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Low-intensity functional electrical stimulation can increase multidirectional trunk stiffness in able-bodied individuals during sitting

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ABSTRACT

The inability to voluntarily control the trunk musculature is a major problem following spinal cord injury as it can compromise functional independence and produce unwanted secondary complications. Recent developments suggest that neuroprostheses utilizing functional electrical stimulation (FES) may be able to facilitate or restore trunk control during sitting, standing, and other tasks involving postural control. In spite of these efforts, no study to date has used low-intensity FES to increase multidirectional trunk stiffness and damping in an attempt to bolster stability while minimizing muscle fatigue. Therefore, we set out to investigate how multidirectional trunk stiffness changes in response to low-intensity FES of a few selected trunk muscles. Fifteen healthy participants sitting naturally were randomly perturbed in eight horizontal directions. Trunk stiffness and damping during natural and FES-supported sitting conditions were quantified using force and trunk kinematics in combination with two models of a mass-spring-damper system. Our results indicate that low-intensity FES can increase trunk stiffness in healthy individuals, and this specifically for directions associated with the stimulated muscles. In contrast, trunk damping was not found to be altered during FES. The presented results suggest that low-intensity FES is a simple and effective method for increasing trunk stiffness on demand.

1. Introduction

The inability to voluntarily control the trunk musculature is a major problem for individuals with spinal cord injury (SCI). Any injury to the spinal cord between the head and the tenth thoracic vertebra (T10) can cause some degree of trunk function impairment due to the loss or mutilation of respective sensorimotor information [1]. Individuals with SCI who do experience compromised sensorimotor control of the trunk muscles are typically unable to regulate sitting balance on their own, resulting in a loss of functional abilities and independence during activities of daily living (ADL) [2,3]. Moreover, trunk instability is a primary cause of respiratory dysfunction due to isovolumic changes in the ribcage and abdominal compartment configuration [4,5]. Therefore, rapid and optimal improvement of trunk control is of high priority for affected individuals, outweighing their desire, for example, to walk again [6].

In order to compensate for insufficient muscle control and maintain equilibrium during sitting, individuals with SCI tilt the pelvis further backward than able-bodied individuals, allowing them to increase the level of stability in the anterior direction [7,8]. When reaching, they oftentimes use one arm on their lap or thrown over the back of their chair to provide the forces for keeping the trunk from bending forward uncontrollably. As a consequence, it is almost impossible to carry out bilateral hand and arm reaching tasks. Larger or heavier objects generally have to be placed conveniently in the lap or on a close table before they can be manipulated safely using both arms. In addition, the described compensational sitting arrangements can lead to kyphosis [8–10] and pressure sores [11] caused by asymmetric trunk orientation and infrequent weight distribution.

Various efforts have attempted to improve sitting stability of individuals with SCI during forward-reaching, primarily by customizing wheelchair configurations. These include modified tilt and reclination angles [12], novel types of seat cushions [13], and the use of footrests [14] or chest straps [15]. Furthermore, recent developments in the field of neurorehabilitation suggest that also neuroprostheses utilizing functional electrical stimulation (FES) may have the potential to facilitate or even restore trunk control during sitting and other functional tasks. For example, FES has been used in open-loop control

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schemes to activate the paralyzed trunk musculature during sitting to increase seated postural stability [16,17] and facilitate bimanual tasks that individuals with SCI are otherwise unable to complete [16,18]. In addition, Yang et al. examined the effect of FES on wheelchair performance, concluding that stimulation resulted in an increase in the user's control with respect to wheelchair propulsion speed [19]. Besides these experimental approaches, also model-based studies have been performed for the purpose of identifying adequate closed-loop control strategies [20,21] and the necessary torque levels for facilitating trunk stability via FES [21,22].

All of these efforts offer valuable insights into the feasibility and effectiveness of FES for enhancing or restoring trunk stability in individuals with SCI. At the same time, larger FES activation levels that can stabilize the upper body against external perturbations have been shown to lead to muscle fatigue [17–20], compromising the functional abilities and safety of the user. Another method of using FES technology is to apply low-intensity FES with the goal of increasing trunk stiffness and damping. Weak muscular co-contractions have been shown to significantly increase trunk stiffness and contribute to postural control in seated healthy individuals [23–27]. Increasing trunk stiffness via low-intensity FES may, however, not only enhance postural stability in any horizontal direction, but also mitigate muscle fatigue as one of the largest challenges associated with FES solutions [20,28–31]. Based on these considerations, we hypothesize a positive effect of low-intensity FES on multidirectional trunk stiffness during sitting. Using a recently proposed methodology for estimating trunk stiffness and damping [32], the objective of this study was to investigate how multidirectional trunk stiffness changes in healthy individuals in response to low-intensity FES of a few selected trunk flexors and extensors. Such a control set will be fundamental for exploring the effect of low-intensity FES on trunk stiffness and postural control in individuals with SCI and other neuromuscular disorders.

2. Methods

2.1. Subjects

Fifteen healthy and young male individuals were invited to participate in this study (age 26.7 ± 4.6 years; height 176 ± 7 cm; weight 72.5 ± 8.1 kg; mean ± standard deviation). All subjects were free from any prior neurological, vestibular, and sensory impairment as well as from any injuries or disorders of the musculoskeletal system. In addition, none of the subjects reported any prior diagnosis of spinal scoliosis or other conditions affecting seated posture. Each subject gave written informed consent to the experimental procedure, which was approved by the local ethics committee in accordance with the declaration of Helsinki on the use of human subjects in experiments.

2.2. Experimental setup and protocol

For each subject, multidirectional trunk stiffness and damping were identified for (1) natural and (2) FES-supported sitting using a mathematical description of a second-order mass-spring-damper system and force and kinematics data from perturbation experiments. During the testing, the subject sat on a custom-made sitting apparatus without touching the ground with his feet, had the forearms resting on his lap, and maintained an upright posture with eyes closed. Independent of FES condition (natural or FES-supported), each subject was instructed prior to each trial to sit in an upright relaxed posture, and this as naturally as possible. One of the subject's hands held an emergency safety button that, when pressed, shut down the power of the custom-made perturbation system (PAPPS) [33] applying the horizontal forces to the subject (Fig. 1A). Details of the experimental setup and data acquisition can be found elsewhere [32].

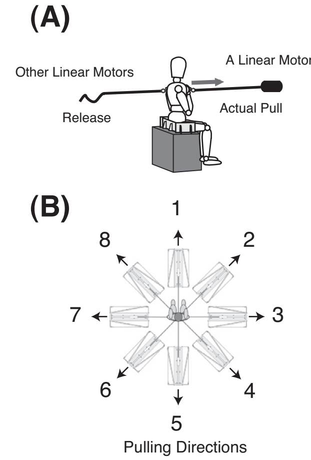

Fig. 1. Schematic of the applied perturbation concept (A) and the eight different perturbation directions (B). The PAPPS delivered horizontal perturbations in the following directions: anterior (1), anterior-right (2), right (3), posterior-right (4), posterior (5), posterior-left (6), left (7), and anterior-left (8). During each trial, a randomly determined PAPPS unit displaced the subject's trunk by 2 cm [32], while the remaining seven units moved toward the subject to prevent any form of interference with or resistance to the evoked response.

The entire perturbation protocol lasted less than 30 min, with 160 pulls in total and 10 pulls per condition (natural or FES-supported) in each of the following directions (Fig. 1B): anterior (1), anterior-right (2), right (3), posterior-right (4), posterior (5), posterior-left (6), left (7), and anterior-left (8). While the natural and FES-supported conditions alternated every 20 pulls, the perturbation direction within each set was randomized to prevent anticipation by the subject (which can significantly influence the neuromuscular state of the trunk and its perturbation response [34]). Every 40 pulls, the subject had a two-minute resting period and was asked to relax his trunk. The execution procedure for each individual perturbation trial has been reported elsewhere [32].

2.3. Delivery of low-intensity electrical stimulation

Four pairs of stimulation electrodes were attached to the skin above the left and right erector spinae and the left and right rectus abdominis muscles, all in the lumbar region. For each back (abdominal) muscle, the cathode and anode were placed vertically and parallel to the spine (midline), with the anode located 4 cm superior of the iliac crest and the cathode 10 cm superior of the anode. The dorsal (ventral) sets were separated laterally from each other by 10 cm, centered with respect to the spine (midline). Note that these locations were chosen based on a previous study that used FES to improve wheelchair propulsion in individuals with SCI [19]. However, in the present study, a larger separation distance between the anode and cathode was chosen to stimulate a larger cross-sectional area of the muscle fibers.

Electrical stimulation was delivered by a portable 4-channel electrical stimulator (Compex Motion II, Compex Motion, Switzerland) using a frequency of 40 Hz and pulse duration of 300 μs. Starting from 0 mA, the stimulation current was slowly increased by 1 mA increments until the experimenter identified the motor threshold of a particular muscle using palpation. During the FES-supported trials, a stimulation current of twice the motor threshold was used. All muscles were stimulated simultaneously, with the objective to generate co-contractions.

2.4. Identification of trunk stiffness and damping

For each executed trial, the kinematics and force data from the first one-second time period following the onset of the perturbation were utilized in the subsequent system identification. Trunk stiffness and damping for each condition (natural and FES-supported) and displacement direction were identified by means of a translational second-order system [27] (*Model I*) and a torsional second-order system [23] (*Model II*). For *Model I*, the trunk was modeled as a point mass concentrated at the trunk's center of mass (COM), which moved in the horizontal plane and was stabilized by translational stiffness k and damping b. The equation of motion of this system is given by:

$$F = m\ddot{x} + b\dot{x} + kx, \tag{1}$$

where F is the applied force, m the mass of the trunk, and $x, \dot{x}, \ddot{x}$ the linear displacement, velocity, and acceleration of the COM, respectively. The effect of any vertical movement caused by the rotational aspect of the trunk was neglected. For *Model II*, the trunk was again modeled as a point mass concentrated at the trunk COM, which rotated around the spinal joint between the fourth and fifth lumbar vertebrae (L4/L5 joint) [35] and was stabilized by rotational stiffness K and damping B. The equation of motion of this system is given by:

$$FL\cos\theta + mgL\sin\theta = I\ddot{\theta} + B\dot{\theta} + K\theta, \tag{2}$$

where F denotes the applied force, L the distance between the center of rotation (L4/L5 joint) and the COM of the trunk, m the mass of the trunk, g the acceleration due to gravity, I the moment of inertia of the trunk, and θ, $\dot{\theta}$, and $\ddot{\theta}$ the angular displacement, velocity, and acceleration of the COM, respectively.

The COM was assumed to be located at the T10 vertebral segment [27], the mass of the trunk (m in Eqs. (1) and (2)) was set to 0.563 of the subject's overall mass [35], and the moment of inertia of the trunk (I in Eq. (2)) was set to $I = m \cdot L^2$ [35]. The conversion from the kinematics time series to angular displacement was performed by placing the origin of the coordinate system on the L4/L5 joint and calculating the angle between the COM displacement vector and a constant vector that spans from the L4/L5 joint to the COM in the equilibrium position.

Using the measured force as the input, the stiffness and damping constants (b, k for *Model I*; B, K for *Model II*) were tuned until a good match between the measured and modeled displacement of the trunk's COM was achieved. The Gauss–Newton optimization algorithm (Optimization toolbox, Matlab ver. 7.5, Mathworks, Massachusetts, U.S.A.) governed the search dynamics, whereas the Percentage-of-Fit (%FIT) was used as the cost function for the optimization procedure:

$$\%\mathrm{Fit} = 100\left(1 - \frac{1}{N}\sum_{i=1}^{N}\left|\frac{y_i - Y_i}{y_i}\right|\right), \tag{3}$$

where N is the number of samples, y_i the experimentally measured and Y_i the modeled displacement of the trunk COM [36]. Only stiffness and damping values from models exhibiting a fitting match of 70% or higher were included in the calculation of the mean trunk stiffness and damping for each individual, condition, and displacement direction.

Using the identified values and the mass of the trunk m, we finally calculated the trunk's undamped natural frequency and damping ratio. The undamped natural frequency ω_0, which is given by:

$$\omega_0 = \sqrt{\frac{k}{m}}\ \text{(translational)} \quad \text{or} \quad \omega_0 = \sqrt{\frac{K}{m}}\ \text{(torsional)}, \tag{4}$$

signifies the frequency at which the undamped system will oscillate when set into motion. The damping ratio ζ, which is given by:

$$\zeta = \frac{b\sqrt{m}}{2m\sqrt{k}}\ \text{(translational)} \quad \text{or} \quad \zeta = \frac{B\sqrt{m}}{2m\sqrt{K}}\ \text{(torsional)}, \tag{5}$$

describes how the system returns to its equilibrium after an impulse force is applied. For a damping ratio greater than 1 (*overdamped*), the system will take a longer time to return to equilibrium position than when the damping ratio is equal to 1 (*critically damped*). For a damping ratio between 0 and 1 (*underdamped*), the system will return faster to equilibrium than the overdamped system, but will exhibit small oscillations around the equilibrium point before settling.

2.5. Statistical analysis

For each condition (natural and FES-supported), the group-average trunk stiffness and damping were identified in dependence of displacement direction using the mean values from all subjects. A two-way analysis of variance (*ANOVA*) with repeated measures was applied to capture potential differences in group-average stiffness and damping between (1) the natural and FES-supported conditions and between (2) the different displacement directions. As a secondary analysis, a paired t-test was used within each displacement direction to identify differences in stiffness and damping between the two FES conditions [37]. The same analyses were performed for the trunk's undamped natural frequency, ω_0, and damping ratio, ζ. For all tests, a significance level of $\alpha = 0.05$ was used. Since the results for the natural condition have been reported elsewhere [32], only the findings for the FES-supported condition in comparison to the natural condition will be presented and discussed.

3. Results

3.1. Experimental attributes and optimization results

Fig. 2 depicts representative examples of the actual displacement (gray lines) and modeled displacement (black lines) of the trunk COM for the two conditions (natural and FES-supported) and the two models (translational and torsional). Subplots show, for a single subject and a backward (5) perturbation, the COM displacement for: (A) natural condition, translational model; (B) natural condition, torsional model; (C) FES-supported condition, translational model; and (D) FES-supported condition, torsional model. Inserts in each subplot provide the time series of the applied input force or torque. A visual inspection suggests that the modeled COM displacement agreed with the measured COM displacement well, and that respective maximum excursions for the FES-supported condition were smaller than for the natural condition.

Across all subjects, the mean preloading force for all eight PAPPS units and trials was 37.6 ± 2.7 N (mean ± standard deviation) for the natural and 38.0 ± 2.6 for the FES-supported condition. The peak force at the perturbing PAPPS unit was 39.9 ± 2.4 N for the natural and 38.7 ± 2.4 N for the FES-supported condition (all perturbation directions and trials). For both preloading and peak forces, no significant differences between the natural and FES-supported conditions were found (t-test; preloading: $p = 0.3954$; peak: $p = 0.9546$). During the FES-supported condition, group average stimulation levels were 20.3 ± 3.8 mA and 24.9 ± 7.4 mA for the rectus abdominis and erector spinae muscles, respectively. From the 2400 trials that were executed in total (15 subjects; 160 trials per subject), less than 0.5% resulted in a percentage-of-fit that was lower than 70% (for both *Model I* and *Model II* together). Those trials were deemed to generate sub-optimal stiffness and damping estimates and were excluded in the subsequent analyses. The remaining trials, which accounted for a mean percentage-of-fit of %FIT = 88.1 ± 6.5% across all subjects and conditions, were used to determine individual and mean values for trunk stiffness and damping as a function of trunk displacement direction.

3.2. Translational model results (Model I)

In Fig. 3, the group-average stiffness and damping results are depicted for the FES-supported condition in comparison to the natural condition when using the translational model (*Model I*). Fig. 3A shows the identified stiffness values and Fig. 3B respective damping values,

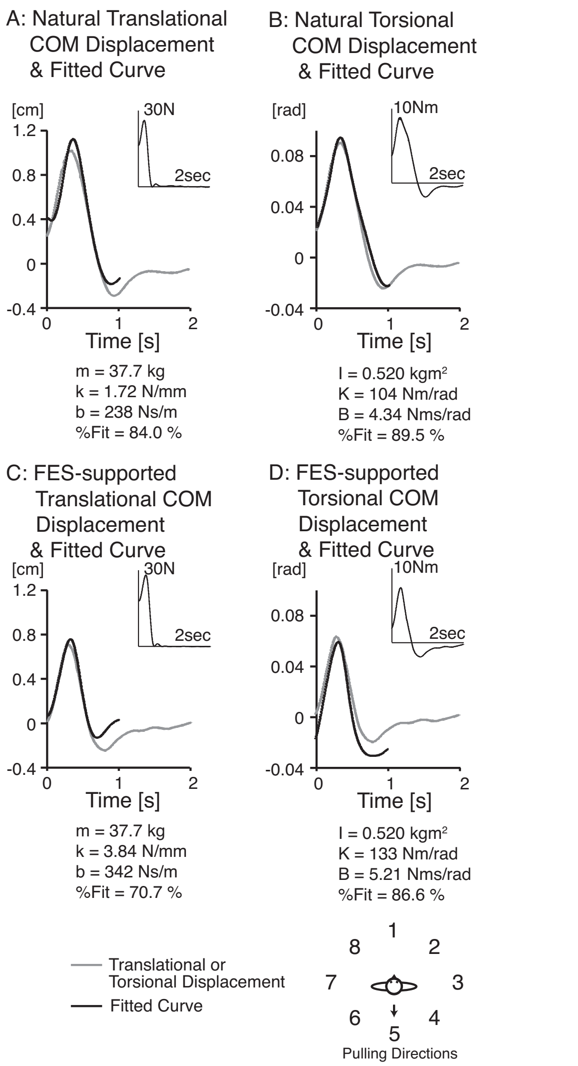

Fig. 2. Representative examples of the actual displacement (gray lines) and modeled displacement (black lines) of the trunk COM for the two conditions (natural and FES-supported) and the two models (translational and torsional). Subplots show, for a single subject and a backward (5) perturbation, the COM displacement for: (A) natural condition, translational model; (B) natural condition, torsional model; (C) FES-supported condition, translational model; and (D) FES-supported condition, torsional model. Inserts in each subplot provide the time series of the applied input force or torque.

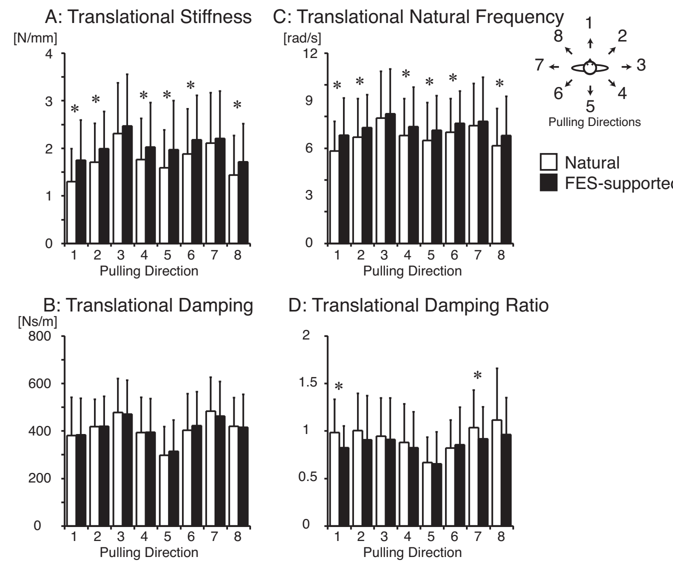

Fig. 3. Group-average stiffness and damping results for the natural condition (white bars) and the FES-supported condition (black bars), using the translational model (*Model I*). Fig. 2A shows the identified stiffness values and Fig. 2B respective damping values, both in dependence of trunk displacement direction. In Fig. 2C and D, the associated undamped natural frequency ω_0 and damping ratio ζ are depicted, respectively.

both as a function of trunk displacement direction. In Fig. 3C and D, the associated undamped natural frequency ω_0 and damping ratio ζ are depicted, respectively.

The two-way *ANOVA* revealed that, for the FES-supported condition, translational stiffness was significantly larger than for the natural condition ($p = 0.0037$). In addition, stiffness values were significantly affected by trunk displacement direction ($p < 0.0001$). The subsequent *t*-tests within each displacement direction indicated that trunk stiffness responded to the low-intensity FES with a significant increase for all directions except the lateral ones (directions 3 and 7; Fig. 3A). When applied to the damping results, both the *ANOVA* ($p = 0.8423$) and the within-direction *t*-tests did not reveal any differences between the natural and FES-supported conditions (Fig. 3B). However, damping values were significantly affected by trunk displacement direction (*ANOVA*; $p < 0.0001$). No interaction effects were found for both stiffness and damping ($p = 0.2098$ and $p = 0.5591$, respectively).

While the bar graph profile for the undamped natural frequency ω_0 generally agreed with the one for trunk stiffness (Fig. 3A and C), the damping ratio ζ showed a more dissimilar profile in comparison with the other plots (Fig. 3D). The *ANOVA* revealed that ω_0 was significantly increased during FES support in comparison to the natural condition ($p = 0.0061$). In addition, ω_0 was also significantly affected by trunk displacement direction ($p < 0.0001$). Same as for the stiffness, ω_0 responded to the low-intensity FES in the *t*-tests with a significant increase for all directions except the lateral ones (directions 3 and 7; Fig. 3C). When applied to ζ, the *ANOVA* did not reveal any differences between the natural and FES-supported conditions ($p = 0.0655$). At the same time, ζ was significantly affected by trunk displacement direction ($p = 0.0001$). The subsequent *t*-tests within each displacement direction indicated that ζ responded to the low-intensity FES with a significant decrease for the anterior and left directions (directions 1 and 7; Fig. 3D). No interaction effects were found for both ω_0 and ζ ($p = 0.1530$ and $p = 0.0764$, respectively).

3.3. Torsional model results (Model II)

In Fig. 4, the group-average stiffness and damping results are depicted for the FES-supported condition in comparison to the natural condition when using the torsional model (*Model II*). Fig. 4A shows the identified stiffness values and Fig. 4B respective damping values, both as a function of trunk displacement direction. In Fig. 4C and D, the associated undamped natural frequency ω_0 and damping ratio ζ are depicted, respectively.

The two-way *ANOVA* ($p = 0.1162$) and within-direction *t*-tests revealed no significant differences in torsional stiffness between the natural and FES-supported conditions (Fig. 4A). At the same time, stiffness values were significantly affected by trunk displacement direction (*ANOVA*; $p < 0.0001$). When applied to the damping results, both the *ANOVA* ($p = 0.3574$) and the within-direction *t*-tests did not reveal any differences between the natural and FES-supported conditions (Fig. 4B). However, damping values were significantly affected by trunk displacement direction (*ANOVA*; $p < 0.0001$). No interaction effects were found for both stiffness and damping ($p = 0.1975$ and $p = 0.3693$, respectively).

The *ANOVA* revealed no significant differences in undamped natural frequency ω_0 between the natural and FES-supported conditions ($p = 0.0726$). At the same time, ω_0 was significantly

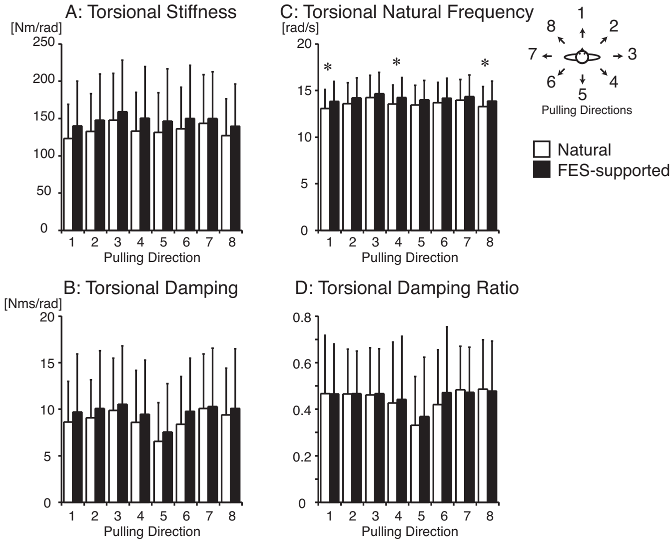

Fig. 4. Group-average stiffness and damping results for the natural condition (white bars) and the FES-supported condition (black bars), using the torsional model (*Model II*). Fig. 3A shows the identified stiffness values and Fig. 3B respective damping values, both in dependence of trunk displacement direction. In Fig. 3C and D, the associated undamped natural frequency ω_0 and damping ratio ζ are depicted, respectively.

affected by trunk displacement direction ($p < 0.0001$). ω_0 responded to the low-intensity FES in the *t*-tests with a significant increase for the anterior, posterior-right, and anterior-left directions (directions 1, 4, and 8; Fig. 4C). When applied to ζ, both the *ANOVA* ($p = 0.6083$) and the within-direction *t*-tests did not reveal any differences between the natural and FES-supported conditions (Fig. 4D). However, ζ was significantly affected by trunk displacement direction (*ANOVA*; $p < 0.0001$). No interaction effects were found for both ω_0 and ζ ($p = 0.1780$ and $p = 0.2213$, respectively).

4. Discussion

The present study is set out to investigate whether low-intensity FES has the ability to increase trunk stiffness and damping in healthy individuals during sitting. In what follows, we discuss the characteristics and significance of the obtained results as well as potential limitations. Note that the validity of the developed methodology and the direction dependency of the identified parameters without low-intensity FES have been reported elsewhere [32].

4.1. Effects of low-intensity FES on multidirectional trunk stiffness and damping

A visual inspection of Figs. 3A and 4A suggests that, independent of trunk displacement direction and system identification model (*Model I* and *Model II*), average trunk stiffness was found to be higher for the FES-supported condition than for the natural condition. An FES-induced increase in trunk stiffness presumably results from augmented stiffness of the stimulated muscles (similar to that during low-intensity co-contractions of antagonist muscles), which in turn enhances spine stability. Note that the two-way *ANOVA* confirmed the visual inspection for the translational model (*Model I*) only. Here, FES resulted in a significant increase in trunk stiffness for all but the lateral directions, which can be explained by the fact that muscles contributing to anterior–posterior trunk stability were the only ones stimulated in this study. First and foremost, our results imply that low-intensity FES has the potential to counteract horizontal perturbations as experienced during ADL. However, they also indicate that lateral abdominal muscles such as the internal and/or external obliques need to be targeted to increase trunk stiffness in the lateral directions.

In contrast to translational stiffness (*Model I*), the *ANOVA* did not reveal significant differences in torsional stiffness (*Model II*) when comparing the two FES conditions. One potential reason for this lack of significance may be that, in the torsional model, the mass of the body was represented as a point mass, with the COM being located at the center of the T10 vertebral segment. This simplification along with respective approximation of the trunk's moment of inertia may not adequately capture the dynamics of the trunk during perturbations, yielding slightly different results in comparison to the more realistic translational model.

Both the *ANOVA* and *t*-test did not reveal any significant differences between trunk damping with and without FES, suggesting that low-intensity FES does not modulate the viscous properties or behavior of the trunk. This agrees with our understanding that trunk damping is caused by viscous elements such as viscera in the abdominal cavity and/or soft mechanical properties of tissue and tendons. While these elements do dampen the trunk motion during perturbations, they should not be affected by low-intensity FES, either directly or via increased muscle stiffness.

The effect of low-intensity FES on the natural frequency of the trunk (ω_0) was similar to that on trunk stiffness (see Figs. 3A/C and 4A/C), indicating an overall increase for the translational model only. This general agreement can be attributed to the proportionality of ω_0 to the square root of trunk stiffness (see Eq. (4)). Since an increase in natural frequency has been linked to less sluggish system dynamics [38], it can be speculated that increasing ω_0 via low-intensity FES may have exactly that effect on the trunk. In contrast to ω_0, low-intensity FES did not have a distinct effect on the damping ratio, ζ. The specific finding that some perturbation directions exhibited an increase and others a decrease in the average value of ζ following FES (see Figs. 3D and 4D) can be explained by the fact that ζ is a function of both trunk damping and the inverse of trunk stiffness (see Eq. (5)).

4.2. Significance of increasing trunk stiffness via FES

As described earlier, also other studies have pursued the potential of FES to increase trunk stability in individuals with SCI [16–19]. While these efforts mark important milestones towards the field's goal of developing neuroprostheses for trunk control, none of them have explored the possibility to specifically increase trunk stiffness and, hence, stability via low-intensity FES. There is no doubt that many individuals with SCI require higher, dynamically regulated levels of FES in order to ensure trunk stability during ADL. However, common tasks such as sitting and standing may be accomplished by bolstering trunk stiffness via low-intensity FES only. In fact, a recent study has found that, during quiet sitting, the trunk musculature exhibits only 3–10% of activation levels obtained during maximum voluntary contractions [39]. This indicates that the sitting posture is supported by the tensegrity structure of the trunk, suggesting that low values of preloading as generated by low-intensity FES may be sufficient for stabilization. While such a "base level" of FES can minimize the occurrence of fatigue, it must be complemented by higher, dynamically regulated levels of FES if demanded by a given situation (such as reaching for an object). Subsequent studies will also have to show that a base level of FES can also increase trunk stiffness in individuals with SCI whose muscle and trunk properties are oftentimes significantly altered following injury.

4.3. Study limitations and future directions

While the main limitations of the experimental and system identification procedures have been reported elsewhere [32], two additional limitations with respect to the FES application have to be addressed. First, measuring the muscle activity of the stimulated muscles via mechanomyography [40] would have allowed us to monitor whether neurally regulated muscle contractions in response to the perturbations are affected by low-intensity FES. Second, stimulating lateral abdominal muscles (such as the external obliques) in addition to the rectus abdominis and erector spinae muscles would have provided, in terms of statistical significance, a more complete picture of the effect of low-intensity FES on multidirectional trunk stiffness. While our results clearly indicate that a base level of FES has the potential to increase multidirectional trunk stiffness, future studies should include lateral stimulation sites as well. Finally, we need to investigate the effect of low-intensity FES on trunk stiffness following SCI as it may be very different in comparison to healthy individuals.

5. Conclusions

Using a recently proposed methodology for estimating trunk stiffness and damping during sitting [32], the present study demonstrates that low-intensity FES is a simple and effective method for increasing trunk stiffness on demand. More specifically, it was found that low-intensity FES increases trunk stiffness in those directions that are associated with the stimulated muscles. In contrast, trunk damping was not altered by the applied low-intensity FES. Subsequent studies will use the developed perturbation and FES protocols to assess the effect of low-intensity FES on trunk stiffness and sitting stability in individuals with neurological impairments such as SCI.

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Conflict of interest

There are no conflicts of interest for the authors of this study. Dr. Popovic is a shareholder in the company MyndTec Inc. However, his activities with MyndTec Inc have no bearing on this project and its outcomes/results.

Ethical approval

The study was approved by the Research Ethics Board of the University of Toronto and the Toronto Rehabilitation Institute, Toronto, ON, Canada (reference number: REB05-017).

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