Neuromuscular Control of the Trunk during Sitting Balance

by

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A thesis submitted in conformity with the requirements for the degree of Doctor of Philosophy in Biomedical Engineering Institute of Biomaterials and Biomedical Engineering University of Toronto

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Abstract

Sitting stability is maintained by continuous and intermittent activation of trunk and lower limb muscles. However, spinal cord injury (SCI) can result in paralysis of the trunk muscles and affects sitting balance. Currently, chest straps and trunk braces are used to correct sitting after SCI. In addition to these passive devices, functional electrical stimulation (FES) could be used to artificially contract trunk muscles and regulate sitting balance. The objectives of this research project were to: (1) investigate neuromuscular synergies of the trunk during perturbed sitting; (2) investigate the effects of anticipation of the perturbation on trunk muscle responses; (3) compare sitting balance between able-bodied individuals and people with SCI; and (4) examine the effects of electrical stimulation of trunk muscles on sitting balance. The first deliverable (Chapter 2) of this project was the development of neural network models that encode trunk muscle responses during perturbed quiet sitting. These models describe complex neuromuscular synergies of the trunk and allow us to understand which muscles are used to stabilize
sitting balance. The second deliverable (Chapter 3) was a study which showed that anticipation of direction and time of perturbation resulted in approximately 15 ms faster trunk muscle activations. These results indicate that the reflexive and patterned neuromuscular responses of the trunk can be modulated by the central nervous system using anticipatory information. The third deliverable (Chapter 4) was a study that investigated differences in sitting balance between able-bodied individuals and individuals with SCI. This study has shown that trunk control is significantly compromised in individuals with SCI, which is the main cause of their reduced balance control during quiet sitting. The fourth deliverable (Chapter 5) was a study that investigated the effects of FES of trunk muscle on sitting balance control. The results showed that co-contractions of trunk muscles with FES could increase trunk stiffness. These findings combined advanced our understanding of the neuromuscular control of the trunk during sitting balance. Moreover, they helped us develop a neuroprosthesis, which could improve sitting balance in individuals with SCI, and consequently contribute to improving their mobility, independence and, ultimately, quality of life.
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Matija Milosevic, 2015
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### Abbreviations

<table>
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<tr>
<th>Description</th>
<th>Abbreviation</th>
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<tbody>
<tr>
<td>angle of the COM</td>
<td>$\theta$</td>
</tr>
<tr>
<td>motor transmission time delay</td>
<td>$\tau_1$</td>
</tr>
<tr>
<td>sensory transmission time delay</td>
<td>$\tau_2$</td>
</tr>
<tr>
<td>two-dimensional</td>
<td>2-D</td>
</tr>
<tr>
<td>artificial neural network</td>
<td>ANN</td>
</tr>
<tr>
<td>anterior-posterior</td>
<td>AP</td>
</tr>
<tr>
<td>95% confidence ellipse</td>
<td>AREA-CE</td>
</tr>
<tr>
<td>body height</td>
<td>B</td>
</tr>
<tr>
<td>passive damping</td>
<td>$B$</td>
</tr>
<tr>
<td>centroidal frequency</td>
<td>CFREQ</td>
</tr>
<tr>
<td>central nervous system</td>
<td>CNS</td>
</tr>
<tr>
<td>center of mass</td>
<td>COM</td>
</tr>
<tr>
<td>center of pressure</td>
<td>COP</td>
</tr>
<tr>
<td>feet COP</td>
<td>COP$_F$</td>
</tr>
<tr>
<td>global COP</td>
<td>COP$_G$</td>
</tr>
<tr>
<td>seat COP</td>
<td>COP$_S$</td>
</tr>
<tr>
<td>external oblique muscle</td>
<td>EO</td>
</tr>
<tr>
<td>functional electrical stimulation</td>
<td>FES</td>
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<tr>
<td>frequency dispersion</td>
<td>FREQD</td>
</tr>
<tr>
<td>height of the COM</td>
<td>$h$</td>
</tr>
<tr>
<td>moment of inertia</td>
<td>$I$</td>
</tr>
<tr>
<td>internal oblique muscle</td>
<td>IO</td>
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</tbody>
</table>
$K$  mechanical stiffness

$K_D$ derivative gain

$K_P$ proportional gain

$L_3$ lumbar erector spinae muscle

$M$ body mass

$m$ moving mass

$MD$ mean distance

$MFREQ$ mean frequency

$ML$ medial-lateral

$MV$ mean velocity

$NMS$ neuromusculoskeletal

$P50$ 50% power

$P95$ 95% power

$PD$ proportional-derivative

$R^2$ coefficient of determination

$RA$ rectus abdominis muscle

$RANGE$ range

$RD$ radial distance

$SCI$ spinal cord injury

$SOM$ self-organizing maps

$T9$ thoracic erector spinae muscle

$UCM$ uncontrolled manifold analysis

$y_{COP}$ anterior-posterior COP
Chapter 1

1. Introduction

1.1. Problem Statement and Motivation

In Canada, approximately 86,000 people are living with a spinal cord injury (SCI), and some 4,300 new cases of SCI occur every year. Based on the current demographic trends in Canada, it can be expected that by the year 2030, some 121,000 people will be living with an SCI (~50% increase from 2010), and that incidence could rise to 5,800 new cases per year (Farry and Baxter 2010). Estimates in the United States vary between 200,000 people with chronic SCI (Anderson, 2004) and 1,275,000 people who self-reported being paralyzed due to an SCI (The Christopher and Dana Reeve Foundation, 2009). The lifetime cost of SCI is approximately $1.47 million for paraplegia and $3.0 million for tetraplegia, and it is approximated that SCI present a total annual healthcare cost of $3.6 billion in Canada (Krueger 2011).

Sitting instability is a significant problem for individuals with SCI because it can affect their ability to perform activities of daily living and results in secondary complications such as pressure sores, kyphosis and respiratory dysfunction, some of which require further hospitalization and can be fatal. Individuals with paraplegia reported trunk stability as the third most important recovery priority that could improve their quality of life (Anderson, 2004). My research examines trunk muscle control during
sitting balance as a way to design a functional electrical stimulation (FES) control strategy that can be used to improve trunk stability and sitting balance after SCI. FES can be used to activate muscles artificially by applying short electric impulses over the muscle nerve. A device that uses FES to activate muscles and move segments of the body in a functional manner is called a neuroprosthesis (Popovic and Thrasher 2004). In order for FES to be used effectively for the recovery of a functional tasks, one must first understand the neuromuscular and biomechanical mechanisms used during these functional tasks. Therefore, the goal of my study is to advance the understanding of the mechanisms responsible for sitting balance and contribute to the development of a sitting neuroprosthesis that will use FES to activate trunk muscles. The ability to control trunk has a direct impact on an individual’s ability to reach, grasp and release objects in activities of daily living, to perform transfers and to become mobile. The neuroprosthesis for sitting is expected to improve trunk stability, sitting balance and help with the performance of activities of daily living, which will contribute to the overall improved quality of life for individuals with SCI.

1.2. Thesis Objectives

The main objectives of my thesis research are to:

(1) Investigate neuromuscular synergies of trunk muscles during perturbed sitting.

- **Objective:** The first objective of my research was to analyze directional-dependence of the neuromuscular synergies of the trunk muscles during eight directional perturbations and to quantify responses of the trunk muscles using descriptive models.
• **Research Question:** Which trunk muscle synergies activate to stabilize sitting balance perturbations? Specifically, are there directional dependencies of neuromuscular synergies of the trunk muscles during different perturbations and can these trunk muscle responses be captured using descriptive models?

• **Hypothesis:** It was hypothesized that the trunk muscle synergies will have unique, direction-dependent responses which can be modeled.

(2) Investigate effects of anticipation of the perturbation on trunk muscle responses.

• **Objective:** The second objective of my research was to identify the neuromuscular responses during sitting balance perturbations and to compare the trunk muscle responses during anticipated and unanticipated sitting perturbations.

• **Research Question:** What are the response latencies of the muscle activations during sudden perturbations and how does the anticipation of perturbation modulate trunk muscle responses?

• **Hypothesis:** It was hypothesized that trunk muscles will respond using latencies which indicate reflexive and non-voluntary responses. Moreover, it was expected that amplitude of the trunk muscle responses will be larger during anticipated perturbations compared to unanticipated perturbations.

(3) Compare sitting balance between able-bodied individuals and people with SCI.

• **Objective:** The third objective of my research was to compare sitting balance of people with cervical SCI and able-bodied individuals, and to investigate the effect of foot support and trunk control on sitting balance.
• **Research Question:** What is the level of the sitting balance impairment of individuals with cervical SCI compared to able-bodied people, and what are the mechanisms responsible for this impairment?

• **Hypothesis:** It was hypothesized that individuals with cervical SCI will have significantly compromised sitting balance and that trunk control will be the main mechanism responsible for the poorer sitting balance among individuals with SCI.

(4) Examine the effects of electrical stimulation of trunk muscles on sitting balance.

• **Objective:** The last objective of my research was to characterize the changes in sitting balance control that resulted from the application of FES to the trunk muscles by comparing FES-assisted sitting and sitting without FES.

• **Research Question:** How does the application of FES on the trunk muscles affect sitting balance control among able-bodied individuals?

• **Hypothesis:** It was hypothesized that FES of the trunk muscles will stiffen the trunk and result in improved sitting balance, characterized by reduced postural sway.

1.3. Background

1.3.1. Trunk Stability and Sitting Balance

The human spine system is inherently unstable, and relies on the surrounding musculature and soft tissues to maintain stability of the trunk (Cholewicki, Panjabi and Khachatryan, 1997). Numerous musculoskeletal elements of the trunk make the system
biomechanically redundant (i.e., the same movements can be executed by a variety of muscular patterns), and trunk stability requires complex muscle coordination (Bergmark, 1989). Trunk muscle co-activation can provide stability of the trunk (Stokes et al., 2000), but a more desired strategy for trunk muscle activations has been suggested to include tonic stiffness through muscle co-activations, in combination with phasic, feedback driven activations to stabilize against external perturbations (Moorhouse and Granata, 2006; Preuss, Grenier and McGill, 2005). Trunk muscles may act across a range of perturbation directions (Stokes and Gardner-Morse, 1999).

Neuromuscular response latencies of the trunk have been reported in the range between 25-150 ms (Stokes et al. 2000). This suggests that the system has medium latency responses, i.e., that monosynaptic or simple polysynaptic reflexes regulate trunk balance during sudden unexpected perturbations (Stokes et al. 2000). Therefore, the central nervous system (CNS), in response to sudden unexpected perturbations, controls trunk stability using neuromuscular structures that do not involve conscious perception and high level decision making, i.e., most of the control is done by neuromuscular systems that primarily involve reflexes and low level CNS control structures.

Functional tasks that involve sitting balance require high levels of trunk stability (Cholewicki et al., 2000). Literature on trunk stability has focused primarily on the vertebral instability related to back pain (Cholewicki, Panjabi and Khachatryan 1997) and postural instability in neurological conditions such as SCI (Seelen et al., 1998; Kukke and Triolo, 2004; Triolo et al. 2009). Recent investigations of trunk stability during sitting have focused much effort on analyzing and modeling the able-body trunk function. Neuromuscular responses of the trunk have revealed insights into the mechanisms of
trunk control which are essential for both standing and sitting balance (Preuss and Fung 2008). Quiet sitting posturography has also revealed similarities of standing and sitting postural control (Vette et al. 2010). The dynamic role of abdominal and back muscles has been identified during sitting, and their direction-dependent neuromuscular responses were described using the Gaussian model (Masani et al. 2009). Moreover, tonic stiffness has been reported in the range of 1-3% for abdominal, and 4-6% for back muscles, relative to maximal voluntary contractions (Masani et al. 2009). These results support the notion that the trunk system behaves like an underdamped oscillator, which is not controlled by stiffness alone (Thrasher et al., 2011). A multi-segmented kinematic analysis of the spine has identified complexities of the spinal system and the superiority of full kinematic analysis over simplified models (Preuss and Popovic 2010a). Preuss and Popovic (2010b) quantitatively defined limits of stability in sitting, which describe balance of able-bodied individuals, and they were shown to cover approximately 33% of the base of support. Overall, the presented literature emphasizes the importance of tonic stiffness and phasic, feedback driven, muscle responses as essential for stabilizing the complex biomechanics of the trunk. The first deliverable, which meets Objective 1 of my thesis (Chapter 2), was a study that developed models which can encode trunk muscle responses during perturbed sitting. These quantitative models describe complex neuromuscular synergies of the trunk muscles and allow us to understand which muscles are used to stabilize sitting balance following multidirectional perturbations.
1.3.2. *Anticipatory Control of Sitting*

The role of anticipatory postural mechanism during sitting has received relatively little attention, although there is evidence to suggest that it may be imperative to trunk stability. Anticipatory reactions could be caused by either self-inflicted perturbations or by predictable external perturbations. During rapid arm movements, which are self-inflicted perturbations, the CNS initiates anticipatory postural adjustments (APA) of postural muscles before the movement in order to minimize postural disturbances in a feedforward manner (Aruin 2002). Results of APA during sitting are somewhat conflicting. Absence of APA has been shown in sitting during fast pointing movements (van der Fits et al. 1998) but, clear evidence of APA has been demonstrated during isometric maximal force-exertion tasks (Le Bozec et al. 2001; Aruin and Shiratori 2003). APA could change or even disappear depending on the magnitude of perturbation, voluntary movement causing the perturbation and the postural task (Latash 2008; Aruin 2002). Consequently, APAs are used to understand how the CNS regulates motor control of one or more body parts during different biomechanical, sensory or balance conditions that CNS is anticipating.

Anticipatory activations of the trunk have also been demonstrated during predictable external perturbations. These feedforward trunk muscle activations were reported primarily in abdominal muscles among subjects expecting sudden perturbations (Lavender et al., 1989; Marras et al. 1987). It was shown that agonist trunk muscles activated prior to predictable perturbations and switched off immediately after, whereas the antagonist muscles remained inactive prior to and activated following external loading perturbations in able-bodied individuals (Cholewicki et al. 2000). Pre-activation
of trunk muscles is not always sufficient, so responses after the perturbation could be required to maintain trunk stability (Cholewicki et al. 2000). The amount of pre-activation of trunk muscles has an influence in decreasing the reactive responses (Stokes et al. 2000). Excessive co-activations may also lead to quick muscle fatigue (Hunter et al., 2004). Taken together, the presented literature suggests the importance of anticipatory trunk muscle activations for trunk stability during sitting. The second deliverable, which meets Objective 2 of my thesis (Chapter 3), was a study that showed that anticipation of direction and time of the perturbation resulted in approximately 15 ms faster trunk muscle activations. These results indicate that the reflexive and patterned neuromuscular responses of the trunk can be modulated by the CNS using anticipatory information.

1.3.3. Sitting Balance after Spinal Cord Injury

Injury to the spinal cord often leads to impairment of the motor and/or sensory functions below the level of the injury. Injuries of the spinal cord in the cervical and thoracic vertebral region lead to motor and/or sensory impairment of the trunk muscles (Chen et al., 2003; Seelen et al. 1998; Triolo et al. 2009). Individuals with neurological SCI levels T10 to C1 have severely affected trunk control which has an impact on their sitting balance (Chen et al., 2003), and this could result in backward pelvis tilting and compromised seated posture (Hobson and Tooms, 1992). As a result, people with SCI use innervated, non-postural muscles (e.g., latisumus dorsi and pectoralis major) to compensate for the impairment, but their sitting balance still remains compromised (Seelen et al., 1998). Moreover, during reaching tasks most individuals with SCI use one arm to maintain support by holding the wheelchair or by pivoting it on the lap for support
Other mechanical solutions for improving sitting balance include chest straps (Reft and Hasan 2002), seat cushions (Aissaoui et al., 2001), foot rests (Potten et al. 2002) and wheelchair adjustments (Potten et al. 1999). These solutions are limiting since they improve balance at the cost of reduced trunk mobility.

Anticipatory adjustments were suggested as a potential surrogate measure for assessing balance among individuals with SCI (Gagnon et al. 2009). It has been shown that initial posterior displacement of the center of pressure occurred before the start of the forward-reaching movements among SCI subjects during sitting (Potten et al., 2002). It is also known that SCI affects neuromuscular control and sensory pathways that contribute to impaired control of sitting. Nevertheless, the mechanisms of sitting balance after SCI remain relatively unexplored. The third deliverable, which meets Objective 3 of my thesis (Chapter 4), was a study that investigated differences in sitting balance control between able-bodied individuals and individuals with SCI. This study has shown that trunk control is significantly compromised in individuals with SCI as compared to able-bodied individuals, which is the main cause of their reduced balance control during quiet sitting.

**1.3.4. Functional Electrical Stimulation (FES)**

Functional electrical stimulation (FES) systems work by applying short electric pulses to a group of muscle nerves, which cause an action potential and a muscle contraction to occur. Surface FES systems deliver a potential difference between two electrodes placed over the nerve that is innervating the muscle of interest, whereas implantable systems apply the stimulation directly to the nerve that is innervating the muscles (Popovic and Thrasher 2004). The magnitude of stimulation can be varied by...
changing the pulse amplitude, width or frequency. Neuroprostheses are devices that use FES to activate paralyzed muscles when voluntary control is not possible because of neurological impairment to the CNS such as stroke and SCI (Popovic and Thrasher 2004). Clinical use of FES includes restoration of upper extremity and lower extremity function, bladder and bowel function, and respiratory function (Peckham and Knutson 2005).

FES systems are usually open-loop controlled. Simple open-loop control systems (i.e., feedforward-based control systems) deliver the stimulation to the targeted muscles without taking into consideration errors caused by perturbations and or the FES systems themselves. They simply deliver preprogrammed stimulation sequences. These systems are appropriate for controlling tasks such as bladder voiding and hand opening, and act as finite-state controllers (Braz et al., 2009). The closed-loop control systems (i.e., feedback-based control systems), which are rarely used in clinical applications, provide disturbance rejection capabilities derived from the error information (Abbas and Gillette 2001). Error signals for feedback control of FES are usually obtained from biological signals or man-made sensors (Braz et al., 2009, Sinkjaer et al., 2003). The closed-loop controlled systems implement error rejection algorithms in addition to tracking the desired trajectory(ies), while open-loop control systems act as finite-state controllers that rely on a pre-existing set of commands to activate the muscles (Braz et al., 2009). To date, due to various technical challenges, the FES systems that implement closed-loop control are rare and majority of the FES devices used today implement open-loop controllers. Examples of open-loop controlled FES devices are systems for walking, reaching, grasping and bladder voiding (Lynch and Popovic 2008).
Application of open-loop controlled FES for sitting balance and functional task performance among individuals with SCI has gained attention recently, and results of the first studies seem encouraging. Kukke and Triolo (2004) demonstrated that stimulation of the lumbar erector spinae in four individuals with SCI significantly improved forward reaching. It has also been shown that FES stimulation of lumbar erector spinae, quadratus lumborum, and gluteus maximus improved spinal alignment, sitting posture, pulmonary function, trunk stability, and reaching function among patients with SCI (Triolo et al., 2009). Yang et al. (2009) also demonstrated that a surface FES system applied to the trunk muscles enhanced manual wheelchair propulsion, among individuals with SCI. These studies demonstrated better performance of functional task when open-loop FES systems have been used to improve sitting or sitting balance. An assessment of multi-intensity stimulation revealed that high-intensity (vs. low-intensity) stimulation produced better functional results (Yang et al., 2009). Overall, these studies have shown encouraging results in support of FES-assisted sitting. The fourth and last deliverable of my project, which meets Objective 4 of my thesis (Chapter 5), was a study that investigated the effects of FES of trunk muscles on sitting balance control. Based on the results obtained from Chapters 2, 3 and 4, FES was used to artificially activate trunk muscles during sitting balance. The results showed that co-contractions (i.e., open-loop control) of trunk muscles with FES could increase trunk stiffness. This is a necessary proof of principle step in establishing the feasibility for using FES to improve sitting balance by increasing trunk stiffness.
1.4. References


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Chapter 2

2. Visualization of Trunk Muscle Synergies during Sitting Perturbations using Self-Organizing Maps (SOM)

The material presented in this chapter has been published in the article:


*NOTE:* The content of this chapter is identical to the material presented in the publication except for the text formatting which was done according to University of Toronto requirements.
2.1. Abstract

The purpose of this study was to demonstrate the use of the self-organizing maps (SOM) method for visualization, modelling and comparison of trunk neuromuscular synergies during sitting. Thirteen participants were perturbed at the level of the sternum, in eight directions during sitting. Electromyographic (EMG) responses of ten trunk muscles involved in postural control were recorded. The SOM was used to encode the EMG responses on a two-dimensional (2-D) projection (i.e., visualization). The result contains similar patterns mapped close together on the plot therefore forming clusters of data. Such visualization of ten EMG responses, following eight directional perturbations, allows for comparisons of direction-dependent postural synergies. Direction-dependent neuromuscular response models for each muscle were then constructed from the SOM visualization. The results demonstrated that the SOM was able to encode neuromuscular responses, and the SOM visualization showed direction-dependent differences in the postural synergies. Moreover, each muscle was modelled using the SOM-based method, and derived models showed that all muscles, except for one, produced a Gaussian fit for direction-dependent responses. Overall, SOM analysis offers a reverse engineering method for exploration and comparison of complex neuromuscular systems, which can describe postural synergies at a glance.

**Keywords:** Balance, electromyography, muscle synergy, perturbation, self-organizing map, sitting, visualization.
2.2. Introduction

Trunk stability is responsible for maintaining upright posture of the spine during standing and sitting. Trunk stability relies on complex synergistic muscle activations, which play an important role during standing and sitting balance control. Previous analysis of trunk stability have examined over 40 muscles of the ‘spine system’ (Bergmark, 1989), which are difficult to evaluate and interpret as a collective system using standard analysis methods. Consequently, there is a need to develop a technique which would allow one to quickly and intuitively analyze synergistic activity of many muscles that act in concert and to derive principles of their synergistic activity from the individual electromyography (EMG) recordings of the muscles of interest during a particular neuromuscular activity (Merletti and Farina, 2008).

In postural control, gross movements that require a number of interdependent and simultaneous muscle responses are known as postural synergies. Postural synergies are control signals for groups of muscles that work together to assure stability of a certain joint or a body segment (Latash, Scholz, and Schoner, 2007). It is known that tonic muscle activation contributes to stability of the trunk during sitting balance (Masani, et al., 2009). It is established that tonic activation of the trunk muscles contributes to the stability of the trunk during sitting and standing balance (Masani, et al., 2009). It has also been established that phasic, feedback-driven, trunk muscle responses help maintain trunk stability during perturbed sitting (Masani, et al., 2009; Thrasher, et al., 2010). In this study, we analyzed postural synergy and the perturbation-induced phasic response of the trunk muscles that work collaboratively to ensure stability of the trunk during perturbed sitting.
The concept of synergy is related to the understanding of how the central nervous system (CNS) activates multiple muscles in order to perform complex movements (Latash, Scholz, and Schoner, 2007). Muscle synergies are analyzed by examining correlations between pairs of muscles (Jensen, et al., 1994). However, correlation analysis is not sufficient when investigating tasks that involve more complex synergies (Latash, Scholz, and Schoner, 2007). Statistical methods using matrix factorization, such as principal component analysis (Mah, et al., 1994), gradient descent (Saltiel, et al., 2001), and cluster analysis (Holdefer and Miller, 2002), offer a solution for investigating more complex mechanisms by analyzing average performances over numerous repeated trials. Uncontrolled manifold (UCM) analysis method (Krishnamoorthy, et al., 2003) evaluates the variability between trials in order to analyze synergies qualitatively. Though all these methods rely on extensive data analysis, which can often be difficult to interpret and conceptualize, they account for the complexities inherent in postural control, and can thus contribute to the understanding of how the CNS controls multiple muscles during complex movements.

Our study presents a self-organizing maps (SOM) method for representing, comparing and modelling complex postural synergies at a glance. The SOM is an artificial neural network (ANN) that uses an unsupervised learning algorithm to project large input datasets onto a two-dimensional (2-D) representation known as a map (Kohonen, 2001). The SOM produces an organized map in which similar patterns, discovered in the input data, are mapped onto nodes close to one another on the map. Thus the map becomes a projection of the input data and allows visualization of large datasets on a 2-D display, while maintaining their topological order (Kohonen, 2001).
Consequently, visualization of numerous EMG responses following perturbations results in a method for representing and comparing postural synergies at a glance.

The SOM algorithm has been used as a robust method for classification of neuromuscular disorders based on EMG recordings (Pattichis, Schizas, and Mittleton, 1995) and for exploration of gait coordination based on locomotion kinematic data (Lamb, et al., 2008; Lamb, et al., 2011; Schollhorn, et al., 2009) for a review of other applications of SOM in biomechanics). To date, there have been limited applications of SOM for visualization of neuromuscular synergies in posturography. Moreover, postural muscle synergies represent a general construct used by the CNS (Torres-Oviedo and Ting, 2007) and may reveal insight into neural strategies used by healthy and impaired nervous systems (Safavynia, Torres-Oviedo, and Ting, 2011). The SOM presents topological relationships of high-dimensional, non-linear data visually, thus making it an attractive tool for analyzing postural muscle synergies.

The objective of this study is to use the SOM method to represent and compare postural muscle synergies by producing a visualization of complex neuromuscular responses following perturbations. Furthermore, the objective is to produce response models of each muscle and compare the results obtained with the SOM analysis to the results obtained by Masani et al. (Masani, et al., 2009) using curve fitting. Overall, the aim of this study was to demonstrate the use of SOM for visualization and comparison of neuromuscular synergies in posturography. The SOM analysis was expected to contribute to further understanding and aid the reverse engineering of the neural mechanisms responsible for sitting balance control, which relies on complex neuromuscular relationships (Bergmark, 1989).
2.3. Methods

The full experimental protocol is reported in our previous study (Masani, et al., 2009; Thrasher, et al., 2010). A brief description follows.

2.3.1. Subjects

This study included thirteen healthy male adults (ages: 21–39 years; mean height: 178.0 (SD: 4.7) cm; mean body mass: 70.3 (SD: 10.0) kg; and all except one were right-handed). Participants had no reported history of lower back problems. The experimental protocol was approved by the local ethics committee and all subjects gave written informed consent before participating.

2.3.2. Experimental Protocol

Subjects were seated in an upright position with legs unsupported, arms crossed over the chest, and eyes closed; subjects wore headphones to eliminate auditory cues. The experiment consisted of eight directional perturbations at the level of sternum, uniformly spaced at intervals of 45° around the subject. Perturbations were applied via manual pulling using a chest harness. The applied perturbation forces were in the range from 131 to 148 N (Masani, et al., 2009). Five trials were taken for each of eight directions (total of 40 perturbations) for each subject. The perturbations were randomly ordered to prevent any anticipation, with approximately 30s between perturbations.

2.3.3. Data Acquisition

Surface EMG recordings were taken from ten muscle groups (five muscles
recorded bilaterally) that were identified as relevant for posture and trunk stability. Ten disposable EMG electrodes (silver-silver chloride disposable electrodes, 10mm diameter) were placed bilaterally, 18mm apart, over the following muscles: 1) rectus abdominis (RA), 3cm lateral to the umbilicus; 2) external oblique (EO), 15cm lateral to the umbilicus; 3) internal oblique (IO), midpoint between the anterior superior iliac spine and the symphysis pubis; 4) thoracic erector spinae (T9), 5cm lateral to the T9 spinous process; and 5) lumbar erector spinae (L3), 3cm lateral to the L3 spinous process. A reference electrode was placed over the clavicle.

Data were acquired using two AMT-8 EMG recording systems (Bortec Biomedical Ltd., Canada) with a pre-amplification gain of 2,000 and a frequency response of 10-1,000Hz. All data were sampled at 2,000 Hz using a 12-bit data acquisition system (NI6071E, National Instruments, USA). All recordings were rectified and low-pass filtered at 2.5Hz using a 4th order, zero-phase-lag Butterworth filter to compute the linear envelope of EMG signals (Masani, et al., 2009; Vera-Garcia, et al., 2006). The phasic response was determined as the peak EMG value in the 0.5s time window immediately following the perturbation (Masani, et al., 2009). Phasic responses of each muscle were selected as the features for analyzing postural synergies following perturbations.

**2.3.4. Self-Organizing Map (SOM)**

A SOM was used to represent and compare postural muscle synergies, and to model the responses of each muscle. The SOM analysis method represented in Figure 2.1 was implemented in Matlab 7 (MathWorks Inc., USA) using the SOM Toolbox for
Matlab (Vesanto, et al., 2000). Acquired EMG signals were processed and the phasic muscle response feature was selected (see: Data Acquisition). The SOM algorithm had two phases: training phase and recall phase (visualization), which are shown in Figure 2.2.

The input dataset consisting of trunk muscle phasic responses was encoded onto a 2-D map representation during SOM training. In the recall phase the resultant map allowed visualization of postural synergies for each of eight directional perturbations. Each direction was assigned a cluster on the map corresponding to where the responses of that direction converged. The organized map contained similar responses grouped close together. Comparison of the EMG response differences was done by analyzing the relative proximity of the clusters of each perturbation direction on the map. Clusters of perturbation directions that were close together had similar neuromuscular responses and clusters of perturbation directions that were far apart had dissimilar responses. Lastly, from the clusters associated with each direction, an averaged response for each muscle was extracted. A direction-dependent model of the responses for each muscle was then constructed using Gaussian curve-fitting, and compared to the results obtained using conventional EMG analysis. A detailed description follows.

![Diagram](EMG_data_input_SOM_output.png)

**Figure 2.1:** Self-organizing map method for data mining, visualization and model extraction of muscle synergies.
2.3.4.1 SOM Training

During SOM training the input data is encoded onto a 2-D output layer known as the map. SOM uses an unsupervised learning algorithm where the output layer nodes
compete to encode the input data.

The input dataset contains a vector for each of eight perturbation directions for each participant (five trials were averaged for each participant) resulting in 104 input data vectors (13 participants x 8 directions). Each input vector contains ten points corresponding to the phasic responses of each analyzed muscle. The input vectors also contained a label with the corresponding direction of perturbation to allow comparisons of direction-dependent responses. The label was used only during the recall phase to show where the clusters for each direction converged on the resultant map, and was not used during the training phase. The input data was logarithmically normalized before SOM training. Training was performed in batch mode, which provided quicker execution of the algorithm (Kohonen, 2001).

The output layer was defined as a 5x5 hexagonal map. In the output layer each node (indexed by $j = 1, \ldots, 25$) is represented by a weight vector $w_j = (w_{j1}, w_{j2}, \ldots, w_{j10})^T$ with the same dimensionality as the vectors from the input space (i.e., ten points). Weight vectors were initially randomly assigned and during the training they were tuned to represent the input data.

During training, each iteration (indexed by $n$) proceeded by sampling a new input vector $x_n = (x_{n1}, x_{n2}, \ldots, x_{n10})^T$ from the input data, that was matched against nodes in the output layer by calculating the Euclidean distance between a given input vector and each node on the map. The algorithm then selected the node that was the best matching unit (BMU), indexed by $c$. The BMU was the node whose weight representation ($w_c$) was closest to the given input vector as measured by the Euclidean distance (Equation 2.1).

$$c = \arg \min_j (\| x_n - w_j \|)$$  \hspace{1cm} (2.1)
The weight vector of the BMU node was then adapted toward the current input vector (Equation 2.2).

\[ w_j(n + 1) = w_j(n) + h_{cj}(n) \cdot [x_n - w_j(n)] \]  

(2.2)

Neighbouring nodes were proportionally modified via the neighbourhood function attempting to distribute knowledge locally around the BMU. A Gaussian neighbourhood function, \( h_{cj}(n) \), which was centred on the BMU node, controlled the region in the output layer over which training occurred; \( r_c \) and \( r_j \) are the position of the BMU node and an arbitrary node, \( j \), on the map (Equation 2.3).

\[ h_{cj}(n) = \alpha_n \exp \left( -\frac{\|r_c - r_j\|^2}{2\sigma_n^2} \right) \]  

(2.3)

The radius of the neighbourhood (\( \sigma_n \)), as well as the learning rate (\( \alpha_n \)), shrank monotonically as \( n \) increased. The algorithm sampled the input data randomly until the weight vectors converged to a stable projection of the input data (Kohonen, 2001; Vesanto, et al., 2000). After SOM training, the map was encoded with the representation of the input data.

2.3.4.2 SOM Visualization

Visualization produced using the SOM–based method reduced dimensionality, yet maintained non-linear topological relationships (Kohonen, 2001). Visualization was generated by superimposing a histogram of BMU hits for a particular dataset onto the nodes of the map, indicating BMU locations. The resultant map contained groupings of
similar patterns that were discovered in the input data mapped onto spatially proximal nodes, consequently forming clusters of input data. Visualization of the hits histograms was produced for each perturbation direction (shown in Figure 2.3). Such a method of data representation enabled us to compare complex neuromuscular responses caused by different perturbations (i.e., direction-specific postural synergies).

Visualization of all perturbation directions on a single map allowed rapid visual inspection of response differences. The arrows (shown in Figure 2.4a) represent the label indicating direction of perturbation with the highest BMU hits frequency (solid line), and other BMU hits (dotted line), appearing on each node. Clusters of responses for each direction were obtained by assigning each node a “winning” direction based on the highest frequency of BMU hits in that direction (Figure 2.4). Relative proximity of clusters on the map implies similarity or dissimilarity of the overall weight vectors associated with those clusters.

The Euclidian distance measure was calculated from the centre of each cluster to the centre of all other clusters on the map (Equation 2.4). The Euclidian distance between arbitrary clusters \(a(x_a, y_a)\) and \(b(x_b, y_b)\) is:

\[
d_{a-b} = \sqrt{(x_a - x_b)^2 + (y_a - y_b)^2}
\] (2.4)

The average Euclidian distance shows the proximity between clusters, which infers similarity or dissimilarity of direction-dependent postural synergies.

2.3.4.3. Model Extraction

Models of neuromuscular responses were constructed from weight vector
representations of the convergent clusters on the map. Each cluster contained several nodes and weight vector responses that corresponded to each perturbation direction. Average responses for each cluster were calculated to produce one overall response vector for each perturbation direction. The resulting weight vector for each direction ($w_d$, for $d=1-8$), contained ten points representing the average phasic neuromuscular response of each muscle group. Gaussian curve-fitting was used to build a continuous-direction model for each muscle. Each muscle response was analyzed with respect to the perturbation direction to extract direction-dependent models: the relationship between perturbation angle ($x$) and the EMG response ($y$), for each muscle, was described by a Gaussian function (Equation 2.5), where $a$, $b$, and $c$ are the model coefficients:

$$y = a \cdot \exp\left(\frac{-(x-b)^2}{c^2}\right)$$ (2.5)

2.4. Results

2.4.1. Postural Synergy Visualization

Postural synergy representations for each direction were obtained and projected onto the SOM to produce visualization. Visualization based on BMU hits distribution shows the concentration of clusters for each perturbation direction (Figure 2.3). The relative concentration of clusters on the map infers similarities (i.e., clusters closer together) and dissimilarities (i.e., clusters further apart) between direction-dependent postural responses. The same visualization also shows that perturbations to the front and back, as well as the right and left (directions 1 and 5, and 3 and 7, respectively), have symmetrical cluster locations relative to each other (Figure 2.3). Although the absolute position of the clusters on the grid was arbitrary, cluster symmetry indicates that
neuromuscular responses were dissimilar and opposite for opposing perturbation
directions, and that the muscular reactions for front vs. back and right vs. left directions
represented the most dissimilar and opposite postural synergies.

Visualizing the directions for which each node was active on a single SOM map
using direction-indicating arrows allowed a comparison of postural synergies between
perturbation directions (Figure 2.4). It is possible for more than one direction to appear in
the same node (represented as multiple arrow directions on a single node - Figure 2.4a),
indicating similarity of muscle responses for those directions. Each node could also be
assigned to a particular cluster (grouped based on the “winning” direction of individual
nodes - Figure 2.4b) and projected on a single SOM map illustrating each of the eight
perturbation directions clusters. Adjacent perturbation intervals are represented by
clusters that were closest on the map (i.e., perturbation direction 5 appears between
directions 4 and 6 - Figure 2.4b), indicating the most similar responses between these
directions. In addition to similar input data converging to proximal clusters on the map,
the size of the clusters is proportional to the size of data corresponding to those clusters
(Martin and Obermayer, 2009). Analysis of SOM cluster size (variability) was found to
be valid for variability analysis (Lamb, et al., 2008). Our analyses of the SOM map
(Figure 2.3) and the relative cluster size (Figure 2.4) indicate that if the cluster size is
smaller, there is less data associated with that cluster (i.e., perturbation direction) (Martin
and Obermayer, 2009) hence suggesting smaller variability of neuromuscular responses
associated with that perturbation direction. The results suggest that neuromuscular
responses to directions 1 and 5 (i.e., anterior-posterior directions) are less variable
(average cluster size = 2) than other perturbations directions (average cluster size = 3.5).
Figure 2.3: SOM map visualization of postural synergy response for each perturbation direction. The plots show average responses for all subjects for each direction. The analysis represents postural synergies and compares the direction-dependent differences of perturbations by comparing the relative cluster locations.
Figure 2.4: SOM map visualization of mean cluster responses for all perturbation directions. a) Directions of arrows on each node indicate perturbation directions 1-8; solid arrows represent the “winning” direction with the highest responses; dotted arrows represent all other directions with lesser responses on each node. b) Each cluster is colour-coded and labeled with the associated direction number and the corresponding direction-indicating arrow.

The Euclidian distance measured between the central locations of two clusters on the map represents similarity (i.e., small Euclidian distance) or dissimilarity (i.e., large Euclidian distance) of synergistic postural responses associated with those two clusters. Each perturbation direction is presented on the polar plot (Figure 2.5) showing similarities and dissimilarities of that perturbation to all other perturbation directions. As expected, the synergistic responses of each direction were most similar to the responses of those directions adjacent to it and got progressively more dissimilar as perturbations change. The responses were most dissimilar from the perturbations in the opposite direction to it. This analysis compared synergistic difference between direction-dependent postural responses using the SOM method.
Distances between clusters imply neuromuscular differences between the corresponding clusters. Clusters of each perturbation direction are compared to all other perturbations to show the similarities and dissimilarities of direction dependent neuromuscular responses.

2.4.2. **Direction-dependent Models**

Gaussian regression models (Equation 2.5), for each muscle, were computed by extracting and analyzing the average response vector ($w_d$), for each direction ($d=1-8$). Each direction vector contained 10 points representing the average phasic neuromuscular responses of each corresponding muscle. The direction-dependent model for each muscle was computed to compare the results to previous studies, which used different approaches to analyze direction-dependent neuromuscular responses during sitting. The coefficient of determination ($R^2$) was used to assess the goodness of fit of each muscle model to be represented by the Gaussian function and the results obtained using the SOM method are presented in Table 2.1. The $R^2$ values from Masani et al. (2009) are used for evaluation of the results obtained by SOM.
Table 2.1: Results of Gaussian curve-fitting for each muscle. Muscles are recorded bilaterally: left (L) and right (R); a, b, c are the coefficients of Equation 2.5; and $R^2$ is the coefficient of determination which is compared to $R^2$ obtained by Masani et al. (2009).

<table>
<thead>
<tr>
<th>Muscle</th>
<th>a</th>
<th>b</th>
<th>c</th>
<th>$R^2$</th>
<th>$R^2$ - Masani et al. (2009)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rectus Abdominis (RA)</td>
<td>2.029</td>
<td>165.2</td>
<td>105.7</td>
<td>0.9543</td>
<td>0.990</td>
</tr>
<tr>
<td>External Oblique (EO)</td>
<td>2.636</td>
<td>156.2</td>
<td>133.1</td>
<td>0.9781</td>
<td>0.977</td>
</tr>
<tr>
<td>Internal Oblique (IO)</td>
<td>2.579</td>
<td>134.9</td>
<td>146.5</td>
<td>0.9719</td>
<td>0.845</td>
</tr>
<tr>
<td>Thoracic Erector Spinae (T9)</td>
<td>1.801</td>
<td>28.4</td>
<td>219.3</td>
<td>0.7926</td>
<td>0.944</td>
</tr>
<tr>
<td>Lumbar Erector Spinae (L3)</td>
<td>2.407</td>
<td>3.3</td>
<td>322.0</td>
<td>0.7636</td>
<td>N/A</td>
</tr>
</tbody>
</table>

In Figure 2.6 the perturbation directions were converted from directions 1-8 into angles, where: 1=0°/360°, 2=45°, 3=90°, 4=135°, 5=180°, 6=225°, 7=270°, and 8=315°. The responses were sub-divided anatomically to abdominal and back muscle groups. The abdominal muscles (RA, EO and IO) responses yielded a good fit suggesting that the relationship could be represented using a Gaussian function. The back muscles (T9 and L3) responses provided an acceptable fit using normal distribution, suggesting a tendency that the relationship fits a Gaussian function, with the exception of right L3. The models obtained using the SOM method suggests that the amplitude of phasic response to direction of perturbation relationship during sitting may be quantitatively modelled using the Gaussian function. This finding confirms those of Masani et al. (Masani, et al., 2009) and Preuss and Fung (2008), who used standard analysis methods to quantify the relationship.

Coefficient ‘b’ in Table 2.1 indicates where each muscle model had a maximal
response. The abdominal muscles had maximal responses around $180^\circ$ (back perturbation direction) suggesting that opposing muscles stabilized the perturbations. The maximal responses of back muscles were around $0^\circ$ (front perturbation direction), suggesting the opposing muscles also stabilized the perturbation. The anatomy of the back muscles, both of which are located around the spinal column, is consistent with the location of the peak responses.

**Figure 2.6:** Activation pattern for different directions of perturbation for left (thick line) and right (thin line) muscle group for all subjects. Abdominal muscles: rectus abdominis (RA), external oblique (EO), and internal oblique (IO); Back muscles: thoracic erector spinae (T9), lumbar erector spinae (L3).
2.5. Discussion

The purpose of this study was to present and demonstrate an SOM-based method for representing, comparing and modelling of trunk muscle postural synergies following direction-dependent perturbations during sitting. We used the SOM to project and represent the direction-dependent phasic responses of ten muscles on a 2-D map. Furthermore, the SOM method produced an organized visualization, where similar patterns were mapped close together, therefore allowing comparisons of the neuromuscular responses following eight-directional perturbations. Finally, we produced direction-dependent models for each of the ten muscles that were acquired using the SOM method.

2.5.1. Direction-dependent Neuromuscular Responses

The results obtained by visualizations produced using SOM-based analysis provide important insights and allow quick comparisons of the neuromuscular system relevant to studies of complex mechanisms of sitting balance (Bergmark, 1989). Using the SOM visualization cluster position analysis (Figure 2.5) we have quantified direction-dependent differences of trunk muscle phasic responses during sitting, which are necessary to stabilize the trunk (Masani, et al., 2009). Our study found symmetrical and direction-dependent neuromuscular responses which are consistent with findings from the literature (Masani, et al., 2009; Thrasher, et al., 2010). In our study activation patterns for each muscle obtained from the SOM clusters were modelled using the Gaussian distribution (Figure 2.6 and Table 2.1) and demonstrate maximum EMG response in the anatomically opposite direction to the perturbation. The results indicate that opposing
muscular reactions stabilize the body by stiffening the muscles that would provide the forces in the opposite direction to the perturbation. These results complement previous findings demonstrating the role of direction-dependent abdominal muscle responses that can be modelled using Gaussian distribution (Masani, et al., 2009; Preuss and Fung, 2008). Although our R² coefficients are not as high, the difference could also be attributed to the maximal voluntary contraction normalization by Masani et al. (Masani, et al., 2009). Moreover, the anatomy of back muscles (T9 and L3) extends vertically along the sagittal plane of the spine, whereas the abdominal muscles (RA, EO and IO) extend along the sagittal and transverse planes of the trunk. This musculoskeletal geometry could explain why the back muscles exhibit less direction-dependent responses (i.e., have a more active role in stabilizing all perturbation directions), whereas abdominal muscles have more direction-dependency. These results are consistent with the notion that the CNS may be tuning the activation level based on the musculoskeletal geometry (Vasavada, Peterson, and Delp, 2002).

Furthermore, analysis of variability based on the size of SOM clusters (Martin and Obermayer, 2009) (Figures 2.3 and 2.4) and direction-indicating arrows (Figure 2.4a) offer some insights into the variability of neuromuscular responses during sitting. Our results suggest that the responses to anterior-posterior perturbations (i.e., forward and backward perturbations) are less variable than other perturbation responses, which include the medio-lateral component. Variability of postural synergies is a result of neuromuscular redundancy (Torres-Oviedo and Ting, 2007) (i.e., the same movement can be executed by a variety of muscular patterns (Bergmark, 1989) and does not necessarily reflect dysfunction (Safavynia, Torres-Oviedo, and Ting, 2011). Inter-trial variations of
individual muscles are known to be correlated, thus representing a general construct used by the CNS, and this variability may represent variations of neural commands that activate individual muscle synergies (Torres-Oviedo and Ting, 2007). The smaller variability of neuromuscular responses in the anterior-posterior directions could be explained by the anatomy of the trunk muscles (Stokes and Gardner-Morse, 1999), which provides a greater mechanical advantage to resist perturbations in the anterior-posterior direction.

2.5.2. Muscle Synergy Visualization with SOM

The main advantage of SOM is the ability to represent the results pictorially (Pattichis, Schizas, and Mittleton, 1995). An intuitive topological visualization of muscle synergies could aid clinicians in discriminating pathology, assessment of rehabilitation and creating evidence-based interventions (Safavynia, Torres-Oviedo, and Ting, 2011), by comparing responses of individuals (for example with spinal cord injury) to established norms. Visualization of direction-dependent responses could also be used to assess symmetry of muscular responses of patients (with stroke, for example) or for biofeedback training. Furthermore, the capability of the SOM algorithm to encode redundancies in data (Lamb, et al., 2008; Martin and Obermayer, 2009) and relative ease of interpretation (Kohonen, 2001; Pattichis, Schizas, and Mittleton, 1995; Lamb, et al., 2008; Lamb, et al., 2011), which have been cited as a limiting factor for clinical muscle synergy analysis (Safavynia, Torres-Oviedo and Ting, 2011), is another benefit of SOM over other muscle synergy extraction methods (Jensen, et al., 1994; Mah, et al., 1994; Saltiel, et al., 2001; Holdefer and Miller, 2002; Krishnamoorthy, et al., 2003). Foremost,
the unsupervised, self-organizing structure of the SOM is an important feature of the SOM algorithm (Martin and Obermayer, 2009). SOM generates classes of data automatically, consequently allowing discovery of subcategories of data, and making it suitable for exploratory analysis (Kohonen, 2001; Pattichis, Schizas and Mittleton, 1995; Lamb, et al., 2011) or discovery of new patterns, perhaps justifying poorer goodness of fit obtained for the back muscles, and suggest that these muscles exhibit more uniform direction-dependent responses and should be modelled accordingly.

2.6. Conclusions

This study demonstrated the SOM-based analysis of postural synergies of trunk muscles during direction-dependent perturbed sitting. Complex neuromuscular synergies were visualized and compared by encoding large EMG datasets on a single map and quantitative models of each muscle were produced. The results obtained using SOM analysis are consistent with findings obtained by Masani et al. (Masani, et al., 2009) and other studies (Thrasher, et al., 2010; Preuss and Fung, 2008) therefore adding to the validity of the SOM visualization of postural synergies. Although computational methods and SOM analysis have not yet demonstrated their full potential (Martin and Obermayer, 2009), the presented method was capable of encoding, qualitatively comparing and assessing variability of direction-dependent postural muscle synergies during sitting perturbations. The SOM-based analysis has revealed insights into mechanisms of trunk muscles during sitting perturbations, and can be used as a reverse engineering method for visualization of complex neuromuscular systems at a glance. The benefit of SOM-based
analysis is the visualization, which has produced a way for summarizing and comparing postural synergies, despite of their complexity. Future applications will concentrate on encoding larger input sets with temporal information, including a larger selection of muscles, as well as a range of perturbation magnitudes. Comparisons to patient data and individual responses could provide a postural synergy classification method which can be clinically significant for trunk assessment and rehabilitation.

2.7. Acknowledgements

This study received financial support from the National Science and Engineering Research Council of Canada (54196), the Canadian Institute of Health Research (FRN-69003, FRN-97952 and FNR-94018), Toronto Rehabilitation Institute and Ontario Ministry of Health and Long-Term Care.
2.8. References


Chapter 3

3. Anticipation of Direction and Time of Perturbation

Modulates the Onset Latency of Trunk Muscle Responses during Sitting Perturbations

The material presented in this chapter is currently under review in the article:

*NOTE:* The content of this chapter is identical to the material presented in the publication except for the text formatting which was done according to University of Toronto requirements.
3.1. Abstract

When the trunk is perturbed, muscle responses would likely differ if the subject had prior information about the perturbation. The objectives of this study were to identify the responses of trunk muscles to sudden support surface translations and quantify the effects of anticipation of direction and time of perturbation on the trunk neuromuscular responses. Twelve able-bodied individuals participated in the study. Participants were seated on a kneeling chair and support surface translations were applied in the forward and backward directions with and without cues about the direction and time of perturbation. The trunk started moving 46.7 and 37.9 ms following forward and backward perturbations, respectively. During unanticipated perturbations, latencies of the trunk muscle contractions varied between 103.4 and 117.4 ms. When the subject anticipated the perturbations, trunk muscle latencies were reduced on average by 16.8 ms and the time it took the trunk to reach maximum velocity was also reduced, suggesting a biomechanical advantage caused by faster muscle responses. The results suggested that trunk muscles have medium latency responses and use reflexive mechanisms. Moreover, anticipation of direction and time of perturbation decreased trunk muscles latencies, suggesting that the central nervous system modulated readiness of the trunk based on anticipatory information.
Keywords: Anticipation; perturbation; support surface translation; sitting; trunk muscles.
3.2. Introduction

The neuromuscular system of the trunk is mainly responsible for maintaining trunk stability during sitting and standing (Preuss and Fung, 2008). Muscle activations following disruption of quiet sitting (Masani et al., 2009; Milosevic et al., 2012; Preuss and Fung, 2008; Shahvarpour et al., 2015) and standing (Carpenter et al., 2008; Cresswell et al., 1994; Preuss and Fung, 2008; Stokes et al., 2000; Wilder et al., 1996) have previously been studied with the objective to better understand the neural mechanisms responsible for trunk stability. Perturbations were delivered in the form of direct perturbation to the trunk (i.e., pushing or pulling of the trunk) (Kuramochi et al., 2004; Masani et al., 2009; Milosevic et al., 2012; Shahvarpour et al., 2015; Stokes et al., 2000; Wilder et al., 1996) or by perturbing the surface on which the individual was sitting or standing (Cresswell et al., 1994; Preuss and Fung, 2008). During such experiments, the type of perturbation (Carpenter et al., 2008), the direction of perturbation and the posture of the participant prior to the perturbation (i.e., sitting or standing posture) (Preuss and Fung, 2008) play a critical role in the response to the perturbation. The latencies of the trunk muscle activations with respect to the onset of perturbation are always in the range between 25 and 200 ms (Cresswell et al., 1994; Preuss and Fung, 2008; Stokes et al., 2000), suggesting that the trunk neuromuscular system has short latency responses based on monosynaptic reflexes (Cresswell et al., 1994; Granata et al., 2004) or medium latency responses based simple polysynaptic reflexes (Stokes et al., 2000).

Complexity of the central nervous system (CNS) control of the trunk is revealed in situations when perturbations can be anticipated. Anticipatory postural adjustments are
often investigated during rapid limb movements, and those studies revealed that anticipation can lead to stiffening of the joints and adjustment of the initial posture before the onset of the limb movement (Allison et al., 2003; Cresswell et al., 1994). In case of external perturbations, such as those delivered by direct perturbations of the trunk or by perturbing the surface on which the subject is standing or sitting, neuromuscular responses can be studied with and without the anticipation of perturbation to investigate the effects of anticipating a perturbation. During such perturbations, there are two types of anticipation: i) spatial anticipation - which is the prediction of type of perturbation or direction of perturbation; and ii) temporal anticipation - which is the prediction of the time of perturbation (Wilder et al., 1996). Wilder et al. (1996) reported that the trunk muscle onset response times were affected by temporal anticipation during standing balance. However, they did not investigate the effects of spatial and temporal anticipation systematically. Marras et al. (1987) found that expectation of the time of perturbations increased the exerted trunk muscle forces during standing perturbations. Aleksiev et al. (1996) found that trunk muscle amplitudes during standing were affected by expectation of the time of perturbation. Similarly, it was reported that neck muscles response amplitudes were increased when direct perturbations to the head were applied (Kuramochi et al., 2004). The effect of anticipation of the perturbation on the trunk neuromuscular responses during sitting is still not well understood. To our knowledge no study has investigated the effect of anticipation of direction and time of perturbation on the neuromuscular responses during sitting support surface translations. Therefore, a systematic investigation of the neuromuscular responses during sitting balance support
surface translations, when direction and time of the perturbation could be anticipated and not anticipated, is required.

We hypothesized that anticipation of direction and time of perturbation will modulate trunk muscle responses. The objectives of this study were to identify the responses of the trunk muscles to sudden support surface translations during sitting and to quantify the effects of anticipation of direction of perturbation and time of perturbation on modulation of the trunk neuromuscular responses.

2. Methods

2.1. Participants

Twelve healthy, male individuals participated in this study. The age, weight and height of participants were 26.8 ± 3.3 years, 64.7 ± 7.8 kg, and 171.6 ± 7.8 cm (mean ± SD), respectively. None of the participants had history of neurological or musculoskeletal impairments. Informed consent was obtained from all individual participants included in the study in accordance with the principles of the Declaration of Helsinki. The experimental procedures were approved by the local institutional ethics committee.

2.2. Experimental Protocol

Participants were seated on a kneeling chair and were instructed to maintain a relaxed upright posture of the trunk while keeping their arms crossed on their chest
Perturbations in the forward or backward direction were applied as support surface translations using an instrumented treadmill FIT (Bertec, USA). Perturbations were delivered with or without spatial and temporal cues (i.e., direction and time of perturbation, respectively) in the following conditions: i) both direction and time of the perturbation could not be anticipated (D–T–); ii) the direction could not be anticipated, but the time of the perturbation could be anticipated (D–T+); iii) the direction could be anticipated, but the time of perturbation could not be anticipated (D+T–); and iv) both direction and time of the perturbation could be anticipated (D+T+). In order to examine if the subjects contracted their trunk muscles before the perturbation when the perturbation could be anticipated or if the anticipation only affected the reactive responses, two catch trial conditions were also incorporated: v) the direction could not be anticipated and the time of perturbation could be anticipated, but the perturbation was not delivered (Catch–); and vi) both direction and time of perturbation could be anticipated but the perturbation was not delivered (Catch+). In total, 192 randomly ordered trials were recorded for each participant, including 16 repeated trials for each of the six conditions for the forward and backward perturbation direction (i.e., 16 trials x 6 conditions x 2 directions). Before recording the experimental data, participants were given an opportunity to become familiarized with the experimental procedure. They were perturbed six times in different directions, and these data were not used in the analysis. This was done to ensure that the initial learning of a new task is not contaminating the experimental results. To prevent fatigue, data was collected in four sessions (48 trials per session) with a 5 min break between sessions. Each of the four sessions lasted approximately 15 min. The direction of perturbation (i.e., spatial cue) was indicated to the participant prior to delivering
perturbations using verbal instructions. During one session only forward perturbations were delivered and during another session only backward perturbations were delivered. In the remaining two sessions the perturbations were delivered in the forward and backward direction and the participant could not anticipate which perturbation will be next. The time of perturbation (i.e., temporal cue) was indicated to the participant in all four sessions, during the conditions with anticipated perturbation time, using an audio signal 1 - 3 sec before the perturbation. The triangle-shaped velocity perturbation was applied over a period of 120 ms. The resultant average perturbation displacement during all trials was $7.3 \pm 1.6$ cm and the peak acceleration was $12.2 \pm 4.8$ cm/sec$^2$ (mean ± SD). After each perturbation was delivered, the treadmill was slowly returned to the starting position and the next perturbation trial was started after 5 - 7 sec.
2.3. **Data Acquisition**

2.3.1. *Trunk Muscle Electromyography and Force Signal Measurements*

Trunk muscle activity was recorded using surface electromyography (EMG) unilaterally on the right side of the body, assuming that the participant has symmetric responses (Masani et al., 2009; Milosevic et al., 2012). Disposable EMG electrodes (Ag-AgCl) were placed with 1 cm separation between the electrodes on the abdominal muscles: rectus abdominis, 3 cm right and 1 cm superior to the umbilicus (RA-1) and rectus abdominis, 3 cm right and 1 cm inferior to the umbilicus (RA-2); as well as erector muscles: thoracic erector spinae, 5 cm right of the T9 spinous process (T9) and lumbar erector spinae, 3 cm right of the L3 spinous process (L3). A reference electrode was
placed over the clavicle. Data was acquired using a surface EMG system Bagnoli-8 (Delsys Inc., USA) with a pre-amplification gain of 1,000 and a The frequency response of the signal was between 20 and 450 Hz, which is in the range of the previous studies examining trunk muscle onset latencies during perturbations (Carpenter et al., 2008; Granata et al., 2004; Stokes et al., 2000) and effectively removes motion artifact during fast perturbations while preserving EMG signal energy (De Luca et al., 2010).

The anterior-posterior ground reaction force on the treadmill surface was also recorded using the instrumented treadmill FIT (Bertec, USA) to identify the exact start of the perturbation. All EMG data as well as the force outputs were recorded using a 12-bit data acquisition system NI6071E (National Instruments, USA) with 2,000 Hz sampling rate.

2.3.2. Trunk Center of Mass Measurements

Three-dimensional trunk kinematics were collected with 6 Oqus cameras (Qualisys Motion Capture Systems, Sweden), to assess postural performance of the upper body after perturbations. A total of 24 passive reflective markers were placed on the following upper body segments: i) head (to C7); ii) upper thoracic segment including arms (T1–T6); iii) lower thoracic segment (T7–T12); iv) lumbar trunk segment (L1–L5); and v) pelvis segment. A detailed description of the marker placements is described in Crosbie et al. (1997) and has subsequently been used by Preuss and Fung (2008) during setting and standing perturbations. An additional marker was placed on the treadmill to record the perturbation platform position. All marker positions were recorded in the absolute coordinate frame and were sampled at 200 Hz using the Qualisys Track Manager.
Software (Qualisys Motion Capture Systems, Sweden). The trunk was divided into the head, abdomen, thorax (i.e., combining the measured upper and lower thorax segments) and the pelvis segments. Then a combined upper body center of mass (COM) of the trunk in the anterior-posterior direction was calculated according the established methods described by Winter (2009).

### 3.3.4. Data Analysis

For all trials, the start of the perturbation (t = 0) was defined as the time when the anterior-posterior force component on the treadmill exceeded 15 N.

#### 3.3.4.1. EMG Responses

Only the primary muscle responses were examined in the EMG analysis. Perturbations were applied as forward or backward support surface translations, resulting in backward or forward trunk bending, respectively (Figure 1). Therefore, during forward perturbations (i.e., backward trunk bending), RA-1 and RA-2 muscle responses were analyzed and during backward perturbations (i.e., forward trunk bending) T9 and L3 muscle responses were analyzed. All EMG data were first rectified by taking the absolute value of the signal. Analysis of the EMG signals included computing: 1) the tonic muscle activity, which was defined as the root mean square of the EMG signal in the 50 ms window before the perturbation; 2) the onset time of the muscle response, which was automatically identified using the integrated protocol (IP) algorithm (Santello et. al., 1998; Allison 2003); and 3) phasic muscle activity, which was defined as the root mean square of EMG signal in the 100 ms window after the muscle onset.
**Figure 3.2:** Example of the trunk muscle responses and the chair and center of mass (COM) movements during: a) forward perturbation and b) backward perturbation. Shown is the activity when both direction and time of perturbation could not be anticipated ($D^- T^-$) and both direction and time of perturbation could be anticipated ($D^+ T^+$) for the rectus abdominis muscle superior to the umbilicus (RA-1), rectus abdominis muscle inferior to the umbilicus (RA-2) in blue, and the thoracic erector spinae muscle (T9) and lumbar erector spinae muscle (L3) in red.
3.3.4.2. COM Responses

The trunk COM responses were evaluated to characterize the movement of the upper body and evaluate postural performance. The trunk COM position was low-pass filtered at 50 Hz using a fourth-order, zero-phase-lag Butterworth filter. Analysis of the COM responses included calculating: 1) the trunk movement onset, which was determined using the IP algorithm (Santello et. al., 1998; Allison, 2003); 2) the trunk maximum displacement, which was determined as the maximum displacement of the COM relative to the chair in the 500 ms window after the perturbation; 3) the time to maximum displacement, which was the time it took the COM to reach the maximum displacement; 4) the trunk maximum velocity, which was determined as the maximum velocity of the COM in the 500 ms window after the perturbation; and 5) the time to maximum velocity, which was the time it took the COM to reach maximum velocity.

3.3.5. Missing Data

For all data during the experiment, 7% of trials for RA-1 muscle, 5% of trials for RA-2 muscle, 17% of trials for T9 muscle, and 11% of trials for L3 muscle were rejected when the muscle onset time was identified as an outlier and it was not possible to determine the onset of muscle activity due to excessive co-activation of muscles. For kinematics data, 9% of all data was rejected when the markers required to calculate the COM were missing and could not be interpolated during post processing.
3.3.6. Statistical Analysis

Comparisons between experimental conditions for the EMG and COM analyses were performed using the one-way repeated measures analysis of variance (ANOVA) with Tukey post-hoc multiple comparisons when a significant difference was found on the ANOVA test. Significance level was set to $p<0.05$.

3.4. Results

An example of the EMG and COM data is shown in Figure 1. The catch trials (i.e., Catch$^-$ and Catch$^+$) were compared to the main experimental conditions (i.e., D$^-$ T$^-$, D$^-$ T$^+$, D$^+$ T$^-$ and D$^+$ T$^+$) only for the analysis of the tonic muscle activity to confirm that muscles were not preloaded before the perturbations. Once this was confirmed, only the main experimental conditions were compared, as there were no phasic activities during the catch trials.

3.4.1. Trunk Muscle Responses

3.4.1.1. Tonic EMG Activity

There were no differences in the tonic muscle activity between experimental conditions and catch trials for the RA-1 ($p=0.533$), RA-2 ($p=0.366$), T9 ($p=0.204$) and L3 ($p=0.140$) muscles.
3.4.1.2. EMG Onset Latency

The onset times were significantly different between experimental conditions for RA-1 ($p<0.001$), RA-2 ($p<0.001$), T9 ($p<0.001$) and L3 ($p<0.001$) muscles (Table 1). Tukey post-hoc multiple comparisons indicated that when both direction and time of perturbation were anticipated, trunk muscle onset latencies were shorter for all muscles. Anticipation of the direction and the time of perturbation seem to both contribute to the shorter onset latencies (Table 1).

3.4.1.3. Phasic EMG Activity

There were no differences in the phasic muscle activity between experimental conditions for the RA-1 ($p=0.612$), RA-2 ($p=0.700$), and T9 ($p=0.300$) muscles, while there was a significant difference for L3 ($p=0.005$). Tukey post-hoc multiple comparisons indicated that there was a smaller phasic muscle activity in the L3 muscle only when both direction and time of perturbation were anticipated.
Table 3.1: Analysis of the onset latencies of trunk muscles for: rectus abdominis superior to the umbilicus (RA-1) and rectus abdominis inferior to the umbilicus (RA-2) during forward perturbations, and thoracic erector spinae (T9) and lumbar erector spinae (L3) during backward perturbations. Results show the mean ± S.D. for each muscle of twelve participants. One-way repeated measures analysis of variance (ANOVA) with Tukey post-hoc was used to compare results in four conditions. Condition 1: both direction and time of the perturbation could not be anticipated (D– T–); Condition 2: direction could not be anticipated, but time of the perturbation could be anticipated (D– T+); Condition 3: direction could be anticipated, but time of perturbation could not be anticipated (D+ T–); and Condition 4: both direction and time of the perturbation could be anticipated (D+ T+).

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<tbody>
<tr>
<td>RA-1</td>
<td>115.0 ± 9.7</td>
<td>110.4 ± 10.0</td>
<td>105.0 ± 8.2</td>
<td>99.1 ± 8.3</td>
<td>*</td>
<td>1-2; 1-3; 1-4; 3-4</td>
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<td>RA-2</td>
<td>116.6 ± 11.0</td>
<td>113.8 ± 11.4</td>
<td>109.0 ± 10.0</td>
<td>99.8 ± 9.8</td>
<td>*</td>
<td>1-3; 1-4; 2-4</td>
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<tr>
<td>T9</td>
<td>103.4 ± 8.5</td>
<td>96.1 ± 10.7</td>
<td>95.3 ± 12.5</td>
<td>90.2 ± 10.8</td>
<td>*</td>
<td>1-2; 1-3; 1-4; 2-4</td>
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<tr>
<td>L3</td>
<td>117.4 ± 13.8</td>
<td>108.3 ± 17.6</td>
<td>109.6 ± 20.3</td>
<td>96.1 ± 17.3</td>
<td>*</td>
<td>1-2; 1-4; 2-4; 3-4</td>
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* p < 0.01
3.4.2. Trunk Kinematics Responses

3.4.2.1. Onset of COM Movements

There were no differences in the trunk movement onset time between experimental conditions for the forward perturbations ($p=0.938$) and backward perturbations ($p=0.839$) (Table 2).

3.4.2.2. Maximum COM Displacement

There were no differences in the maximum trunk displacement between conditions for the forward perturbations ($p=0.357$) and backward perturbations ($p=0.139$). Also, the time to maximum displacement was not different between experimental conditions for the forward perturbations ($p=0.973$) and backward perturbations ($p=0.711$) (Table 2).

3.4.2.3. Maximum COM Velocity

There were no differences in the maximum trunk velocity between conditions for the forward perturbations ($p=0.133$) and backward perturbations ($p=0.489$). However, the time to maximum displacement was significantly different between experimental conditions for the forward perturbations ($p=0.003$) and backward perturbations ($p=0.007$) (Table 2). Tukey post-hoc multiple comparison analysis indicated that when both direction and time of perturbation were anticipated, the time required to reach the maximum trunk velocity was faster during both the forward and backward perturbations. Anticipation of the direction and the time of perturbation both seem to have contributed to this result slightly (Table 2).
Table 3.2: Analysis of the center of mass (COM) kinematics. Results show the mean ± S.D. of each variable for twelve participants. One-way analysis of variance (ANOVA) with Tukey post-hoc was used to compare results during forward (F) and backward (B) perturbations in four conditions. Condition 1: both direction and time of the perturbation could not be anticipated (D– T–); Condition 2: direction could not be anticipated, but time of the perturbation could be anticipated (D– T+); Condition 3: direction could be anticipated, but time of perturbation could not be anticipated (D+ T–); and Condition 4: both direction and time of the perturbation could be anticipated (D+ T+).

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<tr>
<td><strong>Trunk movement onset (ms)</strong></td>
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<tr>
<td>F</td>
<td>46.3 ± 5.5</td>
<td>47.2 ± 4.6</td>
<td>46.4 ± 8.6</td>
<td>46.8 ± 6.7</td>
<td>n.s.</td>
</tr>
<tr>
<td>B</td>
<td>37.8 ± 5.1</td>
<td>37.9 ± 5.1</td>
<td>38.4 ± 4.9</td>
<td>37.5 ± 5.2</td>
<td>n.s.</td>
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<tr>
<td><strong>Max. trunk displacement (cm)</strong></td>
<td></td>
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<tr>
<td>F</td>
<td>4.9 ± 0.6</td>
<td>4.9 ± 0.5</td>
<td>4.7 ± 0.5</td>
<td>4.7 ± 0.5</td>
<td>n.s.</td>
</tr>
<tr>
<td>B</td>
<td>5.1 ± 1.0</td>
<td>5.4 ± 0.9</td>
<td>5.2 ± 0.8</td>
<td>5.0 ± 0.8</td>
<td>n.s.</td>
</tr>
<tr>
<td><strong>Time to max. trunk displacement (ms)</strong></td>
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<tr>
<td>F</td>
<td>327.3 ± 150.0</td>
<td>309.4 ± 172.0</td>
<td>332.4 ± 210.9</td>
<td>328.9 ± 173.7</td>
<td>n.s.</td>
</tr>
<tr>
<td>B</td>
<td>372.9 ± 243.8</td>
<td>385.4 ± 222.5</td>
<td>361.6 ± 186.0</td>
<td>358.9 ± 199.2</td>
<td>n.s.</td>
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<tr>
<td><strong>Max. trunk velocity (cm/s)</strong></td>
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<tr>
<td>F</td>
<td>45.4 ± 4.2</td>
<td>44.7 ± 3.4</td>
<td>44.2 ± 3.3</td>
<td>44.3 ± 3.8</td>
<td>n.s.</td>
</tr>
<tr>
<td>B</td>
<td>45.2 ± 2.4</td>
<td>46.1 ± 3.0</td>
<td>46.1 ± 2.4</td>
<td>45.4 ± 3.3</td>
<td>n.s.</td>
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<tr>
<td><strong>Time to max. trunk velocity (ms)</strong></td>
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<tr>
<td>F</td>
<td>104.1 ± 6.7</td>
<td>105.0 ± 5.0</td>
<td>94.9 ± 5.6</td>
<td>96.0 ± 6.1</td>
<td>* 1-3; 1-4; 2-3; 2-4</td>
</tr>
<tr>
<td>B</td>
<td>99.3 ± 4.3</td>
<td>98.1 ± 5.0</td>
<td>95.7 ± 3.1</td>
<td>93.2 ± 5.0</td>
<td>* 1-3; 1-4; 2-4</td>
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* p < 0.01; n.s. not significant
3.5. Discussion

3.5.1. Trunk Response Latencies

The average abdominal muscle response latencies in our study (i.e., RA-1 and RA-2) varied between 99.1 and 116.6 ms during forward perturbations and erector muscles response latencies (i.e., T9 and L3) varied between 90.2 and 117.4 ms during backward perturbations, while the onset of the trunk center of mass movement was on average 46.7 ms after the forward perturbations and 37.9 ms after the backward perturbations. Effectively this suggests that trunk muscles activated approximately 60 ms after the movement of the trunk. During postural responses to a perturbation, a number of phasic muscle responses occur to compensate for the perturbation (Schmidt, 2011). They include: the monosynaptic stretch reflex (M1) with muscle onset latencies between 30 - 50 ms; functional polysynaptic stretch reflex (M2) with the muscle onset latencies between 50 - 80 ms; triggered reactions with muscle onset latencies between 80 - 120 ms; and voluntary reactions (M3) with muscle onset latency between 120 - 180 ms (Schmidt, 2011; Wilder et al., 1996). It was previously suggested that abdominal muscle responses to support surface perturbations are likely automatic postural responses triggered as part of a muscle synergy along with other proximal muscles (Carpenter et al., 2008). Stokes et al. (2000) have reported that the trunk muscle responses are likely regulated by monosynaptic reflexes (probably stretch reflex) and medium latency reflexes during trunk loading. The responses to support surface perturbations observed in our study can likely be classified as medium latency reflexes, as they are too long to be monosynaptic. This
means that simple polysynaptic reflexes regulate the trunk neuromuscular system in response to sudden support surface translation perturbations. The polysynaptic reflex responses originate from the muscle spindles and are generally far stronger and more flexible than the monosynaptic reflexes, as well as more involved in movement compensation (Wilder et al., 1996). The polysynaptic reflexes are autogenetic, meaning that they are self-generated, but prior instructions could change the response because the reflex travels to the higher centers of the motor cortex and the cerebellum during the response (Schmidt 2011).

Average response latencies of the trunk muscles following support surface translations varied between 70 - 250 ms during sitting and 100 - 200 ms during standing (Preuss and Fung, 2008). Our findings generally agree with this body of literature. However, during direct trunk perturbations applied via a chest harness, reported literature suggested faster trunk muscle responses, ranging between 24 and 68 ms (Cresswell et al., 1994) and 30.7 ms (Granata et al., 2004). These can be classified as monosynaptic reflexes. These faster responses may be caused by direct loading of the trunk via a chest harness which can elicit cutaneous afferent reflexes, not present during support surface translations (Carpenter et al., 2008). Hence, the type of perturbation seems to influence the responses latencies of the trunk muscles.

3.5.2. Anticipation of Perturbation

Response latencies of the RA-1, RA-2, T9 and L3 muscles were 115.0, 116.6, 103.4 and 117.4 ms, respectively, when the direction and time of perturbation could not be anticipated and decreased to 99.1, 99.8, 90.2 and 96.1 ms, respectively, when both the
direction and time of the perturbation could be anticipated. Therefore anticipation of
direction and time of perturbation resulted in 16.8 ms average faster trunk muscle
responses, which is a significant improvement. These results suggest that the anticipation
of the direction and time of perturbation decreases trunk muscle response latencies,
making them more reflexive. The improved reaction time is indicative of a more efficient
response strategy by the central nervous system because it indicates that the system can
respond quicker to reduce the load on the spinal disks more effectively which could in
turn reduce the risk of injury during balance disturbances (Schmidt 1991; Wilder et al.,
1996). Our results also indicate that the time for the trunk to reach the maximum velocity
was faster when the perturbation could be anticipated. Therefore, earlier activations of
trunk muscles lead to the biomechanical advantage of being able to reach the maximum
trunk velocity faster, which may be advantageous in reducing the load on the trunk during
anticipated perturbations.

Previous studies showed that expectation of a sudden load increased average trunk
muscle forces (Marras et al., 1987) as well as the amount of muscle activation (Aleksiev
et al., 1996). During perturbations applied to the head, Kuramochi et al. (2004) reported
that neck muscles response amplitudes were greater when the perturbation could be
anticipated, but reaction times were not affected. These results differ from our findings.
This is probably because the perturbation and the experimental paradigms were different
from our study. However, when the trunk muscles were preloaded before the
perturbations, post-perturbation muscle onset latencies were shown to decrease
(Shahvarpour et al., 2015). Our results demonstrated that trunk muscle response latencies
decreased when the perturbation could be anticipated, despite that the muscles were not
preloaded before the perturbation. Since the exact triggering pathways of the trunk muscle reflexes are unknown, it is difficult to speculate why the response latencies were faster with anticipatory information. It was previously suggested that pre-activation of trunk muscles increased the activity of the gamma system, which could have increased the sensitivity of the muscle spindles and the response to their sudden stretching (Stokes et al., 2000). Perhaps an increase in the sensitivity of muscle spindles with anticipatory information triggers the sub-threshold activation of supraspinal neural circuit (i.e., the cerebellum and the motor cortex), which is responsible for decreasing the response latencies of the trunk muscles during sitting perturbations. Overall, the observed influence of the prior information (i.e., anticipation of perturbation) on the trunk muscle reflex responses suggests that the central nervous system can modulate the “readiness” of the trunk using anticipatory information during sitting balance perturbations.

3.6. Acknowledgements

The authors would like to thank Dr. Daichi Nozaki from the University of Tokyo for allowing us to use his facilities to conduct the experiments. Matija Milosevic was supported by the Natural Science and Engineering Research Council of Canada (NSERC) Postgraduate Scholarships (PGS-D). This work was supported by the Canadian Institutes of Health Research (CIHR) Grant # MOP-97952 and NSERC Discovery Grant #249669. The authors acknowledge the support of Toronto Rehabilitation Institute – University
Health Network who receives funding under the Provincial Rehabilitation Research Program from the Ontario Ministry of Health and Long-Term Care.
3.7. References


4. Trunk Control Impairment is Responsible for Postural Instability during Quiet Sitting in Individuals with Cervical Spinal Cord Injury

The material presented in this chapter has been published in the article:


**NOTE:** The content of this chapter is identical to the material presented in the publication except for the text formatting which was done according to University of Toronto requirements.
4.1. Abstract

Individuals with cervical spinal cord injury usually sustain impairments to the trunk and upper and lower limbs, resulting in compromised sitting balance. The objectives of this study were to: 1) compare postural control of individuals with cervical spinal cord injury and able-bodied individuals; and 2) investigate the effects of foot support and trunk fluctuations on postural control during sitting balance. Ten able-bodied individuals and six individuals with cervical spinal cord injury were asked to sit quietly during two 60 s trials. The forces exerted on the seat and the foot support surfaces were measured separately using two force plates. The global center of pressure sway was obtained from the measurements on the two force plates, and the sway for each force plate was calculated individually. Individuals with spinal cord injury had at least twice as large global and seat sways compared to able-bodied individuals, while foot support sway was not significantly different between the two groups. Comparison between global and seat sways showed that anterior-posterior velocity of global sway was larger compared to the seat sway in both groups. Postural control of individuals with cervical spinal cord injury was worse than that of able-bodied individuals. The trunk swayed more in individuals with spinal cord injury, while the stabilization effect of the feet did not differ between the groups. Foot support affected anterior-posterior fluctuations in both groups equally. Thus, trunk control is the dominant mechanism contributing to sitting balance in both able-bodied and spinal cord injury individuals.
**Key words:** sitting balance; postural sway; center of pressure (COP); spinal cord injury (SCI); foot support; trunk control.
4.2. Introduction

Individuals with spinal cord injury (SCI) often experience motor and/or sensory impairment below the level of injury. Cervical injuries lead to impairment in upper limb, trunk, and lower limb muscles, while thoracic injuries lead to impairment in trunk and lower limb muscles (Seelen et al., 1998). Individuals with cervical or thoracic injuries often have impaired sitting balance (Chen et al., 2003; Grangeon et al., 2012). Impaired sitting balance after SCI causes an individual to alter his/her sitting strategy (e.g. individuals with SCI tilt their pelvis in order to achieve greater stability during sitting) and results in compromised sitting posture. Instability during sitting may affect performance of activities of daily living such as reaching and object manipulation (Chen et al., 2003), and could result in secondary health complications such as pressure sores (Minkel, 2000). Continuous, tonic activation of trunk muscles is required to maintain upright sitting posture, and phasic, feedback-driven activations are required to respond to balance disturbances (Masani et al., 2009). Therefore, paralysis of trunk muscles is one of the main reasons for compromised sitting balance after SCI (Minkel, 2000). Individuals with SCI often use innervated, non-postural muscles (e.g. shoulder and neck muscles), to compensate for the sitting impairment by voluntarily contracting these muscles to regulate sitting balance (Seelen et al., 1998). They also use their arms to increase the base of support during sitting, which can help improve their stability (Grangeon et al., 2012). Despite the compensatory methods that an individual with SCI may use, their sitting balance still remains suboptimal.

Center of pressure (COP) sway during quiet standing and quiet sitting has been utilized to assess postural stability during standing balance (Prieto et al., 1996; Vette et
and sitting balance among able-bodied individuals (Dean et al., 1999; Gilsdorf et al., 1990; Kerr and Eng, 2002; Vette et al., 2010). Such assessments are relatively easy to perform in the laboratory, as they do not threaten the stability of the participants, and as such, can be applicable to evaluate sitting balance of individuals with trunk instability resulting from SCI. A small number of studies evaluated sitting balance of individuals with SCI using COP recordings (Chen et al., 2003; Grangeon et al., 2012; Grangeon et al., 2013; Shirado et al., 2004). These studies showed that postural sway is larger among individuals with SCI compared to able-bodied individuals, indicating worse postural stability and compromised sitting balance after a SCI (Grangeon et al., 2012; Shirado et al., 2004). Interestingly, the research indicates there are no differences in postural sway between individuals with low and high thoracic SCI (Chen et al., 2003). However, in these studies the effect of foot support was either ignored (Chen et al., 2003; Shirado et al., 2004) or not analyzed in detail (Grangeon et al., 2012, 2013).

It is recognized that foot support affects sitting balance. For example, in able-bodied individuals both COP displacement and velocity increased by as much as 70% during forward reaching when the subjects were allowed to use foot support compared to reaching without foot support (Kerr and Eng, 2002). Footrests also increased trunk displacement during forward reaching among individuals with thoracic SCI (Potten et al., 2002). However, the effect of foot support on postural stability of individuals with SCI during quiet sitting has yet to be examined in the literature. In prior studies, Chen et al. (2003) and Shirado et al. (2004) used one force plate positioned under the buttocks with both feet supported on the ground, but they did not account for the effects of foot support on COP.
Postural control is the ability to maintain balance. Various factors, including foot support and trunk control, contribute to postural control during sitting balance. Trunk control is the ability to control the trunk, which can be evaluated by analyzing the trunk fluctuations using COP measures obtained from a force plate placed under the buttocks. Similarly, foot support can be evaluated using postural sway fluctuations obtained using a force plate placed under the feet. Grangeon et al. (2012, 2013) used two force plates, one placed under the buttocks on the seat and the other one under the feet, to calculate the COP, but they did not analyze the separate contributions from the foot support and trunk fluctuations. This is likely because individuals with SCI typically do not have full voluntary control of lower limbs. Consequently, the utility of their foot support is often considered marginal despite the fact that it has been shown that foot support provides a significant contribution during transfers in individuals with SCI (Gagnon et al., 2008).

We hypothesized that individuals with cervical SCI will have worse sitting balance compared to able-bodied individuals, and that foot support will have a positive impact on postural stability. The objectives of this study were to: 1) compare postural control of individuals with SCI with able-bodied individuals; and 2) investigate the effects of foot support and trunk fluctuations on the postural control of the entire body during quiet sitting balance.
4.3. Methods

4.3.1. Participant Recruitment

Able-bodied individuals and individuals with SCI were recruited to participate in this study. In order to participate, all participants had to have the ability to maintain unsupported sitting. Individuals were recruited in the able-bodied group if they had no history of neurological impairment or musculoskeletal injury that could affect their sitting balance. Individuals were recruited in the SCI group if they had either motor incomplete or complete, sensory incomplete or complete cervical SCI, and were minimum one year post injury. All participants gave written informed consent in accordance with the Declaration of Helsinki. The experimental procedures used in this study were approved by the local institutional research ethics board.

4.3.2. Experimental Protocol

Participants were seated in upright sitting posture on a height-adjustable chair without back support and with their feet on the ground in all trials. The seating surface and the foot support surface were each instrumented with a force plate (AccuSway\textsuperscript{Plus}, Advanced Mechanical Technology Inc., Watertown, USA). A thin foam cover was placed over the seat surface to prevent risks of skin injury during data collection in both able-bodied and SCI groups. The height of the chair was adjusted such that the knee angle was at approximately $90^\circ$. Each participant was asked to keep a steady sitting balance with his or her arms crossed over their chest and with their eyes open, as illustrated in Figure 4.1.
**Figure 4.1:** Experimental setup for sitting balance utilizing two force plates. COP$_S$(x,y) captured trunk sway on the seat surface and COP$_F$(x,y) captured foot support sway on the ground. Vertical forces (F$_{zS}$ and F$_{zF}$) and AP and ML forces (not shown) were also captured. The origin, O(0,0,0), of the global coordinate frame, which was used to calculate global COP was placed the middle of the force plates, between the seat and foot support surface on the seat surface, where the seat and foot support surface were aligned along the x and y axis and only separated by distance h which was the height difference between the seat and foot support surface along the z axis.
4.3.3. Measurements

Signals were recorded over two 60 s trials using two force plates (Figure 4.1). Seat COP: \( \text{COP}_S(x,y) \) was calculated from the force plate on the seating surface. Foot support COP: \( \text{COP}_F(x,y) \) was calculated from the force plate on the ground. Global COP: \( \text{COP}_G(x,y) \) was then computed from \( \text{COP}_S(x,y) \) and \( \text{COP}_F(x,y) \) as described in the next section. Moreover, force components measured using the seat force plate \( (F_{xS}, F_{yS}, F_{zS}) \) and the foot support force plate \( (F_{xF}, F_{yF}, F_{zF}) \) were collected. The \( x \) and \( y \) denote the respective coordinates of medial-lateral (ML) and anterior-posterior (AP) directions and \( z \) denotes the vertical direction, as shown in Figure 4.1. Radial distance (RD) time series was calculated as \( RD = \sqrt{AP^2 + ML^2} \) to represent the combined AP and ML COP position according to Prieto et al. (1996). The AP and ML time series were used to examine the specific sway directions and the RD measure was used to examine the overall COP sway. All data were sampled at 500 Hz using a 12-bit data acquisition system (NI 6071E, National Instruments, Austin, USA). A low-pass filter with a cut-off frequency of 5 Hz was applied to all signals (Prieto et al., 1996; Vette et al., 2010).

4.3.4. Global COP Calculation

The seat and the foot support force plate surface were aligned along the \( x \) and \( y \) axis and the seat surface was higher than the foot support surface along the \( z \) axis (i.e. distance \( h \)), as shown in Figure 4.1. The origin \( O(0,0,0) \) for all calculations was positioned on the seat surface in the middle of the seat and foot support force plates. The sum of all moments around the origin, \( M(x,0) \) and \( M(0,y) \), in the ML and AP directions was calculated in Equation 4.1:
\[
\sum M(x,0) = (F_{zs} + F_{zf}) \cdot COP_G(0,y) = F_{zs} \cdot COP_S(0,y) + F_{zf} \cdot COP_F(0,y) + F_{yf} \cdot h,
\]
\[
\sum M(0,y) = (F_{zs} + F_{zf}) \cdot COP_G(x,0) = F_{zs} \cdot COP_S(x,0) + F_{zf} \cdot COP_F(x,0) + F_{xf} \cdot h.
\] (4.1)

where \( h \) was the height difference between the seat and foot support plates Contributions of shear forces due to the height difference between the seat and foot support plates (i.e. \( F_y \cdot h \) and \( F_x \cdot h \)) were included because they could be extensive during various sitting manoeuvres (Gilsdorf et al., 1990). Global COP for sitting was calculated by re-arranging Equation 4.1 to obtain \( COP_G(x,y) \) as a function of measured parameters in Equation 4.2:

\[
COP_G(x,0) = COP_S(x,0) \cdot \frac{F_{zs}}{F_{zs} + F_{zf}} + COP_F(x,0) \cdot \frac{F_{zf}}{F_{zs} + F_{zf}} + h \cdot \frac{F_{xf}}{F_{zs} + F_{zf}},
\]
\[
COP_G(0,y) = COP_S(0,y) \cdot \frac{F_{zs}}{F_{zs} + F_{zf}} + COP_F(0,y) \cdot \frac{F_{zf}}{F_{zs} + F_{zf}} + h \cdot \frac{F_{yf}}{F_{zs} + F_{zf}}.
\] (4.2)

In case when \( h = 0 \), \( COP_G(x,y) \) gives a global COP for dual force plates, previously described by Winter et al., (1998).

### 4.3.5. Data Analysis

The \( COP_G(x,y) \), \( COP_S(x,y) \) and \( COP_F(x,y) \) postural sway fluctuations were analyzed separately. Each COP measure was also separately computed for the ML and AP fluctuations (i.e. \( x \) and \( y \) coordinate) as well as for the overall, RD fluctuations. Subsequent references to the global, seat and foot support COP fluctuations for RD, AP and ML directions will be made as \( COP_S \), \( COP_F \) and \( COP_G \), unless specifically stated. Time and frequency domain parameters that quantified all \( COP_G \), \( COP_S \) and \( COP_F \) fluctuations were calculated and averaged over two trials to represent the value for each subject. Each COP measure was referenced by subtracting them from their mean (Preito et al., 1996). Time-domain measures included: a) the mean distance (MD) which

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represented the average distance traveled by the COP; b) the mean velocity (MV) which was the average velocity of the COP time series; c) the 95% confidence ellipse (AREA-CE) which estimated the elliptical area of best fit enclosed by 95% of COP points; and d) the mean frequency (MFREQ) which was a hybrid measure that represented the rotational frequency of the COP series by the ratio of the mean velocity to the mean distance in revolutions per second (Hz). Stability performance parameters (MD and AREA-CE) were used to evaluate postural stability, whereas the control demand parameter (MV) was used to evaluate the amount of postural activity that the system needed in order to achieve stability (Grangeon et al., 2013). The hybrid measure (MFREQ) described the relationship between the control demand and stability performance (Prieto et al., 1996).

The frequency-domain parameters characterized the area or shape of the power spectral density of the COP series, which was calculated in the frequency range from 0.15 to 5.0 Hz, and assessed postural stability regulation (Grangeon et al., 2013; Prieto et al., 1996). They included: a) centroidal frequency (CFREQ), which represented the central mass frequency; and b) the frequency dispersion (FREQD), which was a unit-less measure of the variability of the power spectral density. It has been reported that the natural frequency of an inverted pendulum is inversely proportional to the moment of inertia (Winter et al., 1998) and that CFREQ is also linked to the inertia of the trunk during sitting balance (Grangeon et al., 2013; Vette et al., 2010). Also, using computational simulations it was found that FREQD was correlated to the stiffness of the inverted pendulum, which was used to model the trunk (Maurer and Peterka, 2005).
4.3.6. Statistical Analysis

Comparisons were performed using Mann-Whitney test for independent samples comparison and Wilcoxon signed-rank test for related samples comparison. Non-parametric tests were chosen over the equivalent parametric tests because the Shapiro-Wilk test suggested that all identified measures were not normally distributed. Statistical analysis was performed with a significance level $P < 0.05$.

4.4. Results

4.4.1. Participants

Ten able-bodied individuals and six individuals with cervical SCI participated in this study. Participants’ demographic information is summarized in Table 4.1. The mean age ($P = 0.428$), mean weight ($P = 0.792$), and mean height ($P = 0.445$) of participants in able-bodied and SCI groups were not significantly different, suggesting that the groups were matched.
Table 4.1: Participants’ demographic information, where AB represents able-bodied individuals and SCI represents SCI individuals.

<table>
<thead>
<tr>
<th>Group</th>
<th>Gender</th>
<th>Age (years)</th>
<th>Height (cm)</th>
<th>Weight (kg)</th>
<th>Level of injury</th>
<th>Motor Impairment</th>
<th>Sensory Impairment</th>
<th>Time since injury (years)</th>
</tr>
</thead>
<tbody>
<tr>
<td>AB</td>
<td>F</td>
<td>27</td>
<td>165.1</td>
<td>54.4</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>27</td>
<td>180.0</td>
<td>76.0</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>25</td>
<td>188.0</td>
<td>88.5</td>
<td></td>
<td></td>
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<td></td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>38</td>
<td>180.0</td>
<td>74.0</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>F</td>
<td>29</td>
<td>162.6</td>
<td>53.1</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>41</td>
<td>182.9</td>
<td>83.0</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>F</td>
<td>38</td>
<td>165.1</td>
<td>61.2</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>F</td>
<td>31</td>
<td>162.6</td>
<td>49.9</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>29</td>
<td>177.8</td>
<td>70.3</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>25</td>
<td>180.0</td>
<td>75.0</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SCI</td>
<td>M</td>
<td>25</td>
<td>180.3</td>
<td>61.2</td>
<td>C5</td>
<td>Incomplete</td>
<td>Incomplete</td>
<td>3</td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>25</td>
<td>170.2</td>
<td>72.0</td>
<td>C5</td>
<td>Complete</td>
<td>Incomplete</td>
<td>3</td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>33</td>
<td>175.3</td>
<td>79.4</td>
<td>C4-6</td>
<td>Complete</td>
<td>Incomplete</td>
<td>16</td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>71</td>
<td>n/a</td>
<td>86.8</td>
<td>C4-6</td>
<td>Incomplete</td>
<td>Incomplete</td>
<td>6</td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>40</td>
<td>n/a</td>
<td>67.8</td>
<td>C6</td>
<td>Complete</td>
<td>Complete</td>
<td>19</td>
</tr>
<tr>
<td></td>
<td>M</td>
<td>54</td>
<td>n/a</td>
<td>83.1</td>
<td>C5</td>
<td>Incomplete</td>
<td>Complete</td>
<td>27</td>
</tr>
<tr>
<td>Mean</td>
<td></td>
<td>31.0</td>
<td>174.4</td>
<td>68.5</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SD</td>
<td></td>
<td>5.9</td>
<td>9.5</td>
<td>13.2</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

4.4.2. Differences in Able-bodied and SCI Groups

Figure 4.2 illustrates a representative sample of 15 s of COP$_G$, COP$_S$, COP$_F$ data during sitting balance for an able-bodied individual and an individual with SCI. The planar representation of AP and ML sway is shown on the left hand side of the figure and the separate AP and ML time series are shown on the right hand side of the figure. The plots illustrate that amount of postural sway. The sway area was considerably larger in an individual with SCI compared to the able-bodied individual.

Comparison of COP$_G$ postural sway parameters between able-bodied and SCI groups, in the first two columns of Table 4.2, revealed that overall RD sway amount (MD and AREA-CE) was considerably larger in the SCI group for all measures. Mean
frequency (MFREQ) of COP<sub>G</sub> was smaller in SCI participants for AP and ML directions as well as the overall RD measure. Analysis of COP<sub>S</sub> postural sway parameters between able-bodied and SCI groups, in the third and fourth column of Table 4.2, revealed that amount of sway (MD and AREA-CE) was significantly larger in the SCI group for ML direction, but not for AP direction, though the means were considerably different. Mean velocity (MV) of COP<sub>S</sub> sway was larger in the SCI group only for the AP direction. Similarly, mean frequency (MFREQ) of SCI participants was smaller for the overall RD sway and for ML direction, but it was not different for AP direction. In the last two columns of Table 4.2, comparison of COP<sub>F</sub> sway parameters did not show any significant differences between able-bodied and SCI groups.

4.4.3. Effect of Foot Support

Effects of foot support forces on the COP were analyzed by comparing parameters obtained for COP<sub>G</sub> to those obtained for COP<sub>S</sub> (i.e. COP<sub>G</sub> vs. COP<sub>S</sub>) in the able-bodied and SCI groups separately. Comparison between the first and third columns of Table 4.2 for the able-bodied group, and second and fourth columns of Table 4.2 for the SCI group, revealed that there were no significant differences in amount of postural sway (MD and AREA-CE) between COP<sub>G</sub> and COP<sub>S</sub> in either able-bodied or SCI groups. Mean velocity (MV) of COP<sub>G</sub> sway, when compared to COP<sub>S</sub>, was larger among the able-bodied and SCI group for the overall RD sway and for the AP direction. The sway frequency (CFREQ and MFREQ) was also larger for COP<sub>G</sub>, when compared to COP<sub>S</sub>, for all measures in able-bodied group and for the overall RD sway and AP direction among individuals with SCI.
**Figure 4.2:** Example of the COP sway for global (COP\(_G\)), seat (COP\(_S\)) and foot support (COP\(_F\)) fluctuations during quiet sitting for: A) one able-bodied individual (AB); and B) one individual with SCI. AP represents anterior-posterior and ML medial-lateral sway direction. The planar representations (left) show spatial fluctuations of the combined AP and ML sway with respect to the origin of the global coordinate frame. Time series plots (right) show the corresponding AP and ML postural sway time series separately for COP\(_G\), COP\(_S\) and COP\(_F\). Note that only a representative 15 s of data is shown to reflect the postural sway behaviour.
Table 4.2: Analysis of parameters for the global (COP<sub>G</sub>), seat (COP<sub>S</sub>) and foot support (COP<sub>F</sub>) postural sway during quiet sitting were performed. AB represents able-bodied group and SCI represents individuals with SCI. Postural stability parameters for the radial distance (RD) measure and the anterior-posterior (AP) and medial-lateral (ML) direction were computed. Shown are the group mean (SD) for the mean distance (MD), mean velocity (MV), 95% confidence ellipse (AREA-CE), mean frequency (MFREQ), centroidal frequency (CFREQ) and frequency dispersion (FREQD). Analysis included the Mann-Whitney test to compare AB and SCI groups for each parameter (light grey) and Wilcoxon signed-rank test to compare COP<sub>G</sub> and COP<sub>S</sub> parameters in each group (dark grey).

<table>
<thead>
<tr>
<th>Measure</th>
<th>Global: COP&lt;sub&gt;G&lt;/sub&gt;</th>
<th>Seat: COP&lt;sub&gt;S&lt;/sub&gt;</th>
<th>Foot Support: COP&lt;sub&gt;F&lt;/sub&gt;</th>
<th>Statistics</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>AB</td>
<td>SCI</td>
<td>AB</td>
<td>SCI</td>
</tr>
<tr>
<td>MD (cm)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>RD</td>
<td>0.13(0.04)</td>
<td>0.40(0.39)</td>
<td>0.13(0.06)</td>
<td>0.48(0.43)</td>
</tr>
<tr>
<td>AP</td>
<td>0.10(0.04)</td>
<td>0.34(0.34)</td>
<td>0.11(0.05)</td>
<td>0.42(0.39)</td>
</tr>
<tr>
<td>ML</td>
<td>0.06(0.03)</td>
<td>0.15(0.14)</td>
<td>0.06(0.02)</td>
<td>0.17(0.16)</td>
</tr>
<tr>
<td>AP</td>
<td>0.46(0.18)</td>
<td>0.48(0.13)</td>
<td>0.31(0.10)</td>
<td>0.41(0.12)</td>
</tr>
<tr>
<td>ML</td>
<td>0.22(0.10)</td>
<td>0.25(0.10)</td>
<td>0.22(0.09)</td>
<td>0.27(0.11)</td>
</tr>
<tr>
<td>MV (cm/s)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>RD</td>
<td>0.35(0.15)</td>
<td>0.36(0.07)</td>
<td>0.17(0.05)</td>
<td>0.26(0.06)</td>
</tr>
<tr>
<td>AP</td>
<td>0.77(0.28)</td>
<td>0.41(0.13)</td>
<td>0.65(0.20)</td>
<td>0.40(0.15)</td>
</tr>
<tr>
<td>ML</td>
<td>1.53(0.35)</td>
<td>1.27(0.30)</td>
<td>1.30(0.30)</td>
<td>1.03(0.25)</td>
</tr>
<tr>
<td>CFREQ (Hz)</td>
<td></td>
<td></td>
<td></td>
<td></td>
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<tr>
<td>RD</td>
<td>1.42(0.33)</td>
<td>1.27(0.36)</td>
<td>1.17(0.29)</td>
<td>1.05(0.34)</td>
</tr>
<tr>
<td>AP</td>
<td>1.40(0.20)</td>
<td>1.10(0.33)</td>
<td>1.27(0.23)</td>
<td>1.05(0.31)</td>
</tr>
<tr>
<td>ML</td>
<td>0.60(0.03)</td>
<td>0.58(0.05)</td>
<td>0.59(0.04)</td>
<td>0.57(0.06)</td>
</tr>
<tr>
<td>FREQD (-)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>RD</td>
<td>0.60(0.04)</td>
<td>0.57(0.04)</td>
<td>0.60(0.04)</td>
<td>0.57(0.04)</td>
</tr>
<tr>
<td>AP</td>
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<td>0.59(0.04)</td>
<td>0.58(0.04)</td>
<td>0.58(0.04)</td>
</tr>
</tbody>
</table>

* P < 0.05; ** P < 0.01
4.5. Discussion

4.5.1. Sitting Balance in Individuals with SCI

Global postural sway (i.e. COP<sub>G</sub>) parameters were used to assess the whole-body sitting balance. Results showed that individuals with SCI had much larger overall postural sway, and their mean frequency of fluctuations was smaller, compared to able-bodied individuals. There were no significant differences in velocity of sway between the two groups for the overall postural fluctuations, i.e. COP<sub>G</sub> (Table 4.2). It has been suggested that the amount of COP sway (i.e. mean distance) is associated with postural stability performance, whereas COP velocity was associated with how much activity the postural control system needed to achieve stability (Grangeon et al., 2013; Maki et al., 1990; Prieto et al., 1996). Using this logic, our results suggest that individuals with cervical SCI had less postural stability (suggested by the sway mean distance) with the same amount of postural activity (suggested by the sway velocity results), for both the anterior-posterior and medial-lateral directions. They indicate that individuals with SCI were less stable despite having similar amount of postural control activity. These findings are consistent with results published by Grangeon et al. (2012), who reported that both the anterior-posterior and medial-lateral stability performance during unsupported sitting was lower among individuals with SCI compared to able-bodied controls.

Inability to voluntarily control postural trunk and lower limb muscles (Minkel, 2000) and reliance on higher, non-postural muscles such as shoulder and neck muscles for balance (Seelen et al., 1998) would most likely explain decreased postural performance observed in individuals with cervical SCI. Grangeon et al. (2012) reported similar findings even though their participant group was not homogenous and included
some individuals with lower injuries (thoracic and high-lumbar injuries, in addition to cervical injuries). Our study examined only individuals with mid-cervical SCI (C4 to C6 level of injury) who generally exhibit higher levels of trunk muscle paralysis and as a consequence have limited ability, if any, to actively regulate sitting balance (Minkel, 2000). As expected, postural stability of SCI participants in our study was lower than that of the participants in Grangeon et al. (2012), when comparing group averages between the two studies (e.g. overall mean distance was 35% larger, and mean velocity 39% smaller among SCI participants in our study compared to Grangeon et al. (2012) study). These results suggest that global COP measures of postural stability can dependably evaluate sitting balance impairment after SCI. Moreover, based on these results, it is evident that individuals with cervical SCI have considerably compromised overall global (i.e. \( \text{COP}_G \)) postural control, which is calculated from the seat and foot support force plates, and indicates the performance of both the trunk control and foot support during sitting balance. In the next two sections the trunk control, which was analyzed using measurements on the seat, and the foot support, which was analyzed using measurements of the foot support, are discussed in more details.

4.5.2. Postural Stability of the Trunk

Postural sway parameters measured on the seat force plate (i.e. \( \text{COP}_S \)) showed that individuals with SCI, compared to able-bodied individuals, had considerably larger postural sway, smaller mean frequency, and larger anterior-posterior sway velocity (Table 4.2). These findings imply that individuals with SCI have lower postural stability of the trunk compared to able-bodied individuals. The observed trunk instability among
individuals with SCI was prevalent in the medial-lateral direction. The observed medial-lateral instability of the trunk is consistent with Shirado et al. (2004) who also showed that individuals with complete thoracic SCI had greater medial-lateral COP postural movements of the trunk, compared to able-bodied individuals. This medial-lateral instability in individuals with SCI may be explained by the passive structures of the trunk and lower limbs which provide a larger base of support for the anterior-posterior direction and the frequently observed kyphotic posture in individuals with SCI which increases anterior-posterior support and may cause medial-lateral instability during sitting (Minkel, 2000). Results also showed that individuals with SCI required more postural activity (i.e. larger mean velocity) to stabilize anterior-posterior fluctuations recorded on the seat (i.e. COPS). However, foot support may have influenced anterior-posterior fluctuations. Overall, our results for the seat fluctuations suggest that trunk control is considerably compromised in individuals with cervical SCI during sitting balance.

4.5.3. Effects of Foot Support on Postural Stability

Comparison of global and seat COP sway parameters (i.e. COPG vs. COPS) was used to examine the effects of foot support on postural stability in able-bodied and SCI groups separately, since foot support postural sway (COPF) did not differ between the able-bodied and SCI groups. Results showed that anterior-posterior postural sway mean velocity with foot support (i.e. COPG), compared to sway without the effect of foot support (i.e. COPS) was larger in both the able-bodied group and SCI group. These findings revealed that foot support increased postural activity of the body by adding faster fluctuations in anterior-posterior direction. Grangeon et al. (2013) investigated the
effects of upper limb support during sitting balance and showed that COP sway velocity increased when hands were placed on the thighs for support, compared to when hands were not rested on the thighs.

Results also showed that centroidal frequency as well as the mean frequency of postural sway with foot support (i.e. COP$_G$), compared to sway without foot support (i.e. COP$_S$), was larger in both able-bodied and SCI groups (COP$_G$ was larger than COP$_S$ for both able-bodied and SCI groups as shown in Table 4.2), which is consistent to what Grangeon et al. (2012) showed during upper limb supported sitting. Larger centroidal frequency of a swaying object is associated with smaller effective moment of inertia of the trunk (Grangeon et al., 2013; Vette et al., 2010). It has also been reported that the natural frequency of the swaying object and its inertia are inversely proportional (Winter et al., 1998). Systems with larger moment of inertial are generally more sluggish and require more time to return to equilibrium, which suggests that they are less stable (Vette et al., 2010). Similarly, it follows that that systems with smaller moment of inertial are less sluggish, require less time to return to equilibrium and are consequently more stable. Our findings suggest that effective moment of inertial during postural stability in both able-bodied and SCI individuals was smaller (i.e. larger centroidal frequency), and imply that both groups had more stable anterior-posterior postural stability due to the effects of foot support. Overall, foot support provided stability predominantly in anterior-posterior direction, which is undoubtedly because the passive mechanics of the foot support are aligned with anterior-posterior direction.

Previous research has shown that able-bodied individuals do not need to actively contract lower limb muscles during quiet sitting as long as they remain within the base of
support (Dean et al., 1999). Similarly, individuals with thoracic SCI do not actively contract their hip and lower limb muscles during reaching tasks (Potten et al., 2002). Since both able-bodied and SCI individuals in our study used foot support in the same way during postural stability, it appears that foot support provided passive stability during quiet sitting in both groups. This is consistent with Gagnon et al. (2008) findings, which showed that individuals with SCI used passive foot support during transfers. Taken together, it seems that trunk control is the predominant mechanism contributing to differences in postural stability during sitting in able-bodied and SCI individuals. Foot support provides passive support and only passively influences how postural stability is attained during sitting (Potten et al., 2002).

4.5.4. Limitations and Future Work

Numerous factors such as the neurological injury level and sensory and motor completeness of injury can affect sitting balance performance after SCI (Grangeon et al., 2012; Seelen et al., 1998). It is possible that age, motor and sensory impairment, and time since injury played a role on the sitting balance of individuals with cervical SCI in our study. Therefore, further studies are warranted to systematically examine their effects on sitting balance after SCI. Moreover, future studies should consider full kinematic analysis of the upper body to understand the multi-segmented trunk control during sitting balance.

4.6. Conclusions

The results of our study indicate that individuals with cervical SCI have significantly compromised sitting postural stability as they swayed at least twice more
than able-bodied individuals. The postural sway of trunk was larger in the SCI individuals than able-bodied individuals, while the use of foot support for postural stability improved the stability but the improvement in stability was not different between the groups. Thus, we conclude that trunk control is the dominant mechanism contributing to differences in sitting postural stability between able-bodied individuals and individuals with SCI, and that the overall instability in individuals with cervical SCI is due to postural instability of the trunk. These results emphasize the importance of trunk control in sitting balance and indicate the importance of recovering trunk function in rehabilitation of individuals with SCI as a way to improve their sitting balance required for functional activities.

4.7. Acknowledgements

This work was supported by the Canadian Institutes of Health Research (CIHR) Grants # MOP-97952 and #IMH-94018, and Natural Sciences and Engineering Research Council Discovery and Accelerator Grants #249669. The authors acknowledge the support of Toronto Rehabilitation Institute – University Health Network who receives funding under the Provincial Rehabilitation Research Program from the Ministry of Health and Long-Term Care in Ontario.
4.8. References


Chapter 5

5. Trunk Muscle Co-activation using Functional Electrical Stimulation Modifies Center of Pressure Fluctuations During Quiet Sitting by Increasing Trunk Stiffness

The material presented in this chapter is currently under review in the article:

NOTE: The content of this chapter is identical to the material presented in the publication except for the text formatting which was done according to University of Toronto requirements.
5.1. Abstract

The purpose of this study was to examine the impact of functional electrical stimulation (FES) induced co-activation of trunk muscles during quiet sitting. We hypothesized that FES applied to the trunk muscles will increase trunk stiffness. The objectives of this study were to: 1) compare the center of pressure (COP) fluctuations during unsupported and FES-assisted quiet sitting - an experimental study and; 2) investigate how FES influences sitting balance - an analytical (simulation) study. The experimental study involved 15 able-bodied individuals who were seated on an instrumented chair. During the experiment, COP of the body projected on the seating surface was calculated to compare sitting stability of participants during unsupported and FES-assisted quiet sitting. The analytical (simulation) study examined dynamics of quiet sitting using an inverted pendulum model, representing the body, and a proportional-derivative (PD) controller, representing the central nervous system control. This model was used to analyze the relationship between increased trunk stiffness and COP fluctuations. In the experimental study, the COP fluctuations showed that: i) the mean velocity, mean frequency and the power frequency were larger during FES-assisted sitting; ii) the frequency dispersion for anterior-posterior fluctuations was smaller during FES-assisted sitting; and iii) the mean distance, range and centroidal frequency did not change during FES-assisted sitting. The analytical (simulation) study showed that increased mechanical stiffness of the trunk had the same effect on COP fluctuations as the FES. The results of
this study suggest that FES applied to the key trunk muscles increases the speed of the COP fluctuations by increasing the trunk stiffness during quiet sitting.

**Keywords:** trunk; posturography; quit sitting; functional electrical stimulation (FES); stiffness; inverted pendulum model.
5.2. Introduction

The human spine is inherently unstable and trunk musculature, which surrounds the spine, is primarily responsible for maintaining its stability against multidirectional external forces (Stokes et al., 2000; Masani et al., 2009; Reeves et al., 2006; Milosevic et al., 2012). During quiet sitting, weak tonic activation of the trunk muscles (1-3% of the maximum voluntary contraction for abdominal muscles, and 4-6% for back muscles) provides sufficient multidirectional trunk stiffness to ensure stable quiet sitting (Masani et al., 2009). Neurological injuries, such as spinal cord injury (SCI) and traumatic brain injury, impact sitting balance. They often result in neuromuscular deficits that cause postural instability (Milosevic et al., 2015 and the inability to effectively compensate for external perturbations (Audu et al., 2015; Kukke and Triolo, 2004; Triolo et al., 2009).

Functional electrical stimulation (FES) can generate muscle contractions by delivering short electric pulses between two electrodes placed on the surface of the skin, over the muscle nerve (Popovic and Keller, 2005). Continuous co-activation of trunk muscles with FES was previously shown to improve clinical measures of static balance by correcting spinal alignment (Triolo et al., 2009) and dynamic balance during forward reaching (Kukke and Triolo, 2004) in people with SCI. It was also shown that stimulation of the trunk muscles with FES can produce sufficient trunk muscle contractions to stabilize up to 45% of body weight during sitting balance perturbations (Audu et al., 2015). Triolo and colleagues (Kukke and Triolo, 2004; Triolo et al., 2009) showed that FES of the trunk muscles can cause postural improvements during sitting balance and they assumed that the improvements were caused by increased trunk stiffness. Previous studies have demonstrated that voluntarily co-activation of trunk muscles increases trunk
stiffness (Lee et al., 2005). Therefore, it is logical that trunk muscle co-activation induced by FES, could also increase trunk stiffness.

To date, no study has investigated how FES applied to the trunk muscles influences postural sway during quiet sitting. Postural control during quiet sitting (Reeves et al., 2006; Milosevic et al., 2015; Vette et al., 2010) and standing (Prieto et al., 1996; Maurer and Peterka, 2005; Masani, Vette and Popovic, 2006), which can be evaluated using the center of pressure (COP) sway fluctuations, has been utilized to characterize the balance control of these two biomechanical systems. To understand the effects of trunk stability on the sitting balance control system, Reeves et al. (2006) investigated the COP fluctuations during quiet sitting on an unstable surface. Their findings suggest that additional voluntary trunk muscle co-activations, which are believed to increase trunk stiffness, increased the COP velocity. Using COP sway measures during quiet sitting, it was also shown that individuals with SCI have compromised control of the trunk, which is caused by their neuromuscular impairment (Milosevic et al., 2015). Further, COP sway measures when combined with computational simulations could be used to quantitatively analyze the underlying postural control mechanisms, including the contribution of stiffness and damping during balance control (Maurer and Peterka, 2005).

We hypothesized that co-activating the trunk muscles with FES will modify sitting balance by increasing trunk stiffness. The objectives of this study were to: 1) compare the COP fluctuations during unsupported quiet sitting to FES-assisted quiet sitting - an experimental study and; 2) investigate how FES influences sitting balance - an analytical (simulation) study.
5.3. Methods

5.3.1. Experimental Study

5.3.1.1. Participants

Fifteen male able-bodied individuals (age 26.7 ± 4.6 years; weight 72.5 ± 8.1 kg; height 175.7 ± 6.7 cm) participated in this study. None of the participants had a history of neurological and sensory impairments, and musculoskeletal injury that could compromise their sitting balance. All participants gave written informed consent in accordance with the principles of the Declaration of Helsinki. The experimental procedures were approved by the local institutional ethics committee.

5.3.1.2. Study Protocol

Participants were asked to maintain an upright sitting posture on a height-adjustable instrumented chair without back support, such that their feet were not supported on the ground and with their arms crossed on their chest (Figure 5.1). Participants maintained quiet sitting posture during: a) unsupported sitting; and b) FES-assisted sitting conditions. For each condition, data was collected over two, 30 sec trials.
Figure 5.1: Experimental setup showing the participant’s posture on an instrumented chair without back support during the sitting balance assessments. The force plate was positioned on the seat surface, under the buttocks, to capture trunk sway, while the participant’s feet were not supported on the ground and the participants had their arms crossed on their chest. The figure also shows the: a) front view of the participant illustrating the approximate location of the FES electrodes on the rectus abdominis (RA) muscle and; b) back view of the participant illustrating the approximate location of the of the FES electrodes on the lumbar erector spinae (L3) muscle. The RA and L3 muscles were stimulated bilaterally and were activated simultaneously to generate co-activations.
5.3.1.3. Functional Electrical Stimulation (FES)

During FES-assisted sitting, a portable FES system Complex Motion (Compex, Switzerland (Popovic and Keller, 2005)) was used to deliver transcutaneous electrical stimulation to the trunk muscles by applying rectangular, biphasic, asymmetric charge balanced stimulation pulses with a 300 µsec pulse duration and 40 Hz frequency via self-adhesive gel electrodes (5x5 cm). Stimulation electrodes were placed on rectus abdominis (RA) and lumbar portions of the erector spinae (L3) muscles bilaterally, and were activated simultaneously to generate co-activations (Figure 5.1). These muscles were chosen because they contribute significantly to trunk stability during sitting balance perturbations (Milosevic et al., 2012). The stimulation intensity for each muscle was determined by gradually increasing the stimulation amplitude with 1 mA increments until the experimenter identified the motor threshold by checking for palpable contractions. The stimulation intensity was then set to twice the motor threshold, or the highest tolerable amplitude, which was higher than the motor threshold but less than twice the motor threshold. Trunk flexors and extensors were symmetrically activated to avoid trunk bending. The average stimulation amplitude was 20.3±3.8 mA for the RA and 24.6±7.4 mA for the L3 muscles.

5.3.1.4. Center of Pressure (COP)

Ground-reaction forces were recorded using a force plate AccuSway^Plus (Advanced Mechanical Technology Inc., USA) positioned on the seat, under the buttocks of the participants (Figure 5.1). Force plate signals were sampled at 500 Hz using a 12-bit data acquisition system NI 6071E (National Instruments, USA). COP fluctuations in
anterior-posterior (AP) and medial-lateral (ML) directions were calculated from the recordings (Prieto et al., 1996). A low-pass filter with a cut-off frequency of 5 Hz was applied to all recordings (Vette et al., 1996).

The time and frequency domain parameters were calculated to characterize the COP fluctuations as described in (Prieto et al., 1996). COP fluctuations were referenced by subtracting the mean position from each time series (Prieto et al., 1996; Maurer and Peterka, 2005). Time-domain measures included: a) mean distance (MD), which represented the average distance from the origin travelled by the COP; b) mean velocity (MV), which was the average velocity of the COP time series; c) range (RANGE), which was the maximum distance between the two points on the COP path; and d) mean frequency (MFREQ), which was a hybrid measure that represented the rotational frequency of the COP series by the ratio of the mean velocity to the mean distance in revolutions per second (Hz). The frequency-domain parameters characterized the area or shape of the power spectral density of the COP series and were calculated in the frequency range from 0.15 to 5.0 Hz (Milosevic et al., 2015; Vette et al., 2010; Prieto et al., 1996). They included: a) centroidal frequency (CFREQ), which represented the central mass frequency; b) frequency dispersion (FREQD), which was a unit-less measure of the variability of the power spectral density; c) 50% power (P50), which included the frequencies below which 50% of the total power of the spectral density was concentrated; and d) 95% power (P95), which was the frequencies below which 95% of the total power of the spectral density was concentrated. Statistical analysis was performed to compare all parameters during unsupported and FES-assisted quiet sitting using the Wilcoxon signed-ranks test. A non-parametric test was chosen because Shapiro-
Wilk test has shown that not all selected measures were normally distributed. Significance level was set at $p < 0.05$.

### 5.3.2. Analytical (Simulation) Study

#### 5.3.2.1. Model

We conducted an analytical (simulation) study to investigate the mechanism of changes of COP fluctuations in AP direction, which were observed in the experiments discussed in Section 2.1. The simulation study was performed using Matlab and Simulink (ver. R2011b, MathWorks, Inc., USA). A feedback model of the control system during quiet sitting was developed using: i) an inverted pendulum model to describe the mechanics of the quiet sitting; ii) a proportional-derivative (PD) controller to represent the neural controller of the central nervous system that regulates balance of the trunk; iii) motor and sensory command transmission delays; iv) a neuromusculoskeletal (NMS) torque-generation process, which was modeled using a second order dynamic equation; and v) mechanical stiffness and passive damping of the trunk. All components of the model are shown in Figure 5.2. A detailed description follows.

The size of the inverted pendulum was calculated using the average subjects’ body mass ($M$) and height ($H$), which was $M = 72.5$ kg and $H = 1.76$ m in this study. The moving mass ($m$), height of the center of mass (COM) ($h$), and the moment of inertia ($I$) of the moving part of the body (i.e., head, arms and trunk) with respect to the greater trochanter were estimated as: $m = 0.678M$; $h = 1.142(0.190H)$; and $I = mh^2$, as described by Winter (Winter, 2009), to obtain $m = 49.2$ kg; $h = 0.381$ m; and $I = 7.14$ kg m$^2$. The neural control was modelled as the PD controller with a proportional gain ($K_P$) and a
derivative gain ($K_D$) because it was shown that a PD controller can represent postural
dynamics during both standing (Maurer and Peterka, 2005; Masani, Vette and Popovic,
2006; Masani et al., 2008) and sitting balance (Audu et al., 2015). A constant time delay
was added to correspond to the motor ($\tau_1$) and sensory ($\tau_2$) transmission time delays of
the central nervous system. Onset latencies of paraspinal muscles were shown to be 21.8
ms (Granata, Slota and Bennett, 2004) and transmission time of approximately 25 ms was
obtained empirically (Goodworth and Peterka, 2009). Also, transmission time between
the motor cortex and the trunk muscles, obtained by motor evoked potentials, was shown
to be between 15.2 and 17.6 ms in able-bodied individuals (Fujiwara et al., 2001). Our
model used the transmission time delay of $\tau_1 = 20$ ms, and the feedback time delay of $\tau_2 =
20$ ms. The NMS torque-generation process, was modelled as a critically damped second-
order system (Masani et al., 2008), where $\omega_n$ was the natural frequency of the second-
order system. $T = 1/\omega_n$ corresponded to the twitch contraction time of the muscle (i.e. 
delay from the muscle contraction to the time force is generated). Thelen et al. (1994)
showed this time to be in the range between 111 and 218 ms for trunk muscles. Masani et
al. (Masani et al., 2008) empirically derived the contraction times for the ankle muscles
in the range between 121 and 192 ms. We chose the twitch contraction times to include $T
= 120, 170$ and $220$ ms, since it is known that NMS process affects the postural control
mechanism (Masani et al., 2008). The output of the neural controller delayed by the
transmission time and the NMS system represented the active torque component.
Mechanical properties of the system, including passive damping ($B$) and mechanical
stiffness ($K$) were implemented separately, modeling the mechanical support structure,
corresponding to the passive torque components. Gaussian random white noise was
inserted into the system at the neural controller level, as shown in Figure 5.2, to drive the simulations.

**Figure 5.2:** Block diagram of the model used in the simulation study. The feedback model included the neural controller with transmission delays ($\tau_1$, transmission time delay and $\tau_2$, feedback time delay) and the neuromusculoskeletal (NMS) torque-generation process, as well as mechanical stiffness ($K$) and passive damping ($B$) to control the inverted pendulum. The inverted pendulum was used to describe the mechanics of the quiet sitting. $m$ is the moving mass, $h$ is the height of center of mass (COM), and $I$ is the moment of inertia of the inverted pendulum. $K_p$ and $K_D$, are proportional and derivative gains of the proportional-derivative (PD) controller, respectively, used to emulate the neural controller. An inverted pendulum model of quiet sitting is represented, where $y_{\text{COP}}$ is the center of pressure (COP) position, $\theta$ is the sway angle, and $g$ is the acceleration of gravity. Gaussian random noise was inserted into the system to drive the simulations.
5.3.2.2. Simulation and Analysis

The tested controller gain combinations included: $0 < K_p < 600 \text{ Nm/rad}; 0 < K_D < 200 \text{ Nm/s/rad};$ and $0 < K < 300 \text{ Nm/rad},$ in increments of 20. The selected controller gain values were based on the experimentally derived gain values for sitting balance (Goodworth and Peterka, 2009) to reflect the physiological system. Damping was implemented as: $1 < B < 6 \text{ Nm/s/rad},$ in increments of 5, since it was shown that it does not considerably affect the system dynamics (Masani et al., 2008). Nyquist stability analysis was performed on the open-loop system to determine the gain combinations that stabilize the closed-loop system.

The gain combinations that stabilized the system were then used to simulate COP fluctuations. Ten, 40 sec trials for each gain combination were simulated to produce the COP fluctuations. Only the last 30 sec of each trial were used for the subsequent analyses. This was done to eliminate the COP fluctuations that are present during the first few seconds of the simulation, i.e. the transient phase. The obtained sway angle of the COM ($\theta$) which corresponds to the trunk angle, was used to calculate the anterior-posterior COP (i.e., $y_{\text{COP}}$). As body sway during quiet sitting is small, the COM and COP can be approximated as: $y_{\text{COM}} \approx h \cdot \sin \theta$ and $y_{\text{COP}} \approx y_{\text{COM}} + \frac{I}{mgh} \ddot{y}_{\text{COM}}$ (Masani et al., 2014). The same COP parameters as in the experimental study were then calculated using the simulated COP fluctuations. The relationship between each COP parameter and controller gain parameters were analyzed using partial correlation analysis with the COP parameters as a dependent variable and controller gains as independent variables. The linear relationship between each COP parameter and the mechanical stiffness gain $K$ was evaluated using partial correlation coefficient. Significance level was set at $p < 0.05.$
5.4. Results

5.4.1 Experimental Study: Effects of FES on Sitting Balance

The representative plots of one participant, shown in Figure 5.3, illustrate that the participant seemed to sway faster and slightly less during FES-assisted sitting, compared to unsupported sitting. The obtained COP parameters are presented in Table 5.1. Comparison of time-domain parameters showed that the mean distance (MD) and the range (RANGE) tend to decrease with FES but the results were not statistically different. The mean velocity (MV) was significantly larger in the FES condition for anterior-posterior direction and the mean frequency (MFREQ) was significantly larger in the FES condition for both anterior-posterior and medio-lateral directions. For frequency domain parameters, frequency dispersion (FREQD) was significantly smaller and the 50% and 95% power frequency (P50 and P95, respectively) were significantly larger for the anterior-posterior direction during FES-assisted sitting.
Figure 5.3: Example of the experimentally obtained center of pressure (COP) fluctuations during: a) Unsupported quiet sitting and b) FES-assisted quiet sitting for one participant. AP represents anterior-posterior and ML medial-lateral sway direction. The planar representations (left) show spatial fluctuations of the combined AP and ML sway. Time series plots (right) show the corresponding AP and ML postural sway time series separately. Note that only a representative 15 sec of data is shown to describe the postural sway behaviour.
Table 5.1: Analysis of the anterior-posterior (AP) and medial-lateral (ML) center of pressure (COP) fluctuation parameters including mean distance (MD), mean velocity (MV), range (RANGE), mean frequency (MFREQ), centroidal frequency (CFREQ), frequency dispersion (FREQD), 50% power (P50) and 95% power (P95) frequency. Results show the mean ± S.D. for each COP fluctuation parameter and compare unsupported and FES-assisted sitting in fifteen (n=15) able-bodied individuals.

<table>
<thead>
<tr>
<th>Measures</th>
<th>Unsupported Sitting</th>
<th>FES-Assisted Sitting</th>
<th>Wilcoxon signed-ranks test</th>
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</thead>
<tbody>
<tr>
<td>MD (mm)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AP</td>
<td>0.61±0.26</td>
<td>0.54±0.24</td>
<td></td>
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<tr>
<td>ML</td>
<td>0.52±0.26</td>
<td>0.46±0.28</td>
<td></td>
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<tr>
<td>MV (mm/s)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AP</td>
<td>2.80±0.41</td>
<td>3.04±0.61</td>
<td>*</td>
</tr>
<tr>
<td>ML</td>
<td>2.02±0.54</td>
<td>2.30±0.77</td>
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<tr>
<td>RANGE (mm)</td>
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<tr>
<td>AP</td>
<td>0.40±0.18</td>
<td>0.37±0.14</td>
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<tr>
<td>ML</td>
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<td>0.33±0.14</td>
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<td>MFREQ (Hz)</td>
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<tr>
<td>AP</td>
<td>0.97±0.30</td>
<td>1.13±0.35</td>
<td>*</td>
</tr>
<tr>
<td>ML</td>
<td>0.82±0.22</td>
<td>1.01±0.27</td>
<td>*</td>
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<tr>
<td>CFREQ (Hz)</td>
<td></td>
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<tr>
<td>AP</td>
<td>1.72±0.23</td>
<td>1.82±0.25</td>
<td></td>
</tr>
<tr>
<td>ML</td>
<td>1.71±0.19</td>
<td>1.67±0.27</td>
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<tr>
<td>FREQD (-)</td>
<td></td>
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<tr>
<td>AP</td>
<td>0.59±0.05</td>
<td>0.56±0.03</td>
<td>**</td>
</tr>
<tr>
<td>ML</td>
<td>0.56±0.05</td>
<td>0.54±0.06</td>
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<tr>
<td>P50 (Hz)</td>
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<tr>
<td>AP</td>
<td>0.50±0.10</td>
<td>0.55±0.13</td>
<td>*</td>
</tr>
<tr>
<td>ML</td>
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<td>0.55±0.14</td>
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<tr>
<td>P95 (Hz)</td>
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<td>AP</td>
<td>0.72±0.18</td>
<td>0.85±0.18</td>
<td>**</td>
</tr>
<tr>
<td>ML</td>
<td>0.81±0.28</td>
<td>0.87±0.31</td>
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* p < 0.05; ** p < 0.01

5.4.2. Simulation Study: Effects of Increased Stiffness on COP Fluctuations

Figure 5.4 shows the stable gain combinations for the simulation study, which were selected via the Nyquist stability analysis. Minimum stiffness of $K = 180 \text{ Nm/rad}$ was sufficient to stabilize the system without the need for a neural controller. It can also be observed that only very high proportional and derivative gains, i.e. $K_P = 200 \text{ Nm/rad}$
and $K_D = 60 \text{ Nm s/} \text{rad}$, were able to stabilize the system without any passive dumping and stiffness contributions (i.e., $B = 0 \text{ Nm s/} \text{rad}$ and $K = 0 \text{ Nm/} \text{rad}$).

Table 5.2 summarizes the results of the partial correlations analysis. $K$ was positively correlated with mean velocity, mean frequency and 50% and 95% power (MV, MFREQ, P50 and P95) and negatively correlated with frequency dispersion (FREQD). There was no correlation between $K$ and the mean distance (MD), range (RANGE) and centroidal frequency (CFREQ).

**Figure 5.4:** The gain combinations that stabilized the model used in the simulation study, where $K_P$ is the proportional gain, $K_D$ is the derivative gain of the proportional-derivative (PD) controller used to emulate the neural control, and $K$ is the mechanical stiffness contribution. The figure shows the relationship between the parameters.
Table 5.2: Partial correlation analysis for anterior-posterior (AP) center of pressure (COP) fluctuation parameters ($y_{COP}$) and the mechanical stiffness controller gain. Shown are the coefficients of correlation between each COP measurement and the mechanical stiffness gain which was varied as $0 < K < 300$ Nm/rad, while controlling for the effect of proportional gain ($K_P$) and derivative gain ($K_D$). Included are the mean distance (MD), mean velocity (MV), range (RANGE), mean frequency (MFREQ), centroidal frequency (CFREQ), frequency dispersion (FREQD), 50% power (P50) and 95% power (P95) frequency, which were obtained in the simulation study.

<table>
<thead>
<tr>
<th>Measures</th>
<th>$K$</th>
</tr>
</thead>
<tbody>
<tr>
<td>MD</td>
<td>-0.035</td>
</tr>
<tr>
<td>MV</td>
<td>0.283**</td>
</tr>
<tr>
<td>RANGE</td>
<td>-0.096</td>
</tr>
<tr>
<td>MFREQ</td>
<td>0.927**</td>
</tr>
<tr>
<td>CFREQ</td>
<td>-0.034</td>
</tr>
<tr>
<td>FREQD</td>
<td>-0.543**</td>
</tr>
<tr>
<td>P50</td>
<td>0.422**</td>
</tr>
<tr>
<td>P95</td>
<td>0.422**</td>
</tr>
</tbody>
</table>

* $p < 0.05$; ** $p < 0.01$

5.5. Discussions

5.5.1. FES modifies Sitting Balance

Experimental results indicate that FES applied to the trunk muscles during quiet sitting makes the COP move faster (i.e. larger MV, P50 and P95), while it does not affect the amount of COP fluctuations (i.e. MD and RANGE). Results also showed that frequency dispersion (i.e. FREQD) was smaller during FES-assisted sitting, indicating
that the frequency components of COP fluctuations became narrower with the application of FES (Table 5.1).

It has been previously shown that the COP velocity is larger in elderly people than in young people and that the amount of COP fluctuations is not different between the two populations during standing (Prieto et al., 1996). Prieto et al. (Prieto et al., 1996) attributed these differences to stiffness in the lower limbs, which were observed in elderly people. Reeves et al. (Reeves et al., 2006) reported that voluntary trunk muscle co-contraction during sitting, which was assumed to increase trunk stiffness, increased COP velocity. Using simulations, Maurer and Peterka (Maurer and Peterka, 2005) found correlations between frequency dispersion of the swaying object and the mechanical stiffness of that object. Consequently, our experimental results suggest that changes in COP fluctuation during FES-assisted sitting are due to increased trunk stiffness resulting from FES being applied to the trunk musculature. Stiffness is not necessarily a good balance control strategy for quiet sitting. However, increasing trunk stiffness prior to or during external perturbations can improve response to such perturbations (Reeves et al., 2006; Audu et al., 2015). Contracting trunk muscles after neurological injuries such as SCI could be useful for improving trunk control (Milosevic et al., 2015) and it could improve the functional sitting balance of people with SCI (Kukke and Triolo, 2004; Triolo et al., 2009).

Our results also indicate that COP fluctuations during FES-assisted sitting primarily affected the anterior-posterior direction. This is likely because the muscles that we stimulated were trunk flexor and extensor muscles (i.e. rectus abdominis and lumbar erector spinae), which mainly control anterior-posterior stability (Milosevic et al., 2012).
These results suggest that if FES is applied to the trunk muscles, it could provide direction specific stiffness of the trunk, depending on which muscles are activated. The ability to selectively activate specific muscles may be a unique advantage of FES for control of trunk muscles during sitting balance.

5.5.2. FES increases Trunk Stiffness

The analytical (simulation) study showed that increased mechanical stiffness $K$ had the same effect on COP fluctuations as the application of FES on the trunk muscles. That is, in our simulation study, $K$ was positively correlated with MV, MFREQ, P50 and P95, negatively correlated with FREQD, and not correlated at all with MD, RANGE and CFREQ. In the experimental study with FES-assisted sitting, FES applied to trunk muscles increased MV, MFREQ, P50 and P95, decreased FREQD and it did not change MD, RANGE and CFREQ. These results support the hypothesis that FES applied to the trunk muscles increased stiffness during sitting. This is similar to the findings of Lee et al. (Lee et al., 2006) who found that voluntary co-contraction of trunk muscles increases trunk stiffness. Maurer and Peterka (Maurer and Peterka, 2005) also examined the relationship between sway measures and model parameters during standing and showed that increased stiffness was generally associated with increased sway velocity, which also agrees with our experimental and simulation results.

The simulation study results showed that mechanical stiffness and proportional gain parameters (i.e. $K$ and $K_p$, respectively) stabilized the trunk and had an inverse relationship (Figure 5.4). This is because the summation of $K$ and $K_p$ are approximately equal to the required stiffness of $mgh = 184$ Nm/rad, although it is possible for this value
to be larger. Very large proportional gains can induce unstable posture but these gain combinations would be eliminated by the Nyquist stability criterion. Similar relationships between mechanical stiffness and neural controller gains were obtained by Masani et al. (Masani et al., 2008) in standing balance simulations. This relationship suggests that, in order to compensate for the reduced neural contributions (i.e. lower $K_P$) in people with sitting postural instability due to neurological impairments, it is necessary to increase the mechanical stiffness (i.e. $K$). One possible way of achieving this is by applying FES to the muscles of interest (i.e. those that can increase stiffness along particular trunk axes).

5.5.3. Limitations

Considering that it is difficult to manipulate only one element of postural control in the experiments, it is possible that there are other factors which could have also affected the results in addition to the increased trunk stiffness. Since FES is a noticeable stimulus, it may be that other muscles were voluntarily recruited in addition to those that were stimulated using FES, which may have also contributed to the above findings. Increasing trunk stiffness may not be a desirable balance control strategy during quiet sitting as it could lead to muscle fatigue (Kukke and Triolo, 2004; Triolo et al., 2009). However, increasing trunk stiffness can improve sitting in people with SCI (Kukke and Triolo, 2004; Triolo et al., 2009) and during external perturbations aimed at disrupting balance (Audu et al., 2015).
5.6. Conclusions

Our experimental study results showed that FES applied to the trunk muscles modified the COP fluctuation during quiet sitting. Simulations that were performed as part of this study suggested that FES of the trunk muscles increased trunk stiffness. Previous models of sitting balance with FES provided some evidence in support of the idea that activation of trunk muscles using FES increased trunk stiffness. Our experimental and simulation results provided additional indication that co-activation of trunk muscles using FES indeed increased trunk stiffness during quiet sitting. Since FES can activate muscles in individuals with upper motor neuron deficit, such as people with SCI, it may be a viable strategy to apply FES on the trunk muscles to improve their sitting balance. As such, an FES intervention would be used to increased trunk stiffness and improve balance during quiet sitting. Since this was a preliminary study, further experiments are required to fully confirm the effectiveness of FES to improve quiet sitting balance in individuals with SCI.

5.7. Acknowledgements

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5.8. References


Chapter 6

6. Conclusions

6.1. Scientific Contributions

The results presented in this dissertation have contributed to the knowledge and understanding of the trunk neuromuscular system and development of the neuroprosthesis for sitting balance that uses functional electrical stimulation (FES) technology that can be used to activate trunk muscles during sitting balance. The knowledge obtained in the presented studies has advanced our understanding of: (1) trunk muscles synergies; (2) how anticipation of perturbation affects trunk muscle reflexes; (3) influence of spinal cord injury (SCI) on sitting balance; and (4) how the neuroprosthesis can influence the recovery of trunk function during sitting balance. Each of the presented studies in this thesis is related to the proposed thesis objectives (Section 1.2). The specific contributions of each study are discussed next.

6.1.1. Models of Trunk Control

In the first study, I developed models for trunk muscle synergy activations that can predict which muscles activate to stabilize multi-directional disturbances during sitting balance perturbations (Chapter 2). This study is related to objective (1) of the thesis which was to analyze trunk muscle synergies of five muscles recorded bilaterally
during eight different directional perturbations and to compare the direction-dependent muscle synergies. Analysis showed significant direction-dependent differences in the postural synergies during different perturbations. Moreover, quantitative Gaussian fit models were produced for each muscle. These models have not only revealed insights into mechanisms of trunk muscles control during sitting perturbations, but also allow us to predict which muscles need to be activated to respond to different perturbation directions. The results supported the hypothesis that trunk muscle synergies have unique, direction-dependent responses which can be modeled using descriptive models. Such analysis offers an easy reverse engineering method for exploration and comparison of complex neuromuscular systems, which can describe postural synergies at a glance. The results of this study help us understand the complex control of trunk muscles and its influence on sitting balance. This study has been published in the IEEE Transactions of Biomedical Engineering journal.

6.1.2. Control of Trunk Muscle Reflexes

I also investigated the effect of anticipation of the perturbation on trunk neuromuscular responses during sitting balance perturbations (Chapter 3). This study is related to objective (2) of the thesis which was to identify the responses of the trunk muscles to sudden perturbations and to analyze the effects of anticipation of perturbation on the trunk muscle responses. The results showed that, when the perturbation was delivered by moving the surface on which participants were sitting, trunk muscle onset latencies were approximately 100 ms following the onset of the perturbation, while the trunk started moving approximately 40 ms after the perturbation. Therefore, it appears
that trunk muscles have medium latency responses based on simple polysynaptic reflexes to control trunk stability during sudden perturbations. When the subject anticipated the perturbations, the onset latencies were reduced by approximately 15 ms, which is a considerable improvement and suggests that the central nervous system modulates readiness of the trunk muscles based on anticipatory information to control trunk stability. The amplitudes of trunk muscle responses were unaffected when perturbations could be anticipated. These results supported the hypothesis that trunk muscles respond using reflexive and non-voluntary mechanisms. However, the amplitudes of trunk muscle responses remained unaffected when perturbations could be anticipated which contradicts our hypothesis that amplitudes of the trunk muscles responses will be larger compared to unanticipated perturbations. Regardless, when perturbations can be anticipated, trunk muscle responses had a considerable improvement as suggested by faster response onset latencies. This is the first study to show that trunk muscle reflex latencies can be modulated when the support surface perturbation can be anticipated. It tells us how the central nervous system controls the complex mechanism of the trunk muscles during sitting balance. The study has been submitted for publishing in the Journal of Electromyography and Kinesiology.

6.1.3. Sitting Balance Control after SCI

I demonstrated that individuals with SCI have reduced trunk control which is a major reason for their poor sitting balance (Chapter 4). It is known that individuals with cervical SCI usually sustain impairment of the trunk and upper and lower limbs, resulting in compromised sitting. This study is related to objective (3) of the thesis which was to
compare the sitting balance of individuals with cervical SCI and able-bodied individuals and investigate the effects of foot support and trunk control on their sitting balance. As expected, individuals with SCI had significantly compromised sitting compared to able-bodied individuals. The results also revealed that trunk control is the dominant mechanism contributing to differences in sitting postural stability between able-bodied individuals and individuals with SCI, and that the overall instability in individuals with cervical SCI is due to postural instability of the trunk. These results supported the hypothesis that individuals with SCI had poorer sitting balance and that trunk control is responsible for poorer sitting among individuals with cervical SCI. This emphasizes the importance of trunk control in sitting balance and indicates the significance of recovering trunk function in rehabilitation of individuals with SCI as a way to improve their sitting balance. This study has been published in the Clinical Biomechanics journal.

6.1.4. FES for Sitting Balance Control

In the last part of my thesis, I demonstrated that co-contractions of trunk muscles with FES can increase trunk stiffness (Chapter 5). Based on the results from Chapters 2, 3 and 4, which emphasize the importance of the trunk function during sitting and the role of the trunk muscles responses and reflexes during sitting balance, FES was used to artificially activate the trunk muscles. This study is related to objective (4) of the thesis which was to examine the impact of FES induced co-activation of trunk muscles on sitting balance. The results supported the hypothesis that co-activation of trunk muscles using FES increased trunk stiffness during sitting balance. However, the amount postural sway was not reduced as hypothesized initially. It can be argued that the amount of
postural sway was not reduced since the participants were all able-bodied individuals and they already have optimal postural control. Since some postural sway is inherent in sitting and standing balance control, FES cannot further reduce this sway. Also, increased stiffness of the trunk can be advantageous in responding to external balance disturbances. These are the necessary steps in developing an effective neuroprosthesis for sitting balance, and the results established the feasibility for using FES to increase trunk stiffness. This study was submitted for publishing in the Journal of Neuroengineering and Rehabilitation.

6.1.5. Contributions to Neuroprosthesis Development

Overall, my thesis has contributed to better understanding of the mechanisms responsible for sitting balance and development of a neuroprosthesis for sitting balance. Individuals with SCI have difficulties maintaining sitting balance which can affect their ability to perform activities of daily living and results in secondary health complications. I have shown that although foot support provides passive stability during sitting balance, trunk control is the dominant mechanics which is responsible for sitting instability among individuals with SCI. Therefore, in rehabilitation it is important to develop programs for improving trunk control as a way to improve sitting balance in individuals with SCI. To better understand the trunk function during sitting balance one must first understand the neuromuscular and biomechanical mechanisms used during functional tasks. My study has contributed to the better understanding about direction-dependent trunk muscle synergies which are used to control sitting balance. Moreover, I developed a set of equations for each trunk muscle which can be programmed into an FES stimulator and
utilized to activate trunk muscles during sitting balance disturbances. I also showed that it takes the trunk muscles approximately 100 ms to activate after the perturbation. Therefore, to achieve effective trunk stability, it is necessary to activate trunk muscles using FES at the most 100 ms after the perturbation onset. Finally, I showed that using a sitting neuroprosthesis to activate trunk muscles can increase trunk stiffness. Using FES to augment trunk stiffness can be beneficial for enabling an SCI patient to rapidly respond to sitting balance disturbances. These findings have direct implications on development of a sitting neuroprosthesis that can be used to improve trunk stability and sitting balance after SCI.

6.2. Social Contributions

The Rick Hanson Institute released the first Canadian national atlas of SCI rehabilitation, which outlines the current state of practice as well as the challenges for SCI rehabilitation in Canada and suggestions on how to improve the practice (Craven et al., 2012). It considers disability as a limitation to the individual that can be improved or restored through rehabilitation. Trunk stability and sitting balance have been identified in the core sets of SCI rehabilitation in both post-acute (Kirchberger et al., 2010) and long-term (Cieza et al., 2010) SCI care. My research is focused on understanding the neuromuscular control of the trunk muscles during sitting balance as a way to improve trunk stability and sitting balance control among people with neurological disorders such as SCI. Improving trunk stability and sitting balance would have an important role in the rehabilitation of individuals with SCI. My work has also established an important and
necessary step in developing a sitting balance neuroprosthesis, which aims to improve activity performance during sitting balance. Such a neuroprosthesis would allow an individual with a SCI who does not have voluntary control of trunk muscles to perform activities that require trunk stability and sitting balance to participate in tasks that were otherwise difficult or impossible to do independently. Overall, better trunk stability would help minimize the risk of falling during external balance disturbances and could increase sitting stability during the performance of daily living activities by increasing the functional workspace. Undoubtedly, my research has contributed to the better understanding of trunk function and development of a sitting neuroprosthesis, which can lead to improved quality of life for those living with SCI in the global and Canadian context.

6.3. Future Directions

The next step in my research would be the development of the neuroprosthesis that would be integrated into a wheelchair to activate the appropriate trunk muscles and increase trunk stiffness. This neuroprosthesis should allow subjects to maintain sitting balance despite mild external or internal perturbations (e.g., when a patient in a wheelchair suddenly goes over a bump on the road or voluntarily reaches for an object). The objectives of this future work are to develop a neuroprosthesis for sitting balance and examine the neurophysiological and neuroplastic changes in the central nervous system that occur as a result of using the neuroprosthesis. The next generation neuroprosthesis should implement a closed-loop controller to adjust muscle activation levels using FES
and increase trunk stiffness based on the position of the trunk. Using such a neuroprosthesis is expected to have a neurophysiological effect resulting from stimulation of the trunk muscles. This would become a novel technology that would have neurological recovery benefits in addition to serving as an assistive device for individuals with SCI.
6.4. References

