Wheelchair Neuroprosthesis for Improving Dynamic Trunk Stability

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***Abstract*—Trunk instability is a major problem for individuals with thoracic and cervical spinal cord injury. Functional electrical stimulation (FES) neuroprosthesis, a technology that uses small electrical currents to artificially contract muscles, has previously been utilized to improve trunk stability during quasi-static and dynamic sitting. The aim of this study was to develop the first powered wheelchair-based neuroprosthesis and to test its feasibility for improving trunk stability. Eleven male, able-bodied individuals participated in the feasibility study. While participants were seated, the wheelchair was moved in the forward or backward directions with slow and fast accelerations. Two different FES protocols were tested: 1) co-contraction; and 2) directionally-dependent contraction of trunk extensors and flexors. Sham stimulations with intensities below the motor threshold were applied as the control conditions. Inertial motion sensors were used to quantify the maximum angular displacement and velocity of the trunk. Results showed that both directional contractions and co-contraction reduced trunk displacement and velocity, compared to the control conditions. However, directionally-dependent muscle contractions were more effective in improving trunk stability, compared to co-contractions. Overall, feasibility of the wheelchair-based neuroprosthesis was demonstrated. Future research will incorporate feedback from wheelchair movements and test the neuroprosthesis with individuals who sustained spinal cord injury.**

***Index Terms*—Trunk Stability, Sitting Balance, Functional Electrical Stimulation, Neuroprosthesis, Inertial Motion Sensor.**

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I. Introduction

INDIVIDUALS with thoracic and cervical spinal cord injury (SCI) often suffer from trunk instability because of motor and/or sensory impairment of the trunk muscles [1, 2]. This instability leads to a number of functional limitations that negatively affect one's ability to perform activities of daily living, and could lead to numerous secondary health complications such as pressure sores, kyphosis, pulmonary dysfunction and more [3].

Functional electrical stimulation (FES), which uses electrical impulses to artificially contract muscles, is a promising technology applicable to improving trunk stability. Over the past decade, FES has been utilized to improve seated trunk stability in able-bodied individuals and those with SCI [3-8]. Early work in this field focused on using FES to improve trunk stability in static and quasi-static conditions such as during quiet sitting [3, 5], voluntary reaching tasks [3, 6], and manual wheelchair propulsion [4, 7]. These studies primarily utilized tonic co-contraction of the trunk muscles in an effort to increase trunk stiffness, and were largely successful in improving seated posture (i.e., spinal alignment and pelvic orientation) [3], increasing bimanual workspace [6], increasing maximum propulsive forces during manual wheelchair use [4, 7], and reactive postural stability against external perturbations [3]. Although these methods were effective in stabilizing the trunk, a strategy using co-contractions is not ideal for extended periods of usage because it can lead to rapid muscle fatigue, which is a major concern with all FES neuroprostheses [8]. Moreover, trunk muscles are not always co-contracted during sitting. Trunk muscles are weakly co-contracted to keep quiet upright sitting posture, while the trunk uses phasic and direction-dependent activation of trunk muscles following external perturbations during sitting balance [9, 10]. Specifically, following support-surface translation perturbations in the forward direction, trunk flexors are activated during the acceleration phase and trunk extensors are activated during the deceleration phase, and vice-versa during the backward perturbations [10]. Therefore, direction-dependent activation of muscles may be an effective strategy used to control seated balance during external balance perturbations.

Recent studies using biomechanical and musculoskeletal models of the trunk have allowed researchers to develop and investigate the feasibility of closed-loop FES controllers by monitoring trunk movement [11-13]. Vanoncini and colleagues [8, 11] performed one of the first such studies, in which they used a proportional-integral-derivative (PID) and a linear quadratic regulation controller to stabilize the trunk using FES activation of trunk muscles. They demonstrated the feasibility of a closed-loop system for stabilizing the trunk. Murphy and colleagues [14] then demonstrated the feasibility of a closed-loop FES system for improving seated posture in individuals with SCI. Audu and colleagues [15] recently expanded on this, and developed a feedback-controlled neuroprosthesis for improving seated balance against external perturbations. Their system used a proportional-derivative (PD) controller to modulate the force generated by various trunk extensors using FES, and it was capable of returning the trunk of individuals with SCI to an erect posture following forward pulling perturbations with force magnitudes of up to 45% of the body weight [15]. Overall, these are encouraging results in support of development of closed-loop controllers for improving dynamic seated trunk stability.

Due to complexities and various technical challenges required to implement closed-loop FES systems for use in patient's daily lives, we aimed to develop a more practical and applicable neuroprosthesis for sitting balance. Individuals with cervical SCI typically spend most of their daily life in powered wheelchairs. As such, the timing and direction of the perturbations caused by the wheelchair movement, which can displace the patient's body, can be anticipated by monitoring the wheelchair joystick. Thus, it is possible to contract the trunk muscles before the perturbation caused by acceleration and deceleration of a powered wheelchair, in a feed-forward manner. In physiological investigations, it has been shown that pre-activation of muscles can reduce the amount of feedback trunk muscle responses after the perturbation [16]. Therefore, direction-dependent and feed-forward activation of trunk muscles using an FES neuroprosthesis should improve trunk stability during powered wheelchair movements.

We hypothesized that a neuroprosthesis that activates trunk muscles using FES in a direction-dependent and feed-forward manner could be used to improve dynamic trunk stability during acceleration and deceleration of a powered wheelchair. The objectives of this study were to: 1) develop a powered wheelchair-based FES neuroprosthesis that can activate trunk muscles before the wheelchair moves in the forward and backward directions; and 2) evaluate the feasibility of the neuroprosthesis to improve trunk stability by comparing co-contractions and directionally-dependent contractions of trunk muscles.

II. Methods

A. Participants

Eleven healthy, male, able-bodied individuals participated in this study. The age, weight and height of participants were 21.3±3 years, 72.9±11 kg and 177.7±9 cm (mean ± SD), respectively. None of the participants had a history of neurological or musculoskeletal impairments. Informed consent was obtained from all individuals in accordance with the guidelines of the Declaration of Helsinki. The experimental procedures were approved by the institutional Research Ethics Board.

B. Powered Wheelchair Neuroprosthesis

An FES neuroprosthesis was engineered by interfacing a powered wheelchair (655FS, Fortress Scientific, USA) with a microcontroller (Arduino Uno, Arduino, Italy), which communicated to a control and data acquisition computer running a custom application software developed in LabVIEW (NI 6255, National Instruments, USA). The microcontroller was connected to a high-power motor driver (VNH5019, Pololu Robotics and Electronics, USA) to control the wheelchair motors, and with two FES stimulators (Compex Motion II, Compex, Switzerland) to provide simultaneous independent, bilateral electrical stimulation to the trunk muscles. The system is also capable of outputting five analog signals that were used for triggering and synchronizing the data collection system. Figure 1 shows a flow chart overview of the system.

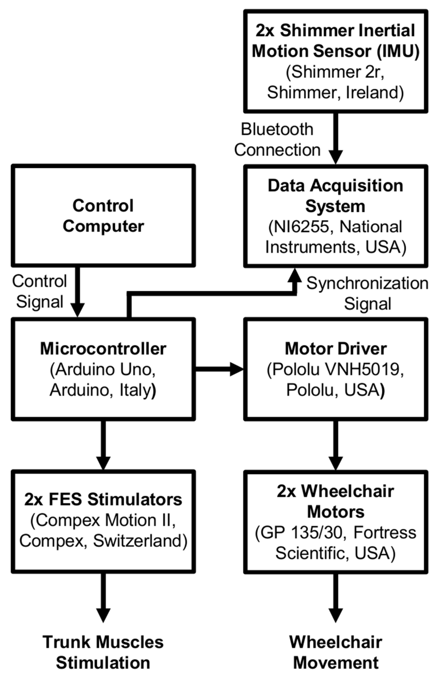

Fig. 1. A flow chart showing the wheelchair neuroprosthesis design.

C. Experimental Protocol

Participants were seated in the wheelchair with arms crossed on their chest, and were instructed to maintain a relaxed upright posture (Figure 2A). Their feet and thighs were secured to the footrest and the seat to isolate movements of the trunk. During the experiment, perturbations were controlled by the experimenter. Perturbations were delivered to the participants in the forward and backward directions, and at two different acceleration magnitudes, corresponding to 20% (slow perturbations) and 40% (fast perturbations) of the body weight (Table 1). This created four perturbation conditions: i) slow-forward; ii) slow-backward; iii) fast-forward; and iv) fast-backward. Each perturbation consisted of

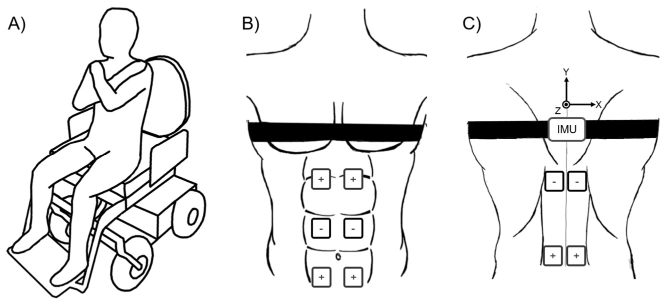

Fig. 2. A) Experimental setup for sitting balance experiment showing the subject sitting in the wheelchair. B) Anterior view of the trunk showing the placement of the stimulating (+) and ground (-) electrodes on the rectus abdominus muscles. C) Posterior view of the trunk showing the placement of the electrodes on the erector spinae muscles, and the placement and orientation of the inertial motion sensor (IMU) at the T5 vertebrae level on the spine.

an acceleration phase and a deceleration phase, producing a triangle-shaped velocity profile. A short auditory tone was played before the start of each perturbation to signal the next trial, but it was played at a random interval between 1-2 seconds prior to the perturbation such that participants would not anticipate the exact time. After each perturbation, the wheelchair was slowly moved back to the starting position, before the subsequent trial commenced approximately 5 seconds later.

Each perturbation was accompanied with one of five experimental conditions: 1) no-FES (NF): muscles were not stimulated; 2) sham-FES co-contraction (CF^-): trunk flexor and extensor muscles were stimulated simultaneously with the below motor threshold intensity; 3) sham-FES directional (DF^-): trunk flexor and extensor muscles were stimulated in a direction-dependent manner with the below motor threshold intensity; 4) full-FES co-contraction (CF^+): trunk flexor and extensor muscles were stimulated simultaneously to generate functional contractions; and 5) full-FES directional (DF^+): trunk flexor and extensor muscles were stimulated in a direction-dependent manner to generate functional contractions. The NF condition served as the primary negative control, and the CF^- and DF^- conditions served as secondary negative controls to take into account the additional voluntary and/or involuntary reactions to FES stimulation in able-bodied participants.

The four perturbation conditions combined with five experimental conditions resulted in a total of 20 conditions. Each of these conditions was repeated 16 times for a total of 320 perturbations. The perturbations were pseudo-randomized and divided equally into four experimental units, each separated by a 10 minute rest period, to minimize fatigue. The first five trials of each unit were used to familiarize the participants, and were not analyzed. As a result, 15 repeated trials of each condition were analyzed for each participant.

TABLE I
AVERAGE PEAK ACCELERATION AMPLITUDES (M/S^2) OF THE WHEELCHAIR DURING PERTURBATIONS

	Slow		**Fast**	
	Forward	Backward	Forward	Backward
Acceleration	2.14 ± 0.48	1.95 ± 0.34	4.26 ± 0.44	3.54 ± 0.50
Deceleration	1.23 ± 0.26	1.05 ± 0.20	3.19 ± 0.40	3.42 ± 0.44

D. Functional Electrical Stimulation (FES)

Stimulation was applied on the rectus abdominus (RA) (trunk flexor) and erector spinae (ES) (trunk extensor) muscles bilaterally, as they have been shown to contribute significantly to trunk stability during anterior-posterior perturbations [5, 9, 10, 17]. FES was delivered using the Compex Motion Stimulator (Compex Motion II, Compex, Switzerland) with biphasic, rectangular, asymmetric charge-balanced stimulation pulses at 40Hz with a with a 300μsec pulse duration [5, 18]. Two self-adhesive stimulating electrodes (5cm x 5cm) were placed bilaterally on two motor points of the RA muscles, with a common ground electrode placed in between, to achieve contraction of the entire muscle (Figure 2B). One stimulating electrode was placed bilaterally on the lower motor point of the ES muscle, with the ground electrode positioned to obtain contraction of the entire muscle (Figure 2C). The locations of the targeted motor points were identified using recommendations by Behringer and colleagues [18] and the exact location was pinpointed using a probe electrode.

The motor threshold for each muscle was identified by gradually increasing the stimulation amplitude with 1mA increments until palpable contractions were identified. The maximum tolerable stimulation was also determined. The stimulation amplitudes for the sham-FES conditions (DF^- and CF^-) were set to 2mA below the motor threshold. The stimulation amplitudes for the full-FES conditions (DF^+ and CF^+) were set to three-quarters of the way between motor threshold and maximum tolerable stimulation. The average sham stimulation amplitude was 6.5±2.6 mA for the RA and 11.9±3.0 mA for the ES muscle. The average full stimulation amplitude was 18.5±4.9 mA for the RA and 33.2±5.5 mA for the ES muscle.

In the directional FES trials (DF^+ and DF^-), FES was applied to one set of muscles during the acceleration phase, and another set of muscles during the deceleration phase [10]. FES was activated 100ms prior to the movement of the wheelchair. Specifically, during the forward perturbations, the RA muscles were stimulated during the acceleration phase, starting 100ms before the onset of the perturbation and relaxed at the onset of the deceleration phase of the perturbation. The ES muscles were stimulated 100ms before the start of the deceleration phase and relaxed 1 second after (end of the perturbation). Similarly, during the backward perturbations, the ES muscles were stimulated during the acceleration phase and the RA muscles were stimulated during the deceleration phase. In the co-contraction FES trials (CF^+ and CF^-), both the RA and ES muscles were stimulated during both the acceleration and deceleration phase, starting 100ms before the start of the perturbation and relaxed 1 second after the onset of the deceleration phase.

E. Data Collection

Two inertial motion sensors (Shimmer 2r, Shimmer, Ireland) were used to collect data about the movement of the wheelchair and the movement of the trunk [20-23]. Each sensor contained a 3-axis accelerometer, a 3-axis gyroscope and a 3-axis magnetometer. One sensor was mounted on the frame of the wheelchair in order to quantify the perturbation acceleration/deceleration profiles [24]. Another sensor was mounted on the participants' back at the T5 vertebrae level to approximate the movements of the trunk [15, 24] (Figure 2C). Data was sampled at 204.8Hz and streamed wirelessly over a Bluetooth connection to a custom application on the control computer, which was developed in LabVIEW software (National Instruments, USA).

F. Data Analysis

Trunk stability measures were calculated during post-processing. All data was filtered using a low-pass Butterworth filter (4th order, zero-phase-lag) with a cut-off frequency of 50Hz [10]. The filtered 3-axis accelerometer and the gyroscope signals were then used to calculate the orientation of the sensor mounted on the back of the participants, using a sensor-fusion algorithm [20, 22, 25].

Using the algorithm, the orientation of the Earth frame relative to the sensor frame $({}^S_E q_{est}(t))$ at time t, which is shown in Equation 1, was calculated using Equations 2 and 3 [25]. A quaternion representation of orientation is used in the algorithm (Equation 1) in which orientation is defined as a four dimensional vector consisting of an instantaneous axis of rotation (defined by q_2, q_3, and q_4) and a magnitude of rotation around this axis (q_1). This approach allows the algorithm to avoid the kinematic singularities present in the Euler angle representation of orientation.

$$({}^S_E q_{est}(t)) = [q_1 \;\; q_2 \;\; q_3 \;\; q_4] \tag{1}$$

$$\left.\begin{aligned} {}^S_E q_{est}(t) &= {}^S_E q_{est}(t-1) + {}^S_E \dot{q}_{est}(t)\Delta t \\ {}^S_E \dot{q}_{est}(t) &= {}^S_E \dot{q}_{\omega}(t) - \beta \frac{\nabla f}{\|\nabla f\|} \end{aligned}\right\} \tag{2}$$

where

$$\left.\begin{aligned} \nabla f({}^S_E q, S_a) &= J^T({}^S_E q_{est}(t-1)) f({}^S_E q_{est}(t-1), S_a) \\ S_a &= [0, a_x, a_y, a_z] \end{aligned}\right\} \tag{3}$$

In Equations 2 and 3, ${}^S_E \dot{q}_{est}(t)$ represents the estimated rate of change of the orientation at time t; ${}^S_E \dot{q}_{\omega}(t)$ represents the rate of change of orientation measured directly by the gyroscope; β represents the gyroscope measurement error in a direction based on the accelerometer measurements $\left(\frac{\nabla f}{\|\nabla f\|}\right)$; and S_a is a quaternion representing the direction the measured acceleration due to gravity. The algorithm calculates the estimated orientation of the Earth frame relative to the sensor frame at time t $({}^S_E q_{est}(t))$, by integrating the estimated rate of change of orientation $({}^S_E \dot{q}_{est}(t))$ and adding it to the estimated orientation at time $t-1$ $({}^S_E q_{est}(t-1))$. f is an objective function that is minimized using a gradient descent algorithm in order to find a single solution for the sensor orientation, using the fact that the direction of acceleration due to gravity in the Earth frame is homogenous and known. J is the Jacobean of this objective function. f and J are defined in Equations 4 and 5 respectively [25].

$$f({}^S_E q, S_a) = \begin{bmatrix} 2(q_2 q_4 - q_1 q_3) - a_x \\ 2(q_1 q_2 + q_3 q_4) - a_y \\ 2\left(\frac{1}{2} - q_2{}^2 - q_3{}^2\right) - a_z \end{bmatrix} \tag{4}$$

$$J({}^S_E q_{est}(t-1)) = \begin{bmatrix} -2q_3 & 2q_4 & -2q_1 & 2q_2 \\ 2q_2 & 2q_1 & 2q_4 & 2q_3 \\ 0 & -4q_2 & -4q_3 & 0 \end{bmatrix} \tag{5}$$

Here a_x, a_y and a_z are the three components of the normalized acceleration vector that is measured by the IMU. The estimated sensor orientation at time t, calculated using this algorithm has levels of accuracy similar to that of other widely utilized Kalman filter based algorithms, but at a low computational expense [25].

To further calculate angular displacement and velocity, the orientation is converted to a rotation matrix $({}^E_S R(t))$ at time t, which can transform vectors from the sensor frame of reference to the Earth frame of reference. This rotation matrix is used to transform a unit vector along the z-axis of the sensor $(z^S(t))$ to a vector in the Earth frame of reference $(z^E(t))$ (Equation 6). This vector, representing the z-axis of the sensor at time t in the earth frame of reference is transformed into a vector in the frame of reference of the sensor at time 0, $z^{S,t=0}$ (Equation 7). The resulting vector is projected on the y-z plane of the sensor (corresponding to the anterior-posterior plane) and used to calculate the angular displacement of the sensor in the anterior-posterior plane at time $(\theta(t))$, with respect to the orientation at time 0 (Equation 8). Therefore, this angular displacement is zeroed with respect to the orientation of sensor at the start of each perturbation.

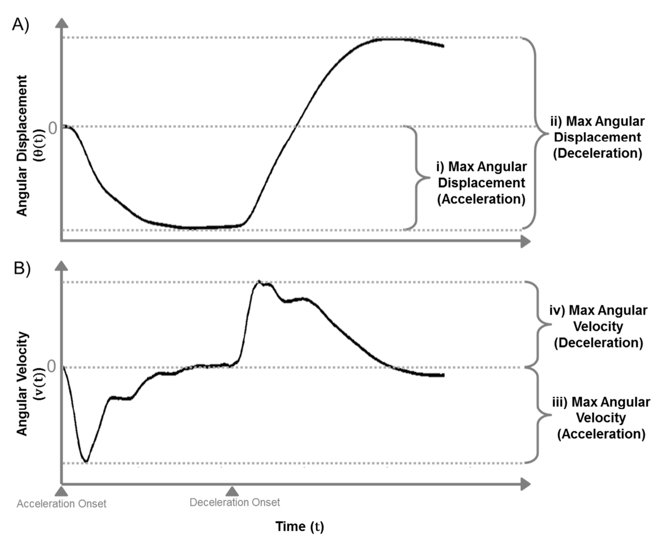

Fig. 3. A) Representative angular displacement profile for a perturbation in the forward direction. The displacement is zeroed at the start of the perturbation: *i)* maximum angular displacement during the acceleration phase of the perturbation; and *ii*) maximum angular displacement during the deceleration phase of the perturbation. B) Representative angular velocity profile for a perturbation in the forward direction. *iii*) maximum angular velocity during the acceleration phase of the perturbation. *iv)* maximum angular velocity during the deceleration phase of the perturbation.

$$\left.\begin{aligned} z^S(t) &= \begin{bmatrix} 0 \\ 0 \\ 1 \end{bmatrix} \\ z^E(t) &= {}^E_S R(t) \times z^S(t) \end{aligned}\right\} \tag{6}$$

$$z^{S,t=0}(t) = {}^E_S R(0)^T \times z^E(t) \tag{7}$$

$$if\ z^{S,t=0}(t) = \begin{bmatrix} x \\ y \\ z \end{bmatrix}, then\ \theta(t) = atan2(y, z) \tag{8}$$

$$v(t) = \dot{\theta}(t) \tag{9}$$

This angular displacement signal was then differentiated to calculate the angular velocity signal for each trial (Equation 9).

From the angular displacement (Equation 8) and angular velocity (Equation 9) signals, the maximum angular displacement and maximum angular velocity were calculated during the acceleration and deceleration phases of the perturbations, as shown in Figure 3. These four measures formed the primary analysis metrics used to characterize dynamic trunk stability.

G. Statistics

Comparisons between the experimental conditions (i.e., NF, CF^-, DF^-, CF^+ and DF^+) were performed using the repeated measure one-way analysis of variance (ANOVA) with LSD post-hoc multiple comparisons when a significant difference was found. Prior to the ANOVA, all measures were tested using the Shapiro-Wilk test to identify if they were normally distributed. Since it was shown that all measures were not normally distributed, a logarithmic transformation was performed to normalize the data before performing the ANOVA analysis. Significance level was set to $p < 0.05$.

III. Results

The results for the maximum angular displacement are shown in Figure 4. The displacement in the control conditions, including no-FES condition (NF) and sham-FES conditions (CF^- and DF^-), was not different. During the acceleration phase, displacement in the full-FES directional condition (DF^+) was significantly lower than the controls at both perturbation magnitudes, in both directions, whereas the full-FES co-contraction condition (CF^+) was significantly lower from the controls in the backward perturbations at both perturbation magnitudes. During the deceleration phase, displacement in the DF^+ condition was significantly lower than the controls at both perturbation magnitudes, in both directions, whereas the CF^+ condition was only significantly lower than the controls during the fast perturbations in the forward and backward directions. Displacement in the DF^+ condition was also significantly lower than the CF^+ condition during the deceleration phase of all perturbations and during the acceleration phase of the slow-forward perturbations.

The results for maximum angular velocity are shown in Figure 5. The velocity in the control conditions was not different. During the acceleration phase, maximum angular velocity in the DF^+ condition was significantly lower than the

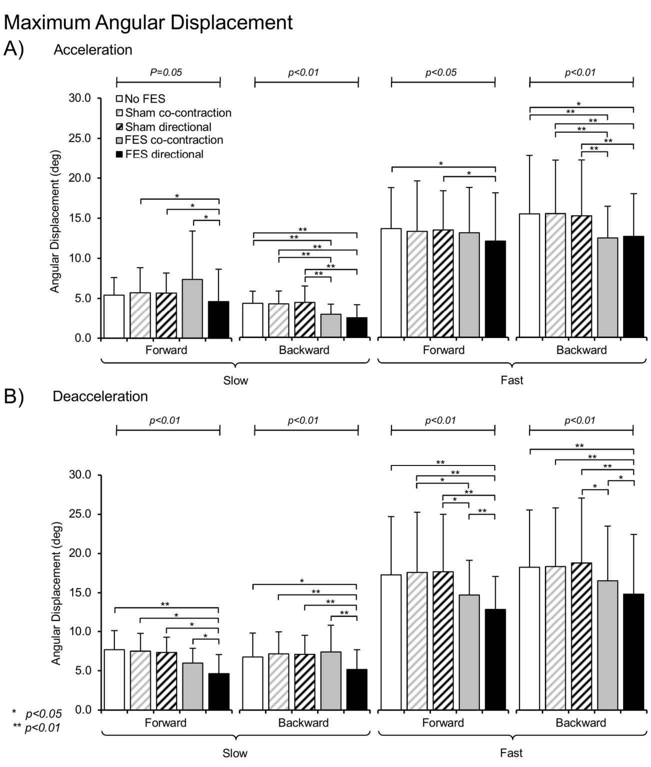

Fig. 4. Maximum angular displacement during: A) acceleration; and B) deacceleration phase of the wheelchair movement for the forward and backward perturbations during slow and fast wheelchair movements. Shown are the mean + standard deviation (SD) for each experimental condition. Note: *p < 0.05 and **p < 0.01.

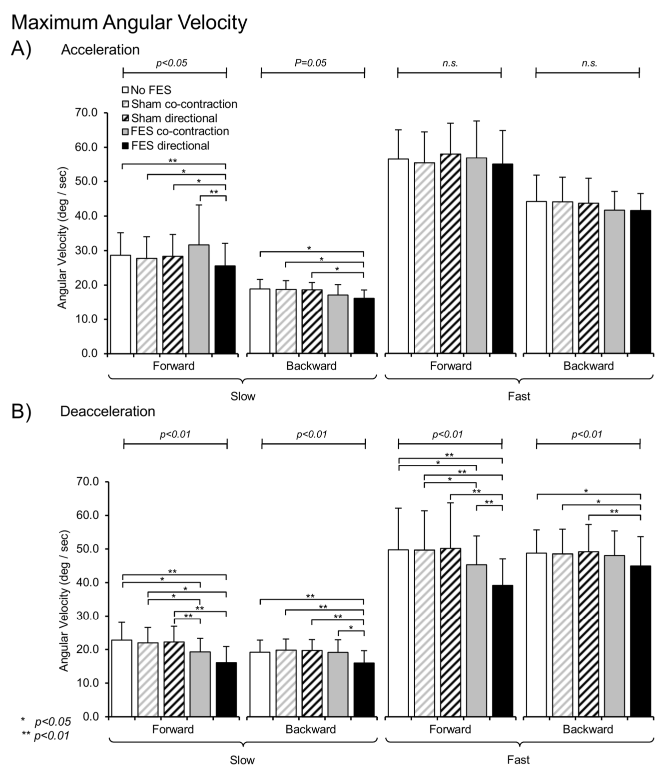

Fig. 5. Maximum angular velocity during: A) acceleration; and B) deacceleration phase of the wheelchair movement for the forward and backward perturbations during slow and fast wheelchair movements. Shown are the mean + standard deviation (SD) for each experimental condition. Note: *p < 0.05 and **p < 0.01.

controls in the slow perturbations in both directions, whereas the CF^+ condition was not significantly different than the controls in any of the perturbations. During the deceleration phase of the perturbations, maximum angular velocity in the DF^+ condition was significantly lower than the controls in both perturbation magnitudes and perturbation directions, whereas the CF^+ condition was significantly lower than the controls only in the forward perturbations, at both perturbation magnitudes. Velocity in the DF^+ condition was significantly lower than in the CF^+ condition during the deceleration phase of slow-backward and fast-forward perturbations, and during the acceleration phase of the slow-forward perturbations.

IV. Discussions

In this study, a powered wheelchair-based neuroprosthesis was developed, which was capable of contracting trunk muscles using FES in a direction-dependent and feed-forward manner. This system, unlike other trunk FES neuroprosthesis systems [8, 15], is the first practical sitting balance neuroprosthesis. In the experiment, we evaluated the feasibility of this neuroprosthesis to improve trunk stability during wheelchair movements. Overall, we demonstrated that the neuroprosthesis reduced trunk displacement and velocity compared to the control trials, and that direction-dependent stimulations seemed more effective compared to the co-contractions.

A. Effects of FES on Trunk Stability

Our results indicate that directional activations using FES to generate functional contractions of the trunk muscles (DF^+) were able to reduce displacement and velocity of the trunk during the acceleration and deceleration phase of the perturbations, compared to the control conditions. Although co-contraction of muscles using FES (CF^+) was also effective in reducing displacement of the trunk, compared to the control conditions, in many circumstances directional activations were more effective than co-contractions.

Large and rapid displacements of the trunk could result in in falls from the chair and potential risks of injury, especially in individuals with SCI [15, 24]. Our results suggest that the trunk was displaced less and with a lower peak velocity when the FES neuroprosthesis was used, which implies improved trunk stability during sitting balance with FES activation of the trunk muscles. Previous studies showed that co-activation of trunk muscles with FES can improve clinical outcomes during sitting balance, including improved spinal alignment and pelvic orientation [3], increased bimanual workspace [6], and improved manual wheelchair propulsion [4, 7]. Moreover, closed-loop controlled FES of the trunk muscles was capable of maintaining trunk posture during reaching [14] and following external perturbations [15] in individuals with SCI. Our study is the first to show that co-contraction (i.e., tonic activation) and directional contraction (i.e., phasic activation) of trunk muscles in a feed-forward manner using FES can improve stability of the trunk during support surface translations. These improvements in sitting balance with FES activation of the trunk muscles are likely due to the increased trunk stiffness [5, 18].

Sham-FES conditions (DF^- and CF^-) were included to take into account the effects of voluntary and/or reflexive muscle responses of the able-bodied participants to FES stimulation. Our results also showed that trunk stability during sham-FES conditions (DF^- and CF^-) was not significantly different from when FES was not utilized during perturbations (NF). Sham-FES provided sensory stimulation before and during the perturbations but did not actually contract the muscles (i.e., stimulations were below the motor threshold intensity). This result suggests that the application of the sensory stimulus did not lead to any additional voluntary and/or involuntary anticipatory reaction in able-bodied participants [5]. Furthermore, the concordance of statistical significance between the test conditions and, the primary and secondary negative controls suggests that the improvements in trunk stability with FES were the result of FES-induced muscle contractions. Sensory stimulation (i.e., afferent input) induced by electrical stimulation could be an effective way to induced spinal plasticity in individuals with incomplete SCI [26]. Therefore, neuroprosthesis should aim to activate both sensory and motor pathways to induce effective neurophysiological changes and facilitate or restore motor function in individuals with incomplete SCI [27]. Our current study demonstrated the biomechanical effectiveness of the powered wheelchair-based neuroprosthesis to improve trunk stability, which was due to the forces generated by FES-induced activation of trunk muscles. Such stimulations have the potential to induce spinal plasticity, but this remains to be investigated in the future studies.

B. Direction-Dependent and Feed-Forward Control of Trunk Muscles

Our results indicate that directional activations (DF^+) were more effective in stabilizing the trunk than co-contractions (CF^+). Although it has been shown that trunk stiffness can be increased by co-contraction of trunk flexors and extensors [5, 18] to improve sitting balance [3, 4, 6, 7], such simultaneous contraction also generates opposing torques that can cancel each other out and lead to an ineffective, resistive torque generation. On the other hand, during directional activations, only muscles that oppose the direction of the perturbation are stimulated, generating only a corrective torque on the trunk, which seems more effective for reducing trunk displacement. It has previously been shown that the central nervous system utilizes such phasic activation patterns in response to support-surface translation perturbations [10]. Our results suggest that a physiologically inspired control strategy using FES to activate the muscles improves biomechanical stability of the trunk more effectively than co-contractions, which are not necessarily a natural central nervous system control strategy.

In both directional and co-contractions protocols, trunk muscles were activated 100ms prior to the onset of the perturbation to achieve maximum force generation at the start of the perturbation [28, 29]. Previously, electromechanical delays have been reported in the range of 50 ms [28] and in order to account for the neuromuscular torque generating process during FES-activation of muscles [29], the neuroprosthesis had to stimulate muscles in a feed-forward manner, prior to the displacement of the body. Moreover, pre-activation of trunk muscles was shown to decrease the reactive, feedback-driven responses [16]. Such stimulation may also be advantageous, especially among individuals with SCI who have distorted and delayed motor control [6]. Furthermore, using such phasic, direction-dependent stimulation of muscles before and during the perturbation also has the potential to reduce muscle fatigue by minimizing unnecessary stimulation durations [8, 18].

C. Limitations and Future Work

The average acceleration amplitudes of the wheelchair were slightly higher during the acceleration phase and during fast movements, compared to deceleration and slow movements (Table 1). Although the movements were within range of typical powered wheelchair motion profiles, our results showed that the neuroprosthesis was slightly more effective during the deacceleration phase and during slow movements. This may be the result of insufficient torque generation by the FES during faster accelerations since the same stimulation amplitudes were used throughout. Future studies should consider a perturbation amplitude-dependent stimulation protocol to improve the performance of the wheelchair neuroprosthesis. Another potential way to improve the performance of the neuroprosthesis in the future may be to combine weak tonic co-contractions during quiet sitting, and larger phasic, directional activations in response to perturbations, which mimics the physiological control strategy of the trunk [9, 10, 17]. Finally, some of the effects during the perturbations in all experimental conditions could be attributed to voluntary and reflexive muscle responses in able-bodied individuals, which is why further testing with individuals who sustained SCI is warranted.

V. Conclusions

In this study, a powered wheelchair-based neuroprosthesis was developed, which was capable of contracting trunk muscles using FES in an effort to improve trunk stability during wheelchair movements. Compared to the control conditions, both directional activations and co-contractions of trunk muscles using FES were able to improve trunk stability during anterior-posterior accelerations and decelerations of the wheelchair, which produced perturbations with magnitudes of up to 40% of the body-weight. Directional activations were more effective in improving trunk stability compared to co-contraction of trunk flexors and extensors, suggesting that physiologically inspired FES activations are a better control strategy for FES neuroprosthesis. Overall, our study developed the first powered wheelchair neuroprosthesis for sitting balance and demonstrated the feasibility of using it to improve trunk stability. Such a neuroprosthesis could allow individuals with thoracic and cervical SCI to maintain sitting balance despite mild perturbations when these perturbations can be anticipated (e.g., by monitoring the wheelchair joystick when a patient accelerates and deaccelerates the wheelchair). However, further investigations are needed to prove the effectives in individuals with SCI.

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