In this proof of concept study, we tested this assertion by examining the potential of a PID control strategy with gravity compensation to effectively maintain balance during quiet stance by regulating FES-induced contractions of the ankle plantarflexors and dorsiflexors in able-bodied individuals. A novel Inverted Pendulum Standing Apparatus (IPSA) was employed to simulate quiet stance whilst minimizing the voluntary control of able-bodied subjects. Quiet and perturbed standing trials were performed in 3 able-bodied subjects. Performance metrics including those pertaining to stability during quiet stance (root mean square difference), perturbation rejection capabilities (settling time, peak deviation), and ability to transition from an offset initial position (settling time), were examined. For all 3 subjects and for all of the metrics examined, our results showed that the proposed closed-loop controlled FES system improved performance in comparison to voluntary control. These results indicate that the PID plus gravity control strategy used in this study offers meaningful benefits over voluntary control in terms of standing stability. Thus, the controller could potentially be applied to the problem of improving or restoring standing ability in some neurologic patient populations.**

***Index Terms*—closed-loop control, functional electrical stimulation, Inverted Pendulum Standing Apparatus, PID controller, plantarflexors and dorsiflexors.**

Manuscript received August 31st, 2013.

Funding for this study was provided by the Natural Sciences and Engineering Research Council of Canada (Grant #249669), the Toronto Rehabilitation Institute, Ontario Ministry of Health and Long-Term Care. H. Rouhani was supported by Swiss National Science Foundation (Grant # PBELP3-137539).

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DOI: 10.2298/JAC1301031S

I. INTRODUCTION

SPINAL CORD INJURY (SCI) patients suffer from a diminished ability or complete inability to maintain balance during quiet stance due to deficits in sensory and/or motor function [1]. Other conditions including stroke [2] and traumatic brain injury [3] can also lead to impaired standing ability. Significant demand exists for techniques aimed at enabling stable stance in these populations as a means of improving quality of life and independence. Furthermore, facilitating stance through activation of paralyzed or paretic muscles minimises the likelihood of secondary complications arising from confinement to a wheelchair, such as osteoporosis, urinary tract infections, spasticity, pressure ulcers and cardiovascular disease [4-7]. Facilitating stance without the need for upper limb support would be particularly advantageous as patients would then be free to use their hands to perform activities of daily living.

To this end, closed-loop control strategies that modulate functional electrical stimulation (FES) levels applied to muscles of the lower limbs in response to fluctuations in balance have shown promise. However, previously investigated control strategies have produced mixed results, require bracing of multiple joints, and/or lack strong physiological bases [8], [9].

Studies performed by our lab and others have suggested that a feedback mechanism based largely on changes in the centre of mass (COM) displacement and velocity could be responsible for modulating balance and overcoming large neural and torque generation delays in able-bodied subjects [10-13]. In these studies, proportional-derivative (PD) or proportional-integral-derivative (PID) controllers fit very well with physiological control strategies. Moreover, multiple studies have modelled the neural control component of able-bodied stance with PD or PID controllers, and have used these models, in combination with passive standing components, to successfully predict experimental results [14-18]. Thus, we have hypothesized that PD or PID control strategies regulating FES levels applied to muscles of the lower limbs, could facilitate stable stance by mimicking the control strategies employed by the central nervous system in able-bodied stance. In fact, in a preliminary study we have experimentally proven that a FES system with PD controller can successfully stabilize a SCI patient's standing posture [19].

In a proof of concept study, presented herein, we have used an Inverted Pendulum Standing Apparatus (IPSA) [20] to simulate quiet stance in able-bodied individuals (Fig. 1).

We then instrumented the ankle joints of able-bodied subjects with an FES system that used a PID control strategy with gravity compensation to actuate the ankle joints and regulate their posture/angles. The IPSA was developed in-house to allow safe investigation of FES control of ankle dorsiflexion/plantarflexion, whilst minimizing voluntary control. During experiments, the subject's feet were attached via foot straps to an inverted pendulum which rotated in the subject's sagittal plane in response to torque applied by the ankle plantarflexors and dorsiflexors to foot plates. Subjects in a standing position were supported by a mechanical frame which locked the knees and hips in extension, thus helping to remove anti-gravity muscle activity in the lower limbs [21]. This frame also avoided multi-link behaviors of human bipedal posture, thus allowing investigations to be limited solely to the ankle joints, which are key controlling joints for postural equilibrium. Furthermore, the frame maintained the subject's body in an upright, stationary position, thus decreasing vestibular and proprioceptive sensory contributions. Visual feedback about the IPSA's inverted pendulum position was also removed by utilizing an eyes closed condition. As such, the device mimicked the subject's body during quiet stance, but voluntary control was minimized as much as possible. Our previous study using the IPSA has provided preliminary evidence for the potential of a PID control strategy for the plantarflexors [20]. This study expands upon and increases the physiological relevance of that previous research by incorporating dorsiflexor control, increasing the weight of the inverted pendulum, and testing in more subjects.

II. METHODS

A. EXPERIMENTS

Trials were performed in three able-bodied male subjects. Demographic information (age, height, weight) is presented in Table 1.

Stimulation electrodes were applied bilaterally to the posterior and anterior calf in order to stimulate the gastrocnemius/soleus muscles (for plantarflexion) and tibialis anterior muscles (for dorsiflexion), respectively. For plantarflexion, 5 cm x 9 cm electrodes were applied bilaterally along the midline of the posterior calf. The anode was placed approximately 2 cm below the popliteal fossa over the gastrocnemius and soleus muscle motor points so as to activate both gastrocnemius heads as well as the soleus muscle. The cathode was placed around the lower end of the gastrocnemius muscle belly just above the ankle joint. For dorsiflexion, 5 cm x 5 cm electrodes were applied. The anode was placed over the motor point of the tibialis anterior, just lateral to the fibula, and the cathode was placed approximately 8 cm below the anode.

A 4-channel stimulator (Compex Motion, Compex SA, Switzerland) was used to produce a balanced, biphasic, rectangular, asymmetric waveform of frequency 20 Hz and pulse width 300 µs. The amplitude of the stimulation current was used as the controlled variable. In a standing position within the standing frame, the subject's knees and hips were locked in extension, and the subject's feet were strapped onto footplates attached to the IPSA (Fig. 1). Initially, the IPSA was locked at a 90° ankle joint angle such that the inverted pendulum was unable to rotate. In this posture, a torque transducer (TS11-200, Durham Instruments,

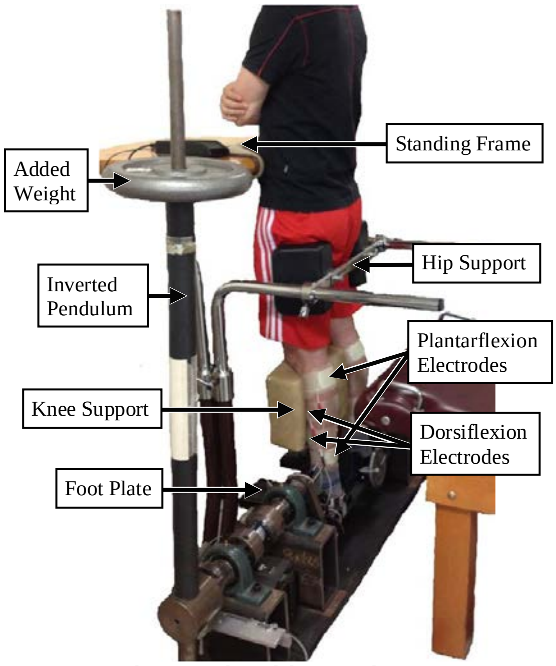

Figure 1. The Inverted Pendulum Standing Apparatus (IPSA) experimental setup.

Germany) was used to measure the isometric torques generated in response to varying stimulation current levels applied to the plantarflexor and dorsiflexor muscles. The stimulation-torque relationships obtained were then employed by the controller to ascertain the requisite stimulation current to apply in real-time. Isometric torque recruitment curves were appropriate for this experiment since the ankle angle did not fluctuate by more than +/- 6° during the experiments from the nominal angle of 5°, indicating that length and velocity components of the muscle contraction dynamic did not play any significant role.

The block diagram of the control strategy employed is depicted in Fig. 2. The inverted pendulum angle in the sagittal plane, obtained using a laser displacement sensor (LK500, Keyence, Japan), was employed as a feedback signal in real-time. The reference angle of the inverted pendulum was set to 5^{o} from vertical to simulate the typical posture observed during able-bodied stance [11]. During trials, the error signal between the angle of the inverted pendulum and the reference angle was inputted into a control system which dictated the stimulation current to apply to the plantarflexors and dorsiflexors in real time. Two separate PID controllers were implemented for each muscle group and a gravity term was incorporated to compensate for the torque due to gravity. The outputs of the controllers indicated the required torques, which were then converted to stimulation current levels based on the established stimulation-torque relationships. Depending on the direction of the requested torque, stimulation was only applied to one muscle group at any given time (i.e., co-contractions were not allowed). The adjustable PID parameters included proportional gain (K_P), integral gain (K_I) and derivative gain (K_D) terms, as well as a derivative filter term (N) to avoid unwanted amplification of high frequency noise. These parameters were initially selected based on the results of previous studies performed by our lab and others [13], [14], [19]. However, the parameters were then tuned based on performance during preliminary

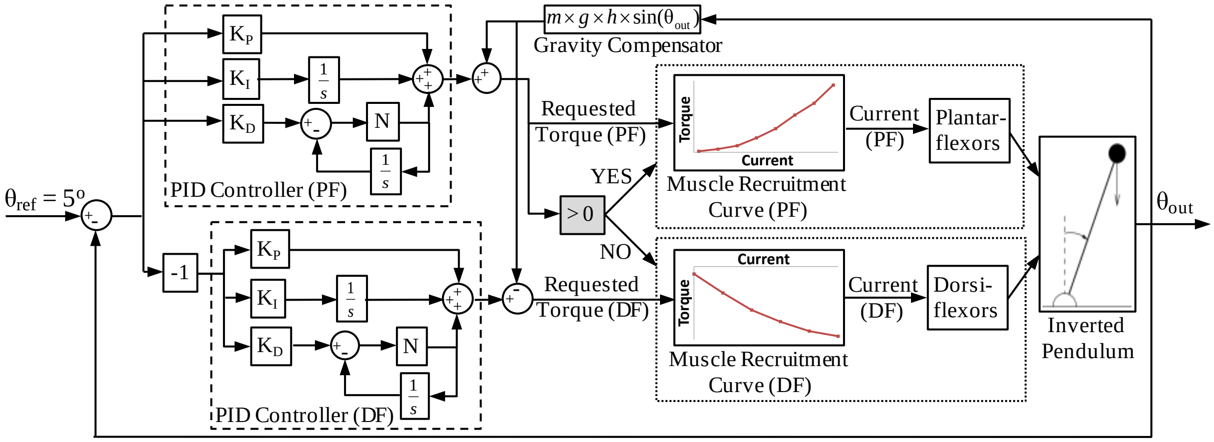

Figure 2. Block diagram of the control strategy employed. DF = Dorsiflexors; PF = Plantarflexors

trials. An additional 22.7 kg was added to the inverted pendulum to yield a total mass of 40.0 kg and centre of mass height of 0.70 m.

During trials, the subjects were instructed to relax with eyes closed. Trials performed included: (i) 10-minute quiet standing trials; and (ii) 1-minute perturbation trials in which 4 moderate posterior perturbations (i.e., in the direction of increasing plantarflexor activation) were manually applied to the inverted pendulum at random times during the 1-minute trial. In both cases, trials were initiated with the inverted pendulum resting on a support surface at an angle offset from the reference angle by approximately 8°. Thus, plantarflexor activation was required to bring the inverted pendulum towards the reference angle, and the ability of the controller to compensate for this offset initial position could be tested.

As a control, trials without FES were also performed during which the subject was instructed to attempt to balance the inverted pendulum at the reference angle throughout. In this paradigm, the eyes closed condition was retained and the subject was given verbal cues at the beginning of a trial. These trials therefore quantified the ability of the subject to balance the inverted pendulum using his remaining intact sensory inputs, the most relevant of which were proprioceptive and somatosensory inputs from the feet and ankles. Thus, any improvements in performance observed with the FES controller activated, in comparison to this control case, could be attributed to the controller. In order to account for the potential effects of fatigue, the order of the trials was randomized, and the subjects were allowed to rest in between trials.

B. Data Analysis

Following completion of trials, various performance metrics were examined. From the 10-minute quiet standing trials, mean inverted pendulum angle was calculated. The inverted pendulum sway was then quantified by calculating the root mean square difference (RMSD) in relation to the mean. For these 2 measures, the first 10 seconds of the trials were disregarded because of the offset initial position of the inverted pendulum. In contrast, by examining the first minute of these trials, settling time to within 1.5° of the reference angle could be calculated in order to investigate ability to compensate for an offset initial position. A more restrictive definition of settling time was not employed due to the presence of small fluctuations of the inverted pendulum angle at times during quiet stance (in both the FES and voluntary control conditions – see Results).

From the perturbation trials, peak posterior deviation of the inverted pendulum and settling time (to within 1.5° of the pre-perturbation angle) following each perturbation were calculated. Mean peak posterior deviation and mean settling time were then calculated by averaging over the 4 perturbations in each trial.

III. Results

In all 3 subjects, establishing appropriate parameters proved to be a relatively straight-forward procedure and the performance of the controller was not unduly sensitive to small variations in parameters. For each subject, the same controller parameters selected (Table 1) were found to be appropriate for both plantarflexors and dorsiflexors.

In Fig. 3, inverted pendulum angles during 10-minute quiet standing trials with and without the closed-loop controlled FES system applied are depicted for each of the 3 subjects. In these trials, the performance of the PID plus gravity controller during quiet stance compared favourably with voluntary control as evidenced by mean inverted pendulum angles which were closer to the reference angle and lower RMSD values for all subjects (Table 2).

Improvements in performance during transitions from offset initial positions were also observed with the control system implemented. In Fig. 4, which depicts magnified views of the beginning of the 10-minute trials depicted in Fig. 3, use of the closed-loop controlled FES system facilitated much improved transitioning from offset initial positions compared with voluntary control trials for all subjects, as evidenced by faster

TABLE 1: DEMOGRAPHIC DETAILS AND PID CONTROLLER PARAMETERS

	Age	Height (m)	Weight (kg)	K_P (Nm/ rad)	K_I (Nm/ rad s)	K_D (Nm s/ rad)	N (1/s)
Subject 1	30	1.70	65	800	0	250	150
Subject 2	22	1.78	75	350	50	250	150
Subject 3	32	1.69	75	350	50	250	150

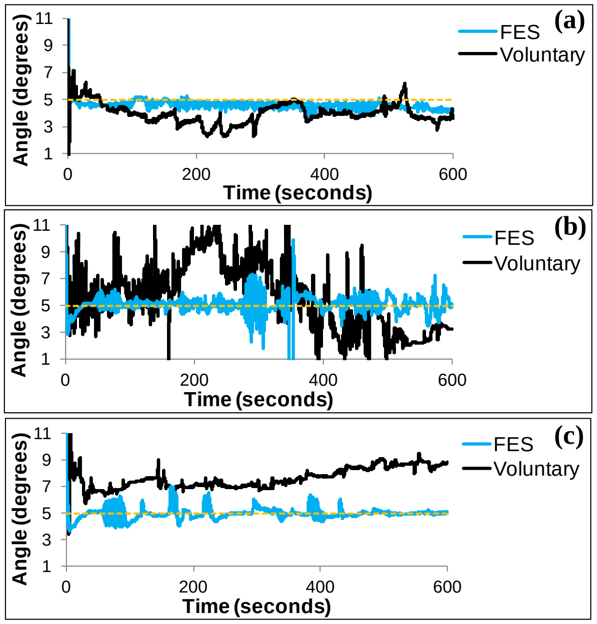

Figure 3. Inverted pendulum angle during 10-minute quiet standing trials with the closed-loop controlled FES system activated (blue) and with voluntary control (black). (a) Subject 1; (b) Subject 2; (c) Subject 3.

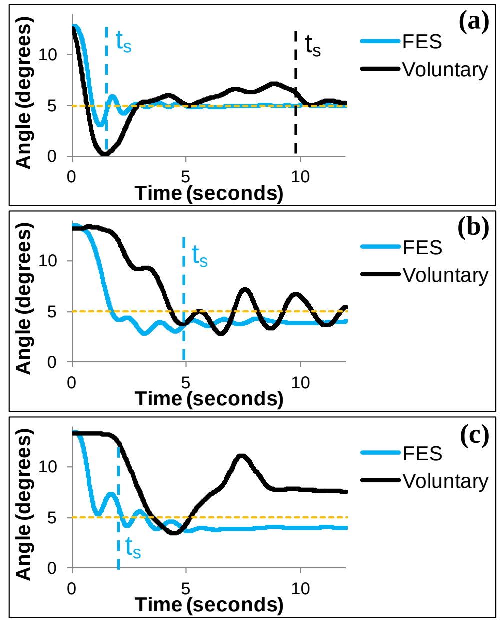

Figure 4. Enlarged view of the inverted pendulum angles during the initial 12 seconds of the 10-minute quiet standing trials in Fig. 3. The inverted pendulum begins from an angle offset from the reference angle. (t_s = settling time to within 1.5° of the reference angle) (a) Subject 1; (b) Subject 2; (c) Subject 3.

settling times to within 1.5° of the reference angle (Table 2).

In Fig. 5, inverted pendulum angles are depicted for 1-minute trials in the presence of external posterior perturbations of the inverted pendulum in FES and Voluntary conditions. From these plots, it is evident that the closed-loop controlled FES system allowed for more rapid and more reliable transition towards the reference angle following perturbations. Decreased mean peak posterior deviations and mean settling times were observed with the closed-loop controlled FES system in all subjects (Table 2).

In Fig. 6, a magnified view of a typical section of the 10-minute quiet standing trial with the closed-loop controlled FES system for Subject 1 (see Fig. 3(a)) is depicted in the top plot, whilst the bottom plot depicts the corresponding plantarflexor and dorsiflexor torques requested and the current amplitudes applied. Note that the absolute values of the requested torques were used such that all plots were visible on the positive vertical axis. The exhibited behavior (small oscillations in the inverted pendulum angle, low but mostly non-zero and gradually fluctuating levels of current applied to the plantarflexors, and only brief, intermittent activation of the dorsiflexors) has been commonly observed across multiple trials and across all 3 subjects.

IV. DISCUSSION

Once appropriate controller parameters were established, the PID plus gravity compensation control strategy was able to maintain the IPSA's stability reliably and reproducibly during prolonged quiet stance and in the presence of external perturbations for all 3 subjects. The control strategy offered considerable benefits over voluntary control in terms of minimizing steady state error, responding to an offset initial position and overcoming perturbations. This is evidenced by reliably improved performance in all 3 subjects for all of the metrics examined. It is also evident that all 3 subjects experienced difficulty in maintaining the inverted pendulum about the reference angle reliably in the voluntary control condition. This verifies that use of the IPSA to suppress vestibular, visual and some proprioceptive sensory inputs has in fact resulted in a disruption to voluntary control. Thus, the results indicate the potential of the control strategy to overcome deficits in voluntary control. This capability could potentially be applied to the problem of facilitating or improving stance in patient populations.

Our results have also verified the potential of the IPSA as a tool to assist in the continued development of closed-loop FES control strategies (PID or other control strategies) for the ankle joint. In particular, the ability to parse out functional improvements in stance is meaningful, since it is difficult to test controller performance in able-bodied subjects under normal circumstances due to the high level of voluntary control. By facilitating the suppression of multiple sensory mechanisms vital for maintaining stability during stance (vestibular, visual, proprioceptive), the IPSA can be used with able-bodied subjects to mimic to some extent the impaired sensory systems observed in patient populations. Thus, important insights can be obtained whilst avoiding potential difficulties related to recruitment and heterogeneity in patient populations. It is, however, worth noting that use of the IPSA will not replace experimentation on patient populations, particularly given that it is impossible to completely remove voluntary control of muscles in able-bodied individuals, and because the IPSA does not account for some of the deficiencies observed in paraplegic musculoskeletal dynamics, such as diminished torque generating ability, rapid fatigue, and spasticity. Despite these limitations, our results verify the potential of the IPSA as a useful platform for working towards the development of a neuroprosthesis for standing.

It is also worth noting that use of the IPSA in patient

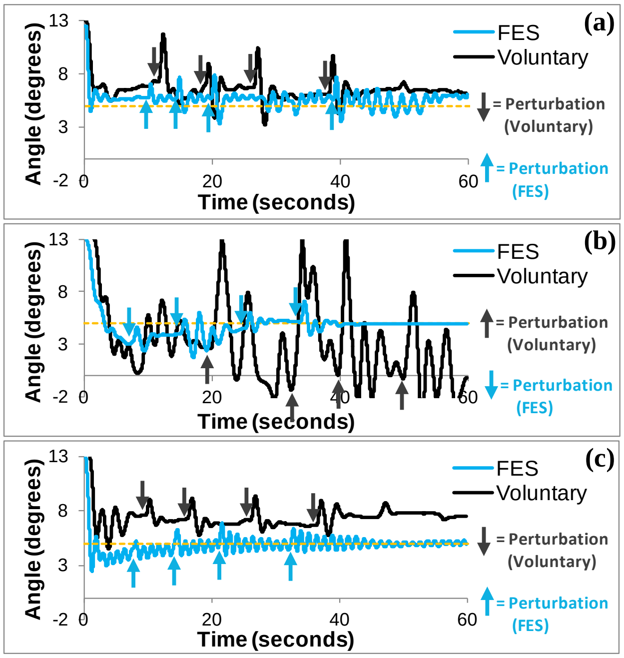

Figure 5. Inverted pendulum angles during 1-minute trials in which 4 posterior perturbations are applied at random times. The arrows represent the perturbation times. (a) Subject 1; (b) Subject 2; (c) Subject 3.

TABLE 2: COMPARISON OF PERFORMANCE WITH AND WITHOUT THE FES CONTROLLER ACTIVATED

Trial	Measure	Subject 1		Subject 2		Subject 3	
		FES[a]	*Vol*[b]	*FES*	*Vol*	*FES*	*Vol*
10-min quiet standing	Mean (deg)	4.6	3.9	5.0	5.7	5.0	7.6
	RMSD (deg)[c]	0.21	0.72	1.0	2.5	0.37	0.74
Offset initial position	Settling time (s)[d]	1.4	9.6	4.9	N/A[e]	2.0	N/A[e]
Perturbation	Mean settling time (s)[d]	1.0	2.7	2.0	N/A[e]	1.3	1.4
	Mean peak deviation (deg)	1.9	3.5	1.9	12	1.8	2.0

a. FES controller
b. Voluntary control
c. Root mean square difference
d. Settling time to within 1.5°
e. Did not to settle to within 1.5°

populations could offer important research and clinical benefits. In particular, since the IPSA allows patients to be fixed in an upright stationary position, safety concerns related to stability and the potential for falls during experimentation or rehabilitation would be effectively eradicated. Thus, the IPSA could be safely and easily utilized for testing control strategies in patient populations. Moreover, our results suggest that the IPSA, in combination with a PID control strategy, could potentially be utilized in a clinical capacity for safely retraining standing ability following incomplete SCI or stroke. Multiple studies have espoused functional and physiological benefits of therapeutic strategies utilizing FES for retraining standing and walking ability in these populations [22], [23]. Similarly, we believe that training sessions using the IPSA and a closed-loop FES system applied to the ankle muscles (using a PID or other type of control strategy) could offer therapeutic benefits by reinforcing requisite muscle activation patterns and encouraging motor learning, whilst also increasing muscle strength. However, this potential application remains to be

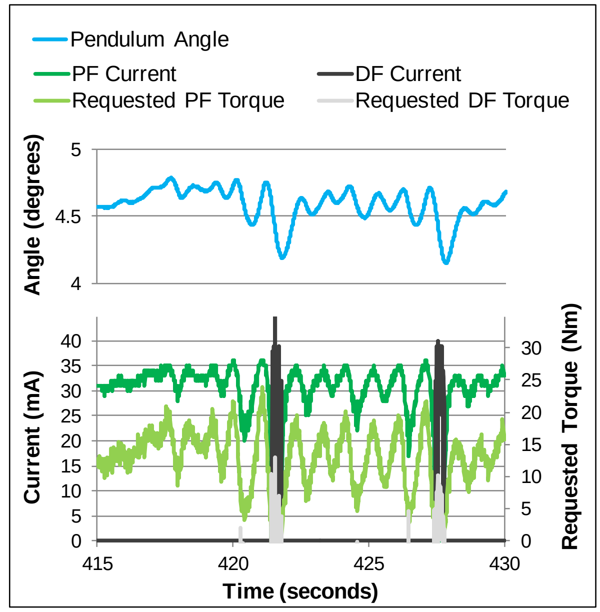

Figure 6. ***Top:*** Inverted pendulum angle during 15 seconds of the 10-minute quiet standing trial with the FES controller activated for Subject 1. ***Bottom:*** The corresponding plantarflexor and dorsiflexor currents, and requested plantarflexor and dorsiflexor torques (PF = plantarflexor; DF = dorsiflexor).

tested.

A promising aspect of the presented control strategy is that the activation pattern appears to match to an extent that observed during quiet able-bodied stance. In particular, as shown in Fig. 6, the ankle muscle activity is typically dominated by activation of the plantarflexors, with only intermittent dorsiflexor activation. This pattern is observed during quiet stance in able-bodied individuals in which the body is generally positioned approximately 5° in front of vertical, residual activity of the plantarflexors stops the body from falling forward, and the dorsiflexors are only briefly activated to counteract posterior sway [11]. Another notable characteristic of the PID control strategy is the small, periodic oscillations observed for prolonged periods in all 3 subjects. These oscillations, though distinct from and typically more pronounced than postural sway during able-bodied quiet stance, are not necessarily problematic. Kiemel et al [24], for example, established that the able-bodied control strategy during quiet stance is based on minimizing muscle activation rather than limiting small fluctuations of the COM. Thus, as long as these oscillations are around the reference angle and small in magnitude, such behavior is not considered detrimental to standing ability. Moreover, despite these oscillations, the maximum current amplitude typically remained considerably below the maximum allowable current amplitude of 60 mA. For example, the maximum current applied to the plantarflexors during the entire 10-minute trials (but after the initial 10-second period of transition from an offset initial position) were only 41 mA, 56 mA and 39 mA for Subjects 1, 2 and 3, respectively. Given that fatigue increases with increasing stimulation current, this suggests that fatigue may not rapidly become an issue, an important consideration given the rapid fatigability of paraplegic muscles. However, body weight matching experiments, in which the total weight of the inverted pendulum is equal to the weight of the subject, will give a clearer representation of fatigability.

The controller parameters found to be most appropriate for Subject 1 differed considerably from those established for Subjects 2 and 3. This suggests that inter-individual variability may need to be factored into controller design. However, given that controller performance was not overly sensitive to moderate changes in controller gains, it is possible that appropriate standardized parameters could be established independent of inter-subject variability. Further experimentation is required in order to investigate relationships between controller gains, controller performance, muscle recruitment dynamics, and subject demographics (e.g. age, height, weight).

For Subject 1, the controller strategy did not include integral components because the gravity compensation term along with appropriate selection of the other controller parameters, typically ensured that steady state errors were minimized. However, as can be observed from Figs. 3(a) and 5(a), the inverted pendulum angle often oscillated about angles slightly offset from the reference angle of 5^{o}. In contrast, in Subjects 2 and 3, the small integral components which were incorporated reliably eradicated any notable offset from the reference angle, suggesting that the integral component offers benefits. On the other hand, however, when considering the clinical objective of maintaining stable stance, small steady state errors may be considered less critical than maintaining stability in the presence of internally and externally generated perturbations. Therefore, integral terms should be incorporated with care.

In conclusion, we demonstrated using the IPSA the potential of a PID plus gravity control strategy to modulate FES levels applied to the ankle plantarflexors and dorsiflexors in closed loop, and to improve stability.

REFERENCES

[1] G. M. Yarkony, R. J. Jaeger, E. Roth, A. R. Kralj, J. Quintern, “Functional neuromuscular stimulation for standing after spinal cord injury,” *Arch Phys Med Rehabil,* vol. 71, pp. 201-206, 1990.

[2] M. de Haart, A. C. Geurts, S. C. Huidekoper, L. Fasotti, J. van Limbeek, “Recovery of standing balance in postacute stroke patients: a rehabilitation cohort study,” *Arch Phys Med Rehabil,* vol. 85, pp. 886-895, 2004.

[3] A. C. Geurts, G. M. Ribbers, J. A. Knoop, J. van Limbeek, “Identification of static and dynamic postural instability following traumatic brain injury,” *Arch Phys Med Rehabil,* vol. 77, pp. 639-644, 1996.

[4] S. Agarwal, R. J. Triolo, R. Kobetic, M. Miller, C. Bieri, S. Kukke *et al.*, “Long-term user perceptions of an implanted neuroprosthesis for exercise, standing, and transfers after spinal cord injury,” *J Rehabil Res Dev,* vol. 40, pp. 241-252, 2003.

[5] S. J. Harkema, C. K. Ferreira, R. J. van den Brand, A. V. Krassioukov, “Improvements in orthostatic instability with stand locomotor training in individuals with spinal cord injury,” *J Neurotrauma,* vol. 25, pp. 1467-1475, 2008.

[6] J. S. Walter, P. G. Sola, J. Sacks, Y. Lucero, E. Langbein, F. Weaver, “Indications for a home standing program for individuals with spinal cord injury,” *J Spinal Cord Med,* vol. 22, pp. 152-158, 1999.

[7] K. M. Bogie, R. J. Triolo, “Effects of regular use of neuromuscular electrical stimulation on tissue health,” *J Rehabil Res Dev,* vol. 40, pp. 469-475, 2003.

[8] H. R. Kobravi, A. Erfanian, “A decentralized adaptive fuzzy robust strategy for control of upright standing posture in paraplegia using functional electrical stimulation,” *Med Eng Phys,* vol. 34, pp. 28-37, 2012.

[9] H. Gollee, K. J. Hunt, D. E. Wood, “New results in feedback control of unsupported standing in paraplegia,” *IEEE Trans Neural Syst Rehabil Eng,* vol. 12, pp. 73-80, 2004.

[10] K. Masani, A. H. Vette, N. Kawashima, M. R. Popovic, “Neuromusculoskeletal torque-generation process has a large destabilizing effect on the control mechanism of quiet standing,” *J Neurophysiol,* vol. 100, pp. 1465-1475, 2008.

[11] K. Masani, M. R. Popovic, K. Nakazawa, M. Kouzaki, D. Nozaki, “Importance of body sway velocity information in controlling ankle extensor activities during quiet stance,” *J Neurophysiol,* vol. 90, pp. 3774-3782, 2003.

[12] T. D. Welch, L. H. Ting, “A feedback model reproduces muscle activity during human postural responses to support-surface translations,” *J Neurophysiol,* vol. 99, pp. 1032-1038, 2008.

[13] K. Masani, A. H. Vette, M. R. Popovic, “Controlling balance during quiet standing: proportional and derivative controller generates preceding motor command to body sway position observed in experiments,” *Gait Posture,* vol. 23, pp. 164-172, 2006.

[14] R. J. Peterka, “Sensorimotor integration in human postural control,” *J Neurophysiol,* vol. 88, pp. 1097-1118, 2002.

[15] R. J. Peterka, “Postural control model interpretation of stabilogram diffusion analysis,” *Biol Cybern,* vol. 82, pp. 335-343, 2000.

[16] A. H. Vette, K. Masani, K. Nakazawa, M. R. Popovic, “Neural-mechanical feedback control scheme generates physiological ankle torque fluctuation during quiet stance,” *IEEE Trans Neural Syst Rehabil Eng,* vol. 18, pp. 86-95, 2010.

[17] J. Y. Kim, M. R. Popovic, J. K. Mills, “Dynamic modeling and torque estimation of FES-assisted arm-free standing for paraplegics,” *IEEE Trans Neural Syst Rehabil Eng,* vol. 14, pp. 46-54, 2006.

[18] J. Y. Kim, J. K. Mills, A. H. Vette, M. R. Popovic, “Optimal combination of minimum degrees of freedom to be actuated in the lower limbs to facilitate arm-free paraplegic standing,” *J Biomech Eng,* vol. 129, pp. 838-847, 2007.

[19] A. H. Vette, K. Masani, M. R. Popovic, “Implementation of a physiologically identified PD feedback controller for regulating the active ankle torque during quiet stance,” *IEEE Trans Neural Syst Rehabil Eng,* vol. 15, pp. 235-243, 2007.

[20] J. F. Tan, J. Zariffa, A. H. Vette, C. L. Lynch, K. Masani, M. R. Popovic, “Use of an inverted pendulum apparatus for the study of closed-loop FES control of the ankle joints during quiet standing,” *Proc. 15th Ann. Conf. of the Int. Functional Electrical Stimulation Society,* vol., pp. 118-120, 2010.

[21] K. Masani, D. G. Sayenko, A. H. Vette, “What triggers the continuous muscle activity during upright standing?,” *Gait Posture,* vol. 37, pp. 72-77, 2013.

[22] M. H. Granat, A. C. Ferguson, B. J. Andrews, M. Delargy, “The role of functional electrical stimulation in the rehabilitation of patients with incomplete spinal cord injury--observed benefits during gait studies,” *Paraplegia,* vol. 31, pp. 207-215, 1993.

[23] A. R. Lindquist, C. L. Prado, R. M. Barros, R. Mattioli, P. H. da Costa, T. F. Salvini, “Gait training combining partial body-weight support, a treadmill, and functional electrical stimulation: effects on poststroke gait,” *Phys Ther,* vol. 87, pp. 1144-1154, 2007.

[24] T. Kiemel, Y. Zhang, J. J. Jeka, “Identification of neural feedback for upright stance in humans: stabilization rather than sway minimization,” *J Neurosci,* vol. 31, pp. 15144-15153, 2011.