AN EVALUATION OF BALANCE PERTURBATION PARADIGMS AND THE EFFECT OF AGE ON THE PERTURBATION CHARACTERISTICS RELATIONSHIP

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ABSTRACT

The ability to recover balance to avoid falling following a postural perturbation is a critical aspect of mobility. Our knowledge of human balance recovery emanates from studies that have intentionally disrupted balance control. These studies have utilized a number of different postural perturbation methods and a variety of perturbation intensities, which makes it difficult to compare between studies, and which may have resulted in inconsistencies and disagreement in the scientific literature. Previous work has suggested that different postural perturbations may induce fundamentally different responses, though the literature is sparse. It is important to understand these differences before a selection of methods could be made to investigate potential balance deficits among different special populations (e.g. older adults; neurological patients). The purpose of the current dissertation was: 1) to directly compare two different types of perturbation, platform-translation and shoulder-pull, on balance-correcting responses, 2) to describe kinematic and neuromuscular responses to different perturbation methods, to explore similarities and differences in the nature of the responses associated with different perturbation methods, and 3) to examine the effect of age on the relationships between perturbation characteristics in determining the balance-correcting response. Four studies were conducted. Firstly, the perturbation characteristics, force and displacement, common to platform-translation and shoulder-pull methods were determined to allow for a reasonable comparison between balance-correcting responses induced by both perturbation methods. These characteristics were used in the second and third study, which: 1) explored differences in dynamic postural stability resulting from two different perturbation methods, and 2) explored similarities and differences in the organization of balance-correcting responses induced with both methods. The fourth study investigated the effect of age on the relationship between perturbation characteristics, which determine the type of balance correcting response (i.e. feet-in-place or stepping) using a shoulder-pull method. These studies suggest that while there are similarities in the balance-correcting responses between perturbation methods, there are also critical differences that are unique to the modes of perturbation utilized. The current dissertation underscores that caution is required when interpreting results of studies utilizing different perturbation methods and that individual differences between participants, which can mask age-related differences, need to be recognized.

DEDICATION

This dissertation is dedicated to my parents, Vitaly and Svetlana, and my girlfriend, Cora Sin. I can't imagine how much tougher this was on you than it was on me. Thank you for your endless support, patience, and understanding. This accomplishment is just as much yours as it is mine.

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LIST OF ABBREVIATIONS

Ant	Anterior
BF	Biceps femoris
BOS	Base of Support
BW	Body Weight
BWD	Backward
CNS	Central Nervous System
COM	Center of Mass
COP	Center of Pressure
EC	Eyes-Closed
EMG	Electromyography
EO	Eyes-Open
ES	Erector spinae
FWD	Forward
GM	Gastrocnemius medialis
HAT	Head, Arms, Trunk
MOS	Margin of Stability
NSS	Stance Side
OA	Older Adults
PLAT	Platform-translation
Post	Posterior
PULL	Shoulder-pull
RA	Rectus abdominis
RF	Rectus femoris
SS	Stepping Side
TA	Tibialis anterior
YA	Younger Adults

CHAPTER 1

General Introduction

1.1. Literature Review

The ability to recover balance following a postural perturbation is a critical aspect of mobility. Postural control can be compromised in special populations, such as older adults, resulting in falls. It has been shown that approximately 30% of adults over the age of 65 and 40% over the age of 75 fall each year. The percentage of fallers increases to 50% for older adults living in long-term care, and to 75% for those who have fallen in the past (Rubenstein, 2006). Falls have been identified as a major cause of pain, disability, psychological grief, reduced quality of life, and early mortality (Murray and Lopez, 1997). According to Statistics Canada (2011), the baby boomers, the individuals who were born between 1946 and 1965, and their parents, the individuals who were born between 1919 and 1940, are the largest populations in Canada, which comprise 29% and 9% of the total Canadian population, respectively. The World War II generation comprises 4% of the total population, and includes individuals who were born between 1941 and 1945. Collectively, these generations make up nearly half of the total Canadian population (Statistics Canada, 2011). The rate at which the Canadian population is aging is concerning because approximately half of the baby boomers are already in their sixties, when the risk of falls and bone fractures due to falls increases substantially (Morrison et al., 2013). Recent findings suggest that the estimated direct cost of falls to the Canadian health care system is approximately $2 billion annually (Accreditation Canada, 2014).

The incidence of falls and the associated costs to the Canadian health care system is likely to increase at an alarming rate in the near future. It is clear that research into causes and prevention of falls, as well as research into the function of the postural control system, deserves careful consideration in populations at risk, such as older adults, in addition to younger adults.

It has been suggested that the postural control system serves two main functions: 1) to build up posture against gravity and 2) to ensure that balance is maintained (Massion, 1994). Since the human body consists of multiple segments and has only two points of contact with the support surface during unassisted bipedal standing, it is inherently unstable, and the task of preserving balance is complex. The behavior of the human body during quiet standing has been modeled as an inverted pendulum or a set of inverted pendula that are linked to one another (Bortolami et al., 2003; Gage et al., 2004). Adequate neuromuscular control of individual segments of the body is vital to maintenance of postural balance, such that even a seemingly simple postural task (e.g. quiet standing) necessitates a complex coordination of joint moments required for the body to preserve upright posture (Hsu et al., 2007). However, ultimately the preservation of postural balance is governed by the relationship between centre of mass (COM), centre of pressure (COP), and the base of support (BOS).

During a quasi-static postural control task, such as quiet standing, the postural control system acts to preserve balance by maintaining vertical projection of the body's COM within the limits of the body's BOS. The COM is defined as the net location of the weighted average of the entire body mass or a system of bodies, i.e. the point where the mass of the entire body is concentrated in three-dimensional space and is defined by the vertical and two horizontal axes (Winter, 2005). The BOS is defined as the area bound by the outermost regions of contact between the body and the support surface or support surfaces (Winter, 2005). For instance, the BOS of an individual during quiet non-assisted bipedal standing is defined by the outermost boundary of their feet including the space between the feet. The horizontal movement of the body's COM within the boundaries of BOS is controlled by the COP, which is defined as the location of the net ground reaction force (Winter, 2005). Thus, the freedom of movement of COP is limited by the size of the BOS. During quiet standing, as vertical projection of COM moves toward the boundaries of BOS, the postural control system responds by creating joint moments to move the COP in front of the translating COM and to move the COM away from the boundaries and toward the centre of BOS (Winter et al., 1990; Winter et al., 1996). It has been suggested that sagittal ankle moments produce anteroposterior movement of COP, while frontal hip moments affect COP movement mediolaterally (Winter et al., 1996). Traditionally, postural control research focused on position limits which were often expressed as peak COM excursions with respect to the peak COP or the boundaries of BOS (Horak et al., 2005; Winter et al., 1996).

The traditional thought, that postural control is dependent on positional limits and that as long as COM is within the BOS the postural balance is maintained, has been challenged. It has been suggested that not only the position of COM but also the velocity of COM plays an important role in maintaining postural balance (Pai and Patton, 1997).

Though Pai and Patton (1997) have agreed that during quiet standing the velocity of COM is quite small and that measures based on position limits could be used to quantify postural control. However, during dynamic situations such as gait or balance recovery following a postural perturbation, the COM velocity can have a robust effect on postural control that may not be reflected effectively in measures based on position limits (Pai and Patton, 1997). Hof et al. (2005) have built on the concept suggested by Pai and Patton (1997) and proposed a measure of postural stability based on the inverted pendulum model. The Margin of Stability (MOS) is a measure that could be used to describe the relationship between COM and BOS during a dynamic situation (Hof et al., 2005). The MOS is defined as the distance between the extrapolated COM position and the boundary of BOS. The calculation of extrapolated COM position is based on the horizontal position of COM, horizontal velocity of COM, and the natural frequency (eigenfrequency) of a simple pendulum (Hof et al., 2005). It is accepted that a larger MOS, as opposed to smaller or negative MOS, during balance recovery is indicative of superior postural stability, since the extrapolated COM is farther from the boundaries of BOS (Arampatzis et al., 2008; Carty et al., 2011; Karamanidis and Arampatzis, 2007). Therefore, unlike measures based on position limits, MOS is better suited for investigations of postural responses during dynamic situations such as those where postural balance is perturb.

Though quiet standing in human participants has been researched extensively, the research is limited in that the quiet standing task innately may not be adequately challenging to expose differences in postural control between populations of interest. For instance, it has been shown that during quiet standing, older adults are able to demonstrate performance similar to that of younger adults (Fujiwara et al., 2007). Moreover, patients with neuropathies and vestibular loss have also demonstrated performance similar to that of healthy controls during a quiet standing task (Horak et al., 1990; Nardone et al., 2000). Therefore, postural control in young and healthy participants as well as special populations is often investigated during dynamic situations, such as those where postural balance is threatened by external or self-induced postural perturbations. In a research setting, the investigation of postural control during a dynamic situation is often achieved by exposing a quietly standing participant to an externally-facilitated postural perturbation. During dynamic situations, the postural control system is challenged to a greater extent than during quiet standing and, therefore, dynamics situations can be more revealing of the mechanisms of postural control. For instance, using postural perturbation paradigms, researchers have shown that older adults are more likely than younger adults to take a step when their postural balance is perturbed and at a lower threshold of instability (Jensen et al., 2001; Mille et al., 2003); and that older adults are more likely to require multiple steps to recover balance (Luchies et al., 1994; McIlroy and Maki, 1996). Moreover, older adults have also been shown to respond slower to a postural perturbation than younger adults (Maki et al., 2001). Collectively, these individual findings create knowledge around dynamic postural control in older adults. However, it is important to note that different perturbation methods were used to perturb participants in these studies. The use of different perturbation methods may have previously caused controversy in postural control literature between two schools of thought: those who argued for centrally activated balance-correcting responses (Bloem et al., 2000; Bloem et al., 2002) and those who argued for distally activated balance-correcting responses (Horak and Nashner, 1986; Nashner, 1982). Recently, it has been suggested that the findings of perturbation studies may be method-specific (Mansfield and Maki, 2009), which alerts that caution is required when interpreting data obtained using different perturbation methods.

There are two general types of perturbation methods that are used to probe dynamic postural control where a perturbation stimulus is applied to: 1) the support surface (floor) that a participant stands on and 2) to the participant's body directly. The support-surface perturbation method is effected by: 1) moving the floor (e.g. Jensen et al., 2001; Maki et al., 2001; McIlroy and Maki, 1996), which is known as support-surface translation (platform-translation), and 2) by tilting the floor (e.g. Bloem et al., 2000; Gage et al., 2008), which is known as support-surface rotation (platform-rotation). The platform-translation method produces a sensation of the surface slipping from under one's feet, such as during a slip on ice, while the platform-rotation method produces a sensation of rotating surface similar to that which can be experienced when paddle boarding or surfing. Unlike the support-surface perturbation, the upper body perturbation methods are effected by: 1) destabilizing a quietly standing participant by pushing or pulling them (e.g. Hsiao-Wecksler et al., 2003; Luchies et al., 1994; Mille et al., 2003), which is known as the push/pull (also cable-pull) method, 2) releasing a participant from a leaning posture (e.g. Arampatzis et al., 2008; Verniba and Gage, 2014), which is known as the tether-release (also lean-and-release) method, and 3) releasing a participant who is actively pushing against a stable object (e.g. Bortolami et al., 2010; Krebs et al., 2001), which is known as the hold-and-release method. When the push/pull method is used, a participant is pushed by a rod or pulled via a cable attached to their upper body or waist via a harness. The tether-release method involves a participant leaning away from the release mechanism while being supported by a tether; when the tether is released, the participant is allowed to fall. The hold-and-release method involves a participant actively applying force with their shoulders or chest against a stable object, such as the extended arms of a clinician or researcher, which when suddenly removed causes the participant to lose balance control. The aforementioned perturbation methods are widely used in postural control research; however, each has characteristics which can be considered advantageous or disadvantageous.

The perturbation onset, direction, and ecological validity are the features of a perturbation method that should be considered when designing a study. The anticipation of perturbation onset by a participant can be attenuated by randomly varying the timing of perturbation. The predictability of perturbation direction can be reduced when using the cable-pull or support-surface perturbation methods, as study protocols often combine multiple-degrees-of-freedom motion with catch trials (e.g. Zettel et al., 2008), which makes it difficult for a participant to predict and anticipate the direction of perturbation. On the contrary, the direction of perturbation can be obvious to a participant when using tether-release or hold-and-release methods. While platform-rotation perturbation can be less predictable in terms of direction of perturbation, it is less likely to occur in the natural environment than platform-translation perturbation; therefore, the use of a rotating floor to perturb a participant may not be as ecologically valid as compared to the use of a translating floor. Likewise, tether-release and hold-and-release perturbation methods are less likely to occur in the natural environment when compared to a push/pull type of perturbation. Platform-translation and push/pull perturbations are likely less temporally and spatially predictable, yet more ecologically valid compared to the other perturbation methods, which may be one of the reasons for the popularity of these perturbation methods among researchers. While there have been substantial research contributions made with the use of platform-translation and cable-pull methods, the comparison between balance-correcting responses induced with these perturbation methods has yet to be made, as recent evidence suggests that platform-translation perturbations may be more destabilizing than cable-pull methods (Mansfield and Maki, 2009).

There are two general types of balance-correcting responses observed in humans: fixed-support and change-in-support. During a fixed-support response, depending on the magnitude of perturbation and the type of support surface (e.g. cluttered or slippery surface), ankle strategy, where primary movement occurs at the ankle joint; hip strategy, where primary movement occurs at the hip joint; or a combination of ankle and hip strategies are observed (Horak and Nashner, 1986). The defining characteristic of a fixed-support response is the lack of change in foot position during the balance-correcting response (Runge et al., 1999), which is why in literature a “fixed-support” response is often used synonymously with “feet-in-place” response. In contrast, change-in-support response is manifested by the change in the size of BOS. The change-in-support responses are often achieved by taking a step in the direction of COM movement and, therefore, are referred to as stepping balance-correcting responses. Alternatively, a change-in-support response can also be achieved by reaching and grasping a stationary object for support. A combination of both stepping and stepping with reaching and grasping has also been described (Maki and McIlroy, 1997). During the stepping response, as opposed to a feet-in-place response, the size of BOS is increased. Thus, larger horizontal displacements of COM, which may result from a larger perturbation stimulus, can be accommodated without the increase in the risk of COM displacing outside the boundaries of BOS. Moreover, the increase in the size of BOS during a stepping response may also allow for greater temporal margin, thereby allowing an individual to generate adequate joint moments to reduce outward velocity of COM. Older adults have been shown to demonstrate lower peak joint moments and lower rate of joint moment generation during balance recovery than younger adults (Pijnappels et al., 2005), which suggests that older adults may not be able to place their foot adequately fast and far ahead of moving COM in order to reduce its velocity and prevent it from leaving the boundaries of BOS. The loss of strength, which is among one of the most notable changes with age due to age-related senile sarcopenia (Narici et al., 2005), and quickness with which counteracting joint moments are generated is, therefore, likely among the reasons for reliance on stepping responses and the use of multiple steps during balance recovery. Ultimately, the type of balance-correcting response, feet-in-place or stepping, depends on the characteristics of the perturbation stimulus.

A perturbation stimulus is often characterized by the magnitude of applied acceleration or force, as well as distance or time over which the perturbation stimulus is applied. Although the methodology differs between studies which utilize platform-translation and cable-pull methods in that the stimuli are applied to different areas of participant's body (i.e. feet during support surface perturbation and upper body or waist during cable-pull perturbation), the mechanical principles of perturbation are similar. Whether the force is the characteristic that is varied to result in the body to undergo acceleration or whether the amount of acceleration is varied directly (often the case when electric or hydraulic actuators are utilized), which implies imbalance of forces acting on the body, the desired effect is movement that perturbs the quasi-static state of the participant's body. Another characteristic of a perturbation is distance or time over which the force or acceleration is applied. Mechanical work is done on an object if an object is moved a distance in the direction of applied force; thus, mechanical work is the product of applied force and distance that an object has moved. Since mechanical work is the product of applied force and distance, in the framework of human postural control research, the amount of force or distance can be varied to do, conceivably, equal amount of mechanical work on a participant. Further, the relationship between the two perturbation characteristics, applied force and distance, can possibly provide insight in postural control beyond the simple understanding of the amount of work done required to perturb an individual. To date, investigation of the relationship between perturbation characteristics which determine the type of correcting response, feet-in-place or stepping, is generally limited to studies using platform-translation (e.g. Jensen et al., 2001) and cable-pull at the waist (e.g. Mille et al., 2003) perturbation methods. There is limited research that examined this relationship following an upper body perturbation in younger or older adults and no studies have compared the relationship between characteristics of balance recovery responses elicited by different perturbation methods, such as platform-translation and cable-pull of the upper body. Moreover, the literature shows discrepancies between the magnitudes of acceleration and displacement used to cause a postural perturbation.

There is promise in utilizing postural perturbation paradigms to investigate dynamic postural control and probe the effects of injury and disease on mobility and fall risk. Dynamic situations, such as those elicited by postural perturbations, are more revealing, as opposed to static scenario (e.g. quiet standing), of mechanism of postural control and balance recovery. However, the knowledge around dynamic postural control and balance emanates from studies which utilize different perturbation methods and different perturbation stimulus characteristics, which makes it difficult to compare findings between studies. Moreover, research that investigated the relationship between perturbation characteristics in determining balance-correcting response and whether age or health affects this relationship is scarce. The existing limited research and anecdotal evidence suggest that balance-correcting responses are method-specific. Thus, work is needed in order to understand the differences between responses induced with different methods before a selection of methods could possibly be used to best understand the deficiencies of different patient groups or special populations (e.g. older adults).

1.2. Dissertation Objectives

The objectives of the current dissertation were: 1) to directly compare two different types of perturbation on postural control and balance recovery, 2) to describe kinematic and neuromuscular responses to different perturbations in order to explore similarities and differences in the nature of the responses, and 3) to examine the effect of age on the relationships between perturbation characteristics in determining balance-correcting response.

1.3. Dissertation Layout

To achieve the dissertation objectives, four studies were conducted and are presented in the subsequent four chapters of the document. The equipment used to induce postural perturbations was designed and constructed by the author (DV). Following the design and construction of the postural perturbation equipment, a multistage Study 1 (Chapter 2; #Thresholds) was conducted during which the equipment was tested, modified, and improved upon. Importantly, common to both platform-translation and shoulder-pull perturbation characteristics (displacement and applied force), which elicited a stepping response in quietly standing participants who behaved naturally, were determined. Studies 2 and 3 (Chapters 3 and 4; #MOS and #EMG) were conducted using the common displacement and applied force parameters established in the first study. Study #MOS explored differences in postural stability during perturbations induced with platform-translation and shoulder-pull perturbation methods. Study #EMG focused on spatiotemporal and neuromuscular similarities and differences in balance-correcting responses induced with platform-translation and shoulder-pull methods. Lastly, Study #YA/OA explored differences in responses between younger and older adults with shoulder-pull perturbation method. The shoulder-pull perturbation was the method of choice in the last study due to higher ecological validity than platform-translation. Moreover, the equipment configured for shoulder-pull involved less moving parts, which improved confidence in the amount of perturbation stimulus applied. The general purpose of this series of studies was to create a solid base for investigation of balance-correcting responses in human participants with standardized methodology which could allow for a direct comparison between perturbation methods to be made.

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CHAPTER 2

Study 1 (#Thresholds): Stepping Threshold with Platform-translation and Shoulder-pull Perturbation Paradigms

2.1. Summary

Introduction: The type of balance-correcting response, feet-in-place or change-in-support (stepping), is predicated on the intensity of perturbation which is often defined by the combination of applied force and displacement. The purpose of the current study was 1) to characterize the custom-built postural perturbation system, which can be configured for platform-translation and shoulder-pull perturbations; 2) to determine the intensity of perturbation required to elicit a stepping response with both methods; and 3) to determine the perturbation stimulus characteristics that are common to both methods.

Methods: First, friction force within the system was measured. The findings were used to calibrate the amount of stimulus used with each participant and both perturbation methods. Then, fourteen young healthy males participated. Unexpected platform translations and shoulder pulls were induced by release of free weights. The weights fell a controlled height exerting a pull on the platform or the participant via a shoulder harness. Participants responded with either feet-in-place or forward stepping responses.

The contents of this chapter represent my work. Parts of this work have been previously presented as a poster at a conference.

Verniba, D., Chaudhari, R., Rahimzadeh Khiabani, R., Gage, W.H., 2014, June. *Stepping thresholds with shoulder pull and translating platform perturbation paradigms.* Presented at the annual meeting of the International Society for Posture and Gait Research (ISPGR), Vancouver, BC, Canada.

The weight (force) and drop height (displacement) were varied to investigate a range of force-displacement combinations required to elicit stepping responses. Force-displacement combinations that elicited stepping responses were recorded and normalized to the participant's body weight (BW) and base of support (BOS; participant's foot length).

Results: The lowest force and associated displacement characteristics that elicited stepping responses showed a significant inverse linear relationship during both platform-translation and shoulder-pull trials. The common force-displacement perturbation characteristics, the intersection between two regression functions, were found to be 8.75%BW and 105%BOS.

Discussion: The amount of friction within the system was measured and corrected for during participant testing. Further, the study identified a linear force-displacement relationship required to elicit a stepping response with platform-translation and shoulder-pull methods. The force-displacement perturbation characteristics methods required to elicit a stepping response common to both were established.

2.2. Introduction

Surface-translation and cable-pull perturbation paradigms are among the most commonly used methods to elicit balance-correcting responses in postural perturbation studies. Balance-correcting responses, elicited with those methods, range from feet-in-place ankle responses to change-in-support responses such as a stepping or upper limb reaching. The individual response is predicated on the intensity of perturbation. In previous research where the translating platform method was used, the nature of the response was determined by a combination of platform acceleration and the duration of acceleration or displacement during platform translation (Jensen et al., 2001; Maki et al., 1996). For research utilizing the motor driven cable-pull method, the nature of the response was determined by a combination of acceleration with which the cable was pulled and the displacement resulting from the pull (Mille et al., 2003; Rogers et al., 2001). Though scarce, there are also studies where free weights were used instead of electrical actuators. In these studies, free weights were allowed to fall a controlled height, and thus, exert force on the participant via a system of cables and pulleys (Chandler et al., 1990; Hsiao-Wecksler et al., 2003).

The investigation into the relationship between perturbation characteristics, which determine the type of correcting response, is sparse and generally limited to platform-translation (Jensen et al., 2001) and waist-pull studies (Mille et al., 2003). To date, no studies have examined such relationship following upper body (shoulder) perturbation. Furthermore, there are no studies which have directly compared platform-translation and shoulder-pull perturbation stimuli in order to examine the relationship between perturbation characteristics.

The purpose of the current study was: 1) to characterize the custom-made perturbation system, which can be configured to produce platform-translation and shoulder-pull perturbation; 2) to investigate the relationship between the perturbation characteristics (applied force and displacement) required to elicit a forward stepping response with platform-translation and a shoulder-pull methods; and 3) to establish the common perturbation characteristics required to produce a forward stepping response using both methods. Thus, the research questions were: 1) what is the magnitude of friction force within the platform-translation and shoulder-pull setup; 2) what perturbation intensity, defined as the combination of applied force and displacement, is required to elicit a forward stepping response in younger adults using both platform-translation and shoulder-pull methods; and 3) what combination of applied force and displacement is common to both perturbation methods?

2.3. Methods

Perturbation equipment design and characteristics

For the current study, DV designed and built the postural perturbation system. The system consisted of six distinct modules (Figure 2.1 and 2.2): a perturbation trigger, a support stand with a mounted electromagnet VISML 600 LED (VSIONIS), a switch tower, two pulley posts, and a translating platform. The stimulation unit (S88K, Grass Technologies, Astro-Med Inc, Rhode Island) served as a perturbation trigger; the stimulation unit was used to stop the flow of electrical current to the electromagnet.

The release of the electromagnet allowed free weights, which were suspended from the switch tower, to fall. The weights produced tension in the steel aircraft cable, which was fed through the pulley system. All pulleys used in the system utilized low friction lubricated metal wheels with ball bearings. The fall of the weights was arrested by a chain that allowed the weights to fall a controlled vertical displacement. The height of the drop was controlled by the addition or removal of chain links. A weight suspended from a movable block was added to the tower design in order to reduce the slack within the aircraft cable. The aircraft cable extended away from the switch tower towards the two pulley posts. The purpose of the pulley posts was to redirect the cable towards a central point where during the experiment both ends of the cable would be linked with either the translating platform or with a participant via the shoulder harness. The translating platform was fitted with low friction lubricated wheels with ball bearings (Figure 2.2). The platform rode atop a guidance rail in order to prevent it from running off the foundation. The dual purpose of the foundation was to insure consistent direction of travel of the platform as well as to smooth the ride of platform, since the floor in the laboratory was uneven.

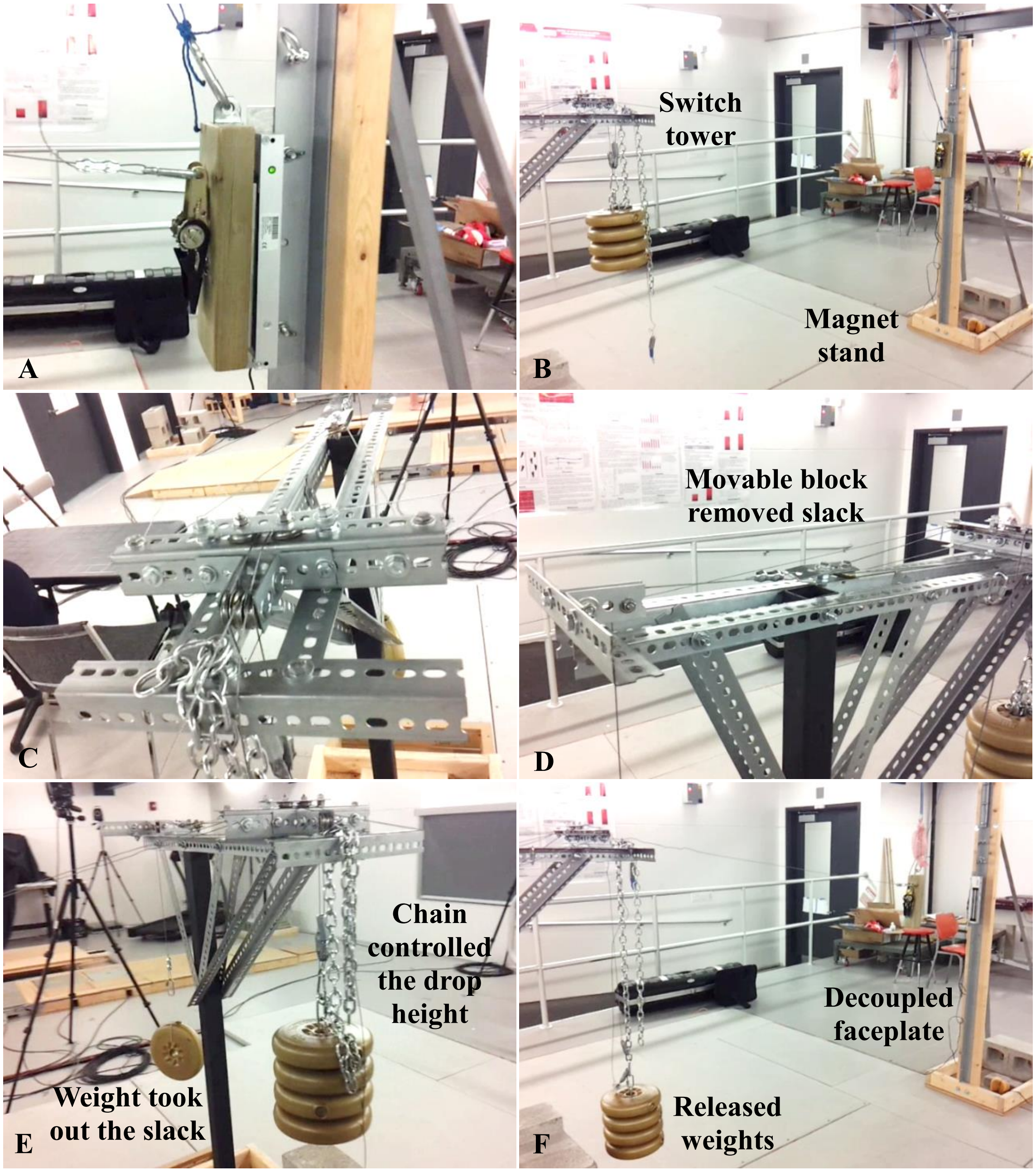

Figure 2.1. A depiction of the perturbation system. The description is from left to right, top to bottom. The system consisted of (A) an electromagnet mounted on a support stand (B). The magnet released free weights that were suspended from the switch tower (B). The weights were linked to an aircraft cable which was fed through a system of pulleys on the switch tower (C). The slack in the pulley system was removed by a movable block (D) and a suspended weight (E). When electric current to the magnet was switched off, the magnet faceplate decoupled from the magnet and released the weights (F). The chain allowed the weights to fall a controlled height. The height was adjusted by adding or removing chain links.

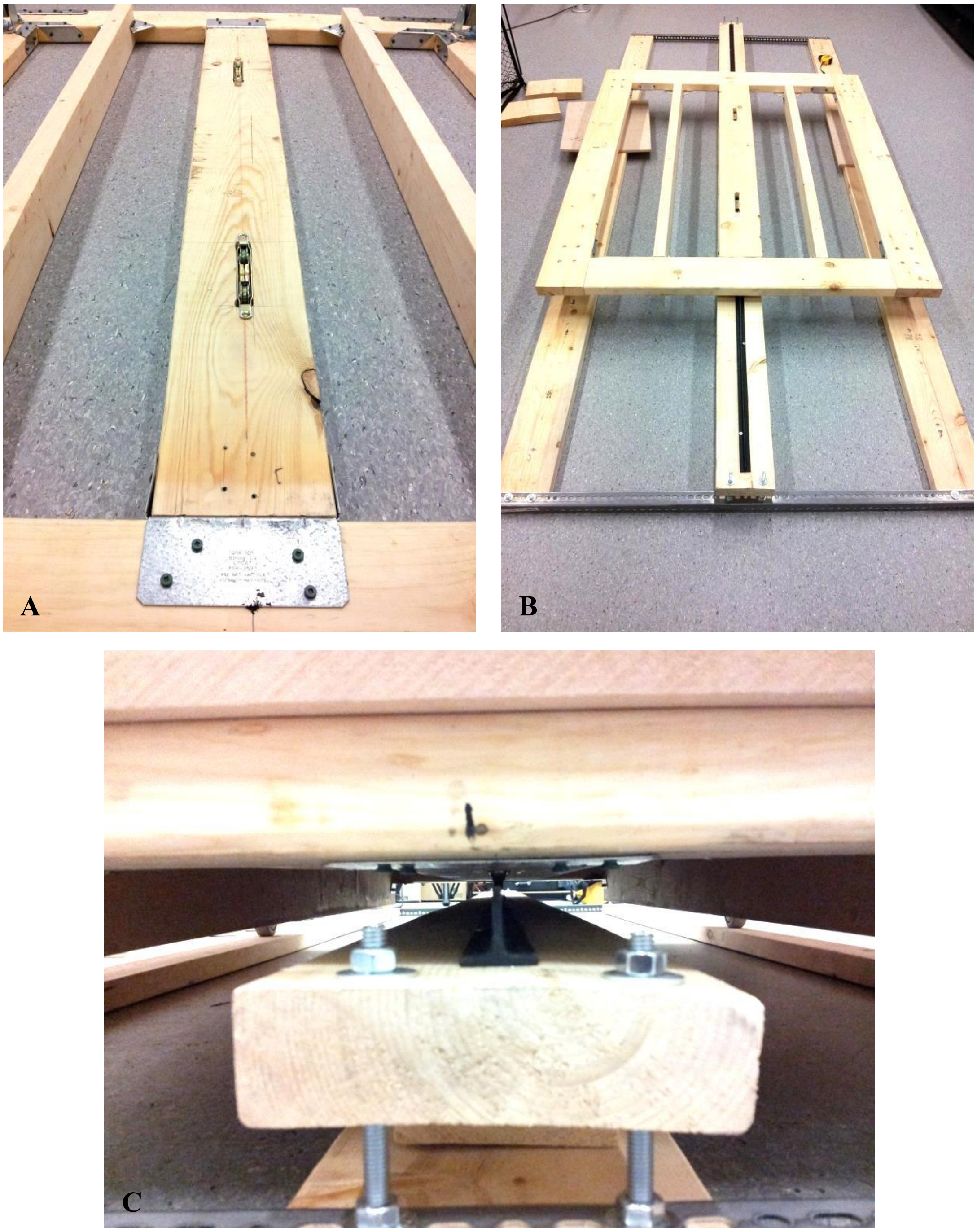

Figure 2.2. The photographs depicting the translating platform and its components during construction. The platform (A) was outfitted with lubricated ball bearing wheels and rolled atop the foundation (B) with a center-mounted rail (C). The purpose of the rail and foundation was to prevent the platform from rolling off the foundation, ensuring consistent heading of the platform during movement, and further reducing friction.

Friction is present in any mechanical system. Although friction within mechanical systems is sometimes beneficial (e.g. vehicle brakes), more often it is unfavorable. In order to make an adequate comparison between perturbation characteristics that are required to elicit stepping responses and to determine perturbation characteristics that are common to both perturbation methods, it was necessary to investigate the friction force within the system with platform-translation and shoulder-pull setups prior to conducting the experiment with participants, as the friction force within the system may differ between the setups. Thus, an investigation to determine the friction force within the cable system and the platform was conducted. To determine the friction force within the cable system, two ends of the cable were linked together and the cable was pulled uniformly. A portable uniaxial load cell (A-Tech Instruments Ltd., Ontario) was coupled in series with the cable and was used to measure the friction force. The load cell was factory calibrated and zeroed prior to testing. Six pull trials for the system alone, without the platform, were conducted. The average friction force was calculated. Next, the platform pulls were performed to determine the rolling friction force. The weight on the platform was progressively increased from 0 to 155kg in approximately 17kg increments. The force on the platform and the rolling friction force data were used to create appropriate adjustment to the free weights used with each participant based on a regression equation for the line of best fit.

Participants

Fourteen young healthy males (age 25.5 ± 2.9 years, height 1.82 ± 0.05m, body weight (BW) 80.6 ± 10.0kg, foot length 27.5 ± 1.0cm; mean ± SD) participated. Participants were excluded if they reported a history of neurological or musculoskeletal disorders; or an injury, pain or surgery on their lower body or back in the six months prior to participation. York University research ethics board provided approval of the methods used in this study. All participants provided informed consent prior to participation.

Set-up and protocol

The platform-translation (PLAT) and shoulder-pull (PULL) perturbation trials were conducted as blocks and performed on two separate days; participants visited the laboratory twice. The two data collection sessions, PLAT and PULL, were counterbalanced across participants. The participant's foot length and body weight were measured before each data collection. Participants remained barefoot for the duration of experiment. Base of support (BOS) length was defined as the participant's foot length. All trials were initiated with participants standing on the platform. For the purpose of PLAT trials, the platform was connected to the cables from both ends. During the PULL trials, stoppers were applied to the platform wheels which prevented it from moving during the experiment. Participants wore a shoulder harness during the PULL trials. The cables were linked to the shoulder harness at approximately the level of the manubriosternal joint from the front and at the T3/T4 vertebrae from the back. Unexpected posterior platform translations and anterior shoulder pulls were induced (Figure 2.3). Unexpectedness of perturbation was achieved by randomly varying the timing of weight release. There were no catch (posteriorly directed perturbations) trials. Participants were instructed as follows: “Behave as naturally as possible. If you don’t have to take a step, don’t take a step. If you feel the need to take a step to avoid falling, do take a step. Do what is natural to avoid falling.” Participants received no explicit instruction in regards to the direction of perturbations.

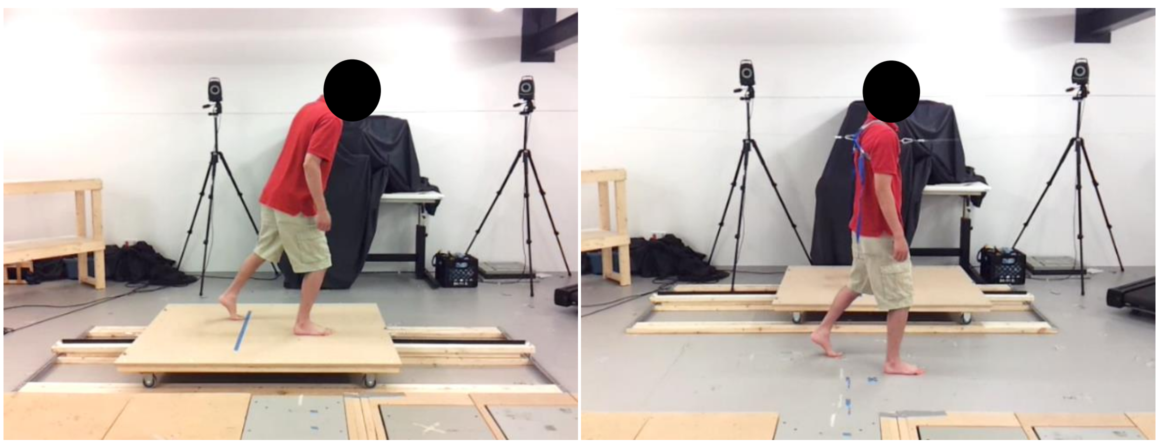

Figure 2.3. The photographs depicting forward stepping responses following the posterior platform-translation (upper panel) and anterior shoulder-pull perturbations (lower panel). Note, not the actual experiment, a demonstration only.

The perturbation stimulus intensity was defined as the combination of applied force (%BW) and displacement (%BOS). The applied force was applied in the range between 2.0%BW and 12%BW (4.45N increments), while the displacement was applied between 50%BOS and 155%BOS (2.2cm increments). The trials were blocked by applied force. The entire range of displacement iterations was applied within each force block, i.e. the testing began with the lowest weight and the lowest displacement. The displacement was progressively increased from trial to trial until it reached the maximum value for each participant, while the force remained unchanged. The force was increased once the entire range of displacement per force block was tested; the testing then resumed with the lowest displacement and the process was repeated until the entire range of displacement was applied. The process was repeated until the combination of the highest force and highest displacement was tested. This protocol was used with half of the participants (n = 7). For the remaining participants (n = 7), the protocol was repeated in reverse order. The testing began with the largest force and the largest displacement. The displacement within each force block was reduced until the lowest displacement was reached, while the force remained unchanged. Once the entire range of displacement was applied, the force was reduced and the protocol was repeated starting with the largest displacement. The process was repeated until the combination of the lowest force and lowest displacement was tested. To offset the effects of fatigue, participants received rest breaks between the force trial blocks. On average, participants performed 2 trials per minute.

Measures of interest

To assess the balance-correcting responses, the primary investigator (DV) stood next to the participant with the line of sight perpendicular to the line of step and in line with the adhesive tape placed in front of participant's toes. Two stepping threshold responses: 1) partial-step, defined as anteriorly directed foot movement that was smaller than 100%BOS, and 2) complete-step, foot movement displacement that was larger than 100%BOS, were recorded by the primary investigator. The feet-in-place balance-correcting response was coded as 0, partial-step as 0.5, and complete-step as 1. The response codes were placed in the 16 x 14 cell pull force-displacement matrix (Figure 2.4). The matrices for all participants were overlaid and the means were calculated for each cell across all individual matrices for PLAT and PULL trials separately. The combination of lowest force and associated displacement, which produced the result of 1 across all matrices for PLAT and PULL trials, were mapped on a scatter plot.

Force

Displacement

		13.4	17.8	22.3	26.8	31.2	35.7	40.1	44.6	49.1	53.5	58.0	62.4	66.9	71.3	75.8	80.3	N
		2.0	2.7	3.3	4.0	4.7	5.3	6.0	6.7	7.3	8.0	8.7	9.3	10.0	10.7	11.3	12.0	BW %
13.2	49																	
15.4	57																	
17.6	65																	
19.8	73																	
22.0	81																	
24.2	90																	
26.4	98																	
28.6	106																	
30.8	114																	
33.0	122																	
35.2	130																	
37.4	139																	
39.6	147																	
41.8	155																	
cm	BOS %																	

Figure 2.4. A sample force-displacement matrix used to record perturbation responses for a participant with the weight of 68kg and BOS length of 27cm. Each blank cell was filled with either 0 (fee-in-place), 0.5 (partial step), or 1 (complete-step).

Statistical analyses

All statistical analyses were conducted using JMP (v8.0, SAS Institute, North Carolina). The linear regression function was fitted to the complete-step threshold scatter plot data. The linear function fit was considered significant at $p < 0.05$. The pull weight and the pull displacement values were expressed as mean ± SE.

2.4. Results

Friction force within the pulley system was determined to be approximately 2N. Figure 2.5 depicts the relationship between the amount of weight on the platform and the platform rolling friction force. Based on the regression function (Figure 2.5), the amount of force used during perturbation was adjusted, as per Table 2.1, for each participant to offset friction force.

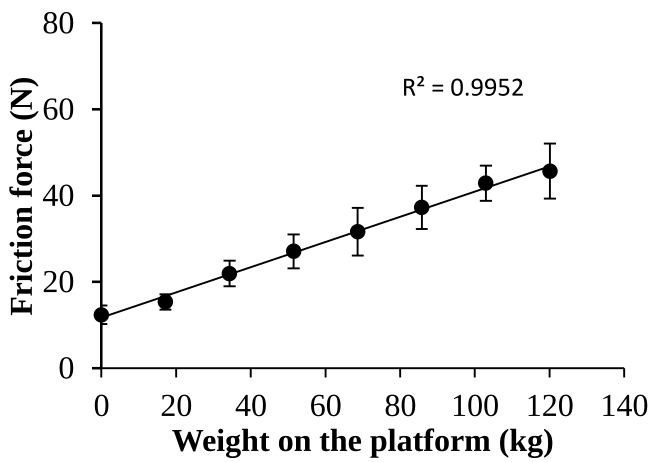

Figure 2.5. A graphical representation of the relationship between the friction force (± SD) and weight on the platform. The regression function was used to establish the force required to compensate for the friction force for each participant.

Table 2.1. Friction force offset per participant's body weight during PLAT trials.

Body Weight (kg)	Offset Force (N)
63.5 - 68.0	33.5
68.0 - 72.5	35.0
72.5 - 77.0	36.0
77.0 - 82.0	38.0
82.0 - 86.5	39.5
86.5 - 91.0	40.5
91.0 - 95.5	42.0
95.5 - 100.0	43.0
100.0 - 104.5	44.5
104.5 - 109.0	46.0

Note: Body Weight and Offset Force were rounded to the nearest 0.5kg and 0.5N, respectively.

The aggregate matrix of all individual force-displacement matrices was created for the PLAT (Figure 2.6) and PULL (Figure 2.7) trials, separately. The combination of the lowest force and associated displacement which resulted in a complete-step responses showed a significant ($R^2 = 0.935$, df = 11, $p < 0.001$) linear inverse relationship between force and displacement for the PLAT trials (Figure 2.8). Likewise, for the PULL trials, complete-step data showed a significant ($R^2 = 0.945$, df = 6, $p < 0.001$) linear inverse relationship between required force and displacement below 123%BOS. Above 123%BOS, a force of approximately 7%BW was required to elicit a complete-step balance-correcting response (Figure 2.8).

The intersection between PLAT and PULL functions represents common force-displacement perturbation characteristics (8.75%BW and 105%BOS) required to elicit a stepping response using both PLAT and PULL perturbation methods (Figure 2.8).

Force

Displacement

BOS %	SE %	2.8	3.4	4.0	4.5	5.1	5.6	6.2	6.8	7.3	7.9	8.4	9.0	9.6	10.1	10.7	11.2
BW % / SE %		0.2	0.2	0.1	0.1	0.1	0.1	0.1	0.0	0.0	0.0	0.1	0.1	0.1	0.1	0.1	0.1
50.3	**0.7**	0.1	0.1	0.1	0.1	0.2	0.1	0.1	0.2	0.5	0.5	0.5	0.8	0.8	0.9	0.9	0.9
58.3	**0.6**	0.0	0.0	0.0	0.1	0.2	0.1	0.1	0.3	0.5	0.6	0.5	0.7	0.9	0.9	0.9	0.9
66.3	**0.6**	0.1	0.0	0.0	0.1	0.2	0.1	0.2	0.4	0.4	0.6	0.7	0.7	0.9	1	1	1
74.2	**0.6**	0.0	0.0	0.0	0.0	0.2	0.1	0.1	0.5	0.6	0.7	0.8	0.8	0.9	1	1	1
82.2	**0.6**	0.0	0.0	0.0	0.2	0.1	0.1	0.2	0.3	0.6	0.7	0.8	0.8	1	1	1	1
90.2	**0.6**	0.0	0.0	0.1	0.0	0.2	0.1	0.3	0.3	0.7	0.8	0.8	1	1	1	1	1
98.2	**0.6**	0.0	0.0	0.0	0.2	0.2	0.1	0.4	0.6	0.8	0.8	0.8	1	1	1	1	1
106.2	**0.7**	0.0	0.0	0.0	0.2	0.2	0.1	0.5	0.5	0.8	0.9	0.9	1	1	1	1	1
114.2	**0.7**	0.0	0.0	0.0	0.2	0.2	0.1	0.5	0.7	0.8	1	1	1	1	1	1	1
122.2	**0.7**	0.0	0.0	0.0	0.2	0.2	0.1	0.6	0.7	0.8	1	1	1	1	1	1	1
130.2	**0.8**	0.0	0.0	0.2	0.2	0.2	0.1	0.5	0.7	0.9	1	1	1	1	1	1	1
138.1	**0.8**	0.0	0.0	0.1	0.2	0.2	0.2	0.6	0.7	1	1	1	1	1	1	1	1
146.1	**0.9**	0.0	0.0	0.1	0.2	0.1	0.3	0.7	0.7	1	1	1	1	1	1	1	1
154.1	**1.0**	0.0	0.1	0.1	0.2	0.3	0.2	0.7	0.8	1	1	1	1	1	1	1	1

Figure 2.6. A map representation of the force-displacement matrix for the range from feet-in-place to complete-step averages (n = 14) for each force-displacement platform translation dyad. The complete-step responses for all participants are coded as 1 and are coloured in light grey. The combination of lowest force and associated displacement, which produced the result of 1, is coloured in dark grey and represents complete-step threshold. The force and displacement values for each dark grey cell within the matrix were used to establish a complete-step threshold relationship.

Force

Displacement BOS %	SE %	2.7	3.3	3.9	4.4	5.0	5.6	6.2	6.7	7.3	7.9	8.5	9.0	9.6	10.2	10.7	11.3
BW % / SE %		0.2	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.2	0.2	0.2
50.7	**0.7**	0.0	0.0	0.0	0.1	0.2	0.2	0.5	0.4	0.3	0.4	0.5	0.5	0.5	0.5	0.5	0.6
58.7	**0.7**	0.0	0.0	0.0	0.1	0.3	0.4	0.5	0.5	0.5	0.6	0.6	0.6	0.6	0.8	0.8	0.7
66.7	**0.6**	0.0	0.0	0.0	0.3	0.4	0.4	0.5	0.5	0.6	0.6	0.7	0.8	0.9	0.9	0.9	0.9
74.7	**0.6**	0.1	0.1	0.1	0.4	0.5	0.5	0.5	0.6	0.7	0.8	0.8	0.9	0.9	0.9	0.9	1
82.7	**0.6**	0.0	0.1	0.1	0.4	0.5	0.6	0.6	0.8	0.9	0.9	0.9	0.9	0.9	0.9	0.9	1
90.7	**0.6**	0.1	0.2	0.2	0.5	0.5	0.6	0.6	0.8	0.9	0.9	0.9	0.9	1	1	1	1
98.7	**0.6**	0.1	0.1	0.4	0.4	0.5	0.6	0.6	0.9	0.9	0.9	0.9	1	1	1	1	1
106.7	**0.6**	0.1	0.3	0.4	0.5	0.5	0.7	0.8	0.9	0.9	0.9	0.9	1	1	1	1	1
114.7	**0.6**	0.1	0.2	0.5	0.6	0.6	0.8	0.8	0.9	0.9	1	1	1	1	1	1	1
122.7	**0.7**	0.1	0.3	0.5	0.6	0.7	0.8	0.9	1	1	1	1	1	1	1	1	1
130.7	**0.7**	0.1	0.3	0.5	0.7	0.8	0.8	0.9	1	1	1	1	1	1	1	1	1
138.7	**0.8**	0.2	0.3	0.5	0.7	0.8	0.8	0.9	1	1	1	1	1	1	1	1	1
146.7	**0.8**	0.2	0.3	0.5	0.7	0.8	0.8	0.9	1	1	1	1	1	1	1	1	1
154.7	**0.9**	0.2	0.4	0.5	0.7	0.8	0.8	0.9	1	1	1	1	1	1	1	1	1

Figure 2.7. A map representation of the force-displacement matrix for the range from feet-in-place to complete-step averages (n = 14) for each force-displacement shoulder pull dyad. The complete-step responses for all participants are coded as 1 and coloured in light grey. The combination of lowest force and associated displacement, which produced the result of 1, is coloured in dark grey and represents complete-step threshold. The associated force and displacement values for each dark grey cell within the matrix were used to establish a complete-step threshold relationship.

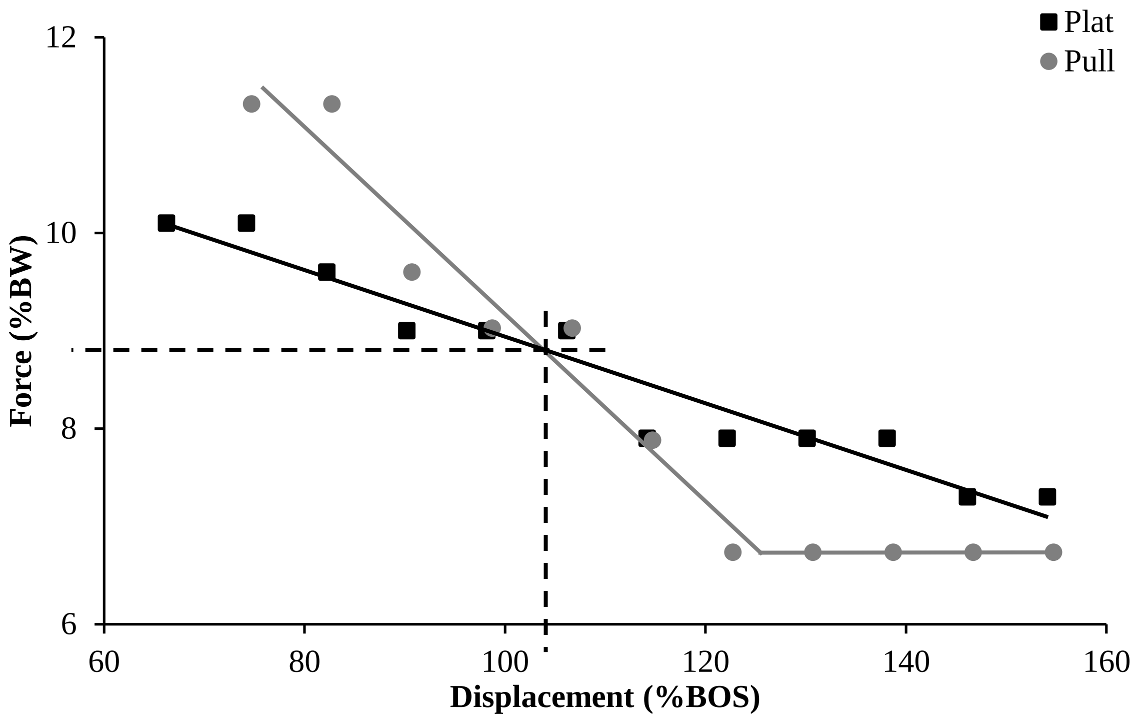

Figure 2.8. The complete-step threshold force-displacement relationship for the PLAT and PULL perturbations depicted in the same figure. The approximate intersection between PLAT and PULL functions (intersecting black dash lines) represents common force-displacement perturbation characteristics (8.75%BW and 105%BOS) required to elicit a stepping response using both perturbation methods.

2.5. Discussion

The purpose of the current study was: 1) to characterize the custom-made perturbation system, which can be configured to produce platform-translation and shoulder-pull perturbation; 2) to determine the intensity of perturbation, defined by force and displacement, required to elicit a forward stepping response with platform-translation and shoulder-pull perturbation methods, and 3) to determine the magnitude of the stimulus required to elicit a stepping correcting response which is common to both perturbation methods.

An inverse linear force-displacement relationship was established for both platform-translation and shoulder-pull methods (Figure 2.8). The interpretation of regression lines is such that any combination of force and displacement on or above the regression lines would produce a complete-step response with the associated perturbation method. Interestingly, it appeared that with the shoulder-pull method, there was a minimum force threshold of approximately 7%BW required to elicit a complete-step response; since beyond approximately 123%BOS, participants did not show complete-step responses with any combination of force lower than 7%BW. With the platform-translation method, however, it appeared that there was no clear minimum force required to elicit a complete-step response in the range examined as the force-displacement relationship appeared to be linear throughout the data set. The common perturbation characteristics were chosen at the intersection between the PLAT and PULL regression lines, which was found to be 8.75%BW of force and 105%BOS of displacement. The intersection between the PLAT and PULL regression lines is the lowest force-displacement combination required to elicit a complete-step response with both perturbation methods.

While previous research (Jensen et al., 2001; Mille et al., 2003; Pai et al., 1998; Rogers et al., 2001) have investigated the relationship between the perturbation characteristics, a direct comparison to the current study is difficult to make. Unlike the current study, where free weights were used to elicit a perturbation, the previous studies utilized perturbation systems that were based on electrical actuators and investigated either platform-translation or waist-pull paradigms exclusively. Further, the current study differs from the previous research in that a larger quantity of combinations of perturbation characteristics was investigated; thus, providing higher data resolution which would allow for enhanced evaluation of the relationship between perturbation characteristics. More importantly, the current study differs from previous research in that both lower and upper body perturbations were investigated using common equipment, procedure, and participants. As a result, the current methodology allowed to determine stepping thresholds for both platform-translation and shoulder-pull perturbation methods as well as determine perturbation characteristics common to both methods. Our plan is to utilize the perturbation characteristic common to both perturbation methods in a study which would compare balance recovery strategies elicited with both methods. In order to make a fair comparison between the balance recovery strategies elicited with two distinct perturbation methods, it is important to ensure that the perturbation stimulus intensity is equivalent between the methods.

2.6. Conclusion

Similar to previous literature, the current study investigated the effect of perturbation force and displacement on balance recovery responses. An inverse linear force-displacement relationship, which is required to elicit a stepping response with platform-translation and shoulder-pull perturbation methods, has been identified. The meaning of the shape of these relationships remains open to interpretation. The next steps may include examination of special populations (e.g. older adults, etc.) to explore factors that may alter the shape and parameters of these relationships. Finally, the current findings can be used to make comparison between balance-correcting strategies elicited with both perturbation methods.

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CHAPTER 3

Study 2 (#MOS): A Comparison of Balance-correcting Responses Induced with Platform-translation and Shoulder-pull Perturbation Methods

3.1. Summary

Introduction: The understanding of postural control mechanisms in humans emanates from studies utilizing a variety of perturbation methods, instructions, and sensory conditions. The use of different perturbation methods may produce method-specific balance-correcting responses. The current study evaluated balance-correcting responses induced with platform-translation and shoulder-pull perturbation methods, and whether absence of vision affects balance-correcting responses differently between perturbation methods.

Methods: Fifteen healthy young males participated. Unexpected forward and backward platform-translation and shoulder-pull perturbations were induced with and without vision (eyes-open and eyes-closed). Participants were asked to behave naturally. Forward stepping trials were analyzed. Margin of stability (MOS), a marker of postural stability, was calculated from the position data of reflective markers placed strategically around the body. MOS was reported at step initiation (start) and at foot contact. Smaller MOS was

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The contents of this chapter represent my work. Parts of this work have been previously presented as a poster at a conference.

Verniba, D., Tokunaga, J., Gage, W.H., 2015, June. *A comparison of platform translation and shoulder pull paradigms through the analysis of balance correcting responses induced using both methods.* Presented at the annual meeting of the International Society for Posture and Gait Research (ISPGR), Seville, Spain.

interpreted as to suggest poorer postural stability and to be less favorable for balance recovery.

Results: MOS was smaller at step start and at foot contact during platform-translation (0.008±0.006m and 0.092±0.007m, respectively) than at each time point during shoulder-pull (0.037±0.007m and 0.173±0.007m, respectively). The absence of vision did not affect MOS at step start. At foot contact during platform-translation with eyes-closed, MOS was larger (0.106±0.008m) than with eyes-open (0.079±0.007m) but not different between eyes-open and eyes-closed during shoulder-pull (0.170±0.007m and 0.177±0.006m, respectively). Participants required a second step to recover balance in 14% of the platform-translation, as opposed to 3% of the shoulder-pull trials.

Discussion: During platform-translation trials participants demonstrated smaller MOS which placed them in a less favorable circumstance for balance recovery. The absence of vision did not affect MOS in shoulder-pull trials, but did in platform-translation trials. Platform-translation appears to be more challenging than shoulder-pull perturbation, which confirms previous findings. The current research highlights differences in MOS and underscores that caution is required when interpreting results of studies utilizing different perturbation paradigms.

3.2. Introduction

The understanding of dynamic postural control in humans emanates from studies that utilize a variety of perturbation methods, participant instructions, and sensory conditions. There are two general types of postural perturbation methods used to probe postural control and balance recovery in human participants: the perturbation stimulus is applied by moving the surface that participant stands on (i.e. support-surface perturbation) or the perturbation stimulus is applied directly to participant's body. Support-surface perturbations can be achieved by using linear translation (e.g. Hlavacka and Horak, 2006; Pai et al., 2000) or angular rotation of the surface that the participant stands on (e.g. Akram et al., 2008; Gage et al., 2007). Pull (e.g. Hsiao-Wecksler et al., 2003; Mille et al., 2003), tether release (e.g. Arampatzis et al., 2008; Verniba and Gage, 2014), or hold and release (e.g. Bortolami et al., 2003; Krebs et al., 2001) methods, to name a few, apply the stimulus directly to participant's body. Though the aforementioned methods are widely used in postural control research, each has its unique characteristics that may be considered advantageous or disadvantageous when investigating a particular hypothesis.

Postural control is dependent on the integration of sensory inputs from somatosensory, visual, and vestibular systems. Researchers often combine sensory manipulations with postural perturbations when investigating dynamic postural control to examine the contribution of sensory information to balance recovery. For example, vision may be removed (eyes closed) to investigate the contribution of somatosensory and vestibular systems to dynamic postural control (Verniba and Gage, 2014). It has been well-established that reduction or disruption in sensory input results in poorer postural control in patient populations (Creath et al., 2008; Hlavacka and Horak, 2006; Horak et al., 1990; Lackner et al., 1999). Interestingly, unlike patients, healthy adults have shown superior balance recovery ability with eyes closed, though likely due to adoption of more conservative balance-correcting responses that might expose the individual to later increased risk of balance disruption (Verniba and Gage, 2014). While it has been shown, using a variety of perturbation methods, that sensory manipulation has a definite effect on postural control in various populations; it is unclear whether sensory manipulation has similar effects on balance-correcting responses. For example, different perturbation methods may result in different forces applied in various body locations, such as the feet and ankles or the head. Further, it can be argued that during perturbation where force is applied near the feet, as opposed to near the head, the head may not displace as fast and as far, which could result in lower stimulation of the vestibular and visual systems and a delayed detection of a postural threat.

There are various measures used to quantify postural control and balance recovery. Centre of mass and centre of pressure displacement during balance recovery are the most common. In recent years, estimation of margin of stability (MOS) has been reported more frequently (e.g. Arampatzis et al., 2008; Suptitz et al., 2013). MOS emphasizes dynamic relationship between the participant’s centre of mass (COM) and base of support (BOS) by taking into account the location of the participant’s COM with respect to the BOS, as well as COM velocity and participant's height (Arampatzis et al., 2008). Smaller MOS during balance recovery suggests poorer dynamic postural control.

While it is evident that various postural perturbation methods (e.g. surface-translation and cable-pull) are different, it is not clear whether balance-correcting responses induced with these methods are stereotyped and whether sensory challenging conditions, such as absence of vision, has different effects on balance-correcting responses induced with both methods. Mansfield and Maki (2009) provided some evidence that balance-correcting responses may be specific to the perturbation method used and that platform perturbations appear to be more destabilizing than waist-pull perturbations (Mansfield and Maki, 2009). However, the literature is sparse and there is a need to understand whether dynamic postural control is perturbation-specific. The purpose of the current study was to evaluate balance-correcting responses induced using two different postural perturbation methods: platform-translation and shoulder-pull. These methods are among the most commonly used, ecologically valid, and spatially and temporally unpredictable perturbation methods. Furthermore, the current study investigated whether absence of vision affects balance-correcting responses with both methods, equally. The research questions were: 1) is there a difference between balance-correcting responses induced with platform-translation and shoulder-pull perturbation methods, and 2) does absence of vision affect balance-correcting responses induced with both methods, equally? It was hypothesized that: 1) participants would show poorer dynamic balance control, defined as smaller MOS, with platform-translation compared to the shoulder-pull method, and that 2) vision would have a larger effect on balance-correcting responses induced with shoulder-pull perturbation, as opposed to platform-translation.

3.3. Methods

Participants

A new group of participants was recruited for this study (#MOS). Healthy males (n = 15, age 24.3 ± 3.0 years, height 181.2 ± 5.9cm, body weight 82.0 ± 14.0kg, foot length 27.2 ± 1.0cm; mean ± SD) participated. Participants were excluded if they reported a history of neurological or musculoskeletal disorders; or an injury, pain or surgery on their lower body or back in the six months prior to participation. York University research ethics board provided approval of the methods used in this study. All participants provided informed consent prior to participation. A sample size calculation was based on preliminary data obtained with the first five participants. The results indicated that a total of 15 participants would provide adequate statistical power (> 80%) to detect difference ($p < 0.05$) between perturbation method means in the MOS measure.

Set-up and protocol

Infrared reflective markers were placed on bony landmarks, as per C-motion recommendation (C-Motion, 2015), to produce a 13-segment kinematic model: head, trunk, left and right upper arm, left and right lower arm, pelvis, left and right thigh, left and right shank, and left and right foot (Leardini et al., 2007). In the current study, all offline processing was conducted using Visual3D software (v4.84.0, C-Motion Inc., Ontario). Visual3D software has been previously successfully utilized in published work (Joao et al., 2014; Verniba et al., 2015). Marker movement was recorded using a 7-camera motion capture system (MX40, Vicon, Colorado). Marker position was sampled at a frequency of 100Hz and subsequently filtered offline using a digital Butterworth 4^{th} order low-pass filter with an 8Hz cut-off. The cut-off frequency was determined using a residual analysis approach (Winter, 2005).

Participants were barefoot for the duration of the experiment. All trials were initiated with participants standing on a custom-made platform, which was used to induce platform-translation perturbations. During platform-translation trials (PLAT), the platform was linked to the perturbation device via cables. The platform was then pulled either anteriorly or posteriorly to elicit a postural perturbation in the sagittal plane. During shoulder-pull trials (PULL), the platform wheels were locked in order to prevent the platform from moving during the trials. Participants wore a shoulder harness that was affixed to the perturbation device via cables. In PULL trials, participants were pulled either anteriorly or posteriorly by the shoulder harness to elicit shoulder-pull perturbations in the sagittal plane.

Unexpected postural perturbations were induced by the release of an electromagnet that was attached to weights and cables. In turn, the cables were attached to either a platform on which the participant stood on or directly to the participant via a shoulder harness. Upon release of the magnet, the weight was allowed to fall a pre-determined height which produced a tension force within the cable system that created a perturbation-inducing movement. The intensity of the perturbation stimulus was set individually for each participant as a combination of two perturbation parameters: the applied force and the displacement that the weights were allowed to fall. The force of the pull and displacement were set to 8.75% of the participant's body weight (BW) and 105% of the participant's BOS length, respectively. The BOS length was defined as the length of the participant's foot. To make a valid comparison between the perturbation methods, it was important to use equivalent perturbation stimuli with both perturbation methods. The parameters of 8.75%BW and 105%BOS were determined experimentally in the #Thresholds study to induce a stepping response in every trial using both PLAT and PULL methods; in combination, these parameters produced perturbation stimuli equivalent to both methods. Participants were instructed as follows: "Behave as naturally as possible. If you don't have to take a step, don't take a step. If you feel the need to take a step to avoid falling, do take a step. Do what is natural to avoid falling." Same instructions were given to all participants in both PLAT and PULL trials.

Each participant completed eight trials per vision condition (eyes-open: EO/eyes-closed: EC), 16 trials per direction of perturbation which elicited forward (FWD) and backward (BWD) stepping response, and 32 trials per perturbation method (PLAT/PULL) for a total of 64 trials per participant (Figure 3.1). For the vision condition trials, participants wore a blindfold and asked to maintain their eyes closed. Upon the completion of each EC trial, participants were allowed to remove the blindfold, open their eyes, and reposition themselves on the platform for the following trial. The order of trials (FWD/BWD with EO/EC) within the perturbation method block was randomized. The trials that resulted in a posterior step (i.e. anterior platform-translation and posterior shoulder-pull trials) were treated as catch trials to prevent participants from anticipating the direction of perturbation and were not included for further data analyses. The trials were blocked by perturbation method. The order of perturbation-method blocks was counterbalanced across participants. On average, participants performed one trial per minute. In addition, rest/recovery breaks were provided between perturbation method blocks to attenuate the effects of fatigue.

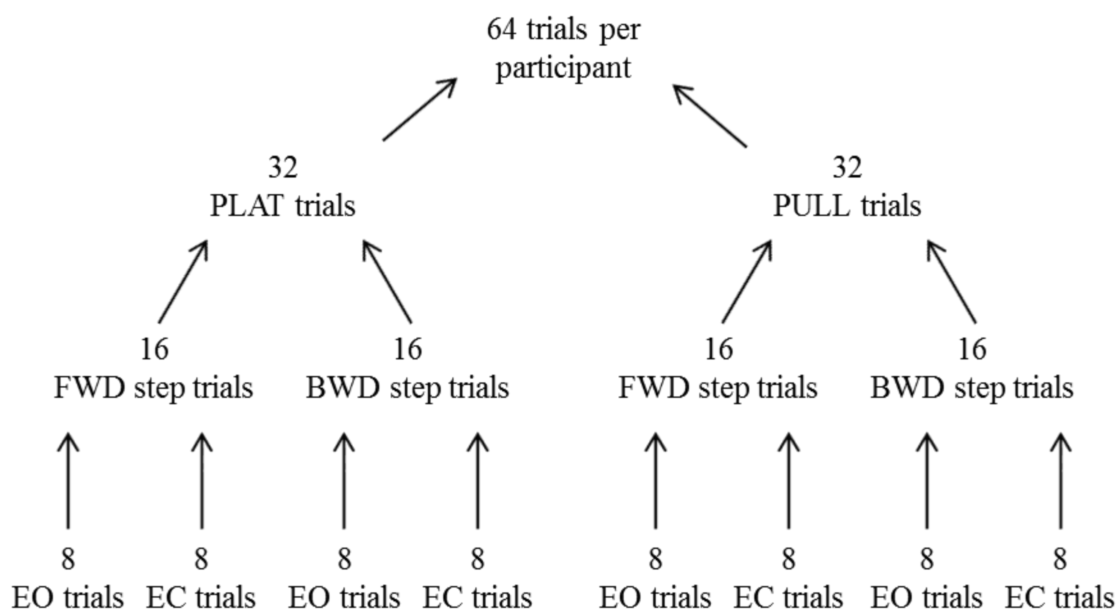

Figure 3.1. A diagram depicting the arrangement of trials and conditions for each participant. PLAT and PULL blocks were counterbalanced between participants, all trials within PLAT and PULL blocks were randomized.

Events and measures of interest

Three events were created using a threshold algorithm: onset (perturbation onset), step start, and foot contact. The onset event was defined as the first positive inflection of the horizontal anteroposterior component of the heel marker velocity signal in PLAT trials, and at the first positive inflection of the horizontal component of the C7 marker velocity signal in PULL trials. Marker velocity signals were derived from marker position signals using a 3-point central finite difference. The first positive inflection of the horizontal component of the heel marker velocity signal in PLAT trials and the first positive inflection of the horizontal component of chest marker velocity signal in PULL trials preceded that of the first inflection of the horizontal component of velocity signal of any other marker affixed to participant's body; thereby, confirming that the segments of the participant's body nearest to the points of perturbation stimulus application were the first to initiate movement with respect to the rest of the body (Figure 3.2). The step start event was defined as the first positive inflection of the vertical component of the stepping foot heel marker velocity signal. The foot contact event was defined as the global minimum of the vertical component of the stepping foot heel marker velocity signal. A similar velocity-based event detection methodology has been previously validated and is discussed in detail elsewhere (O'Connor et al., 2007).

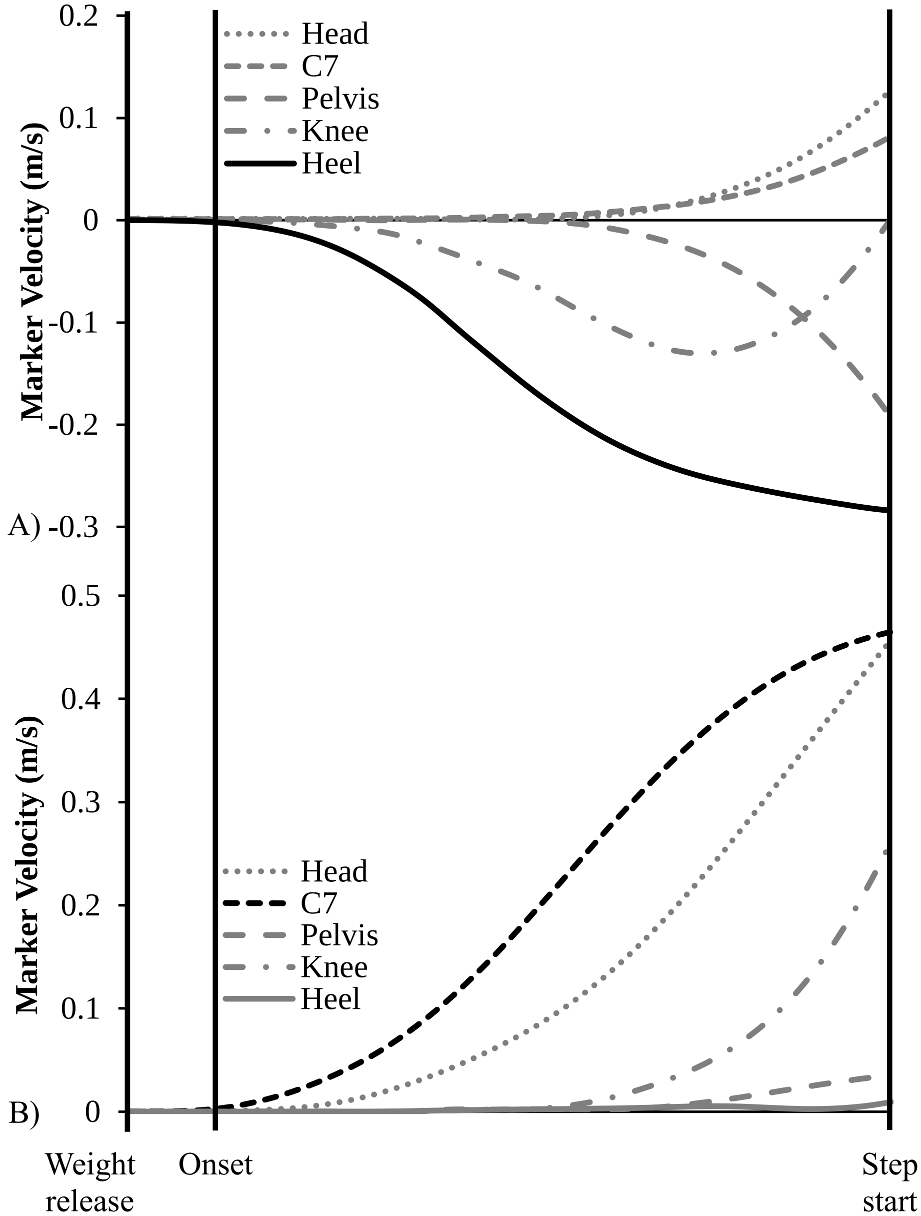

Figure 3.2. A graphical representation of averaged (n = 15) time series of the horizontal velocity signals for the head, base of the neck (C7), pelvis, knee, and the heel markers for A) PLAT and B) PULL trials. All signals were time-normalized between the weight release and step start events; the x-axis represents % time between the events. The signal of the marker that was used to detect initiation of perturbation is highlighted with black color. For the PLAT trials heel marker and for the PULL trials C7 marker velocity signals were chosen as indicators of perturbation initiation and creation of the onset event. The heel and C7 markers were always the first to initiate the motion during PLAT and PULL trials, respectively. The vertical line represents the time point at which the onset event was created.

Whole-body COM was calculated as the weighted average of individual segment COM position data for all 13 segments of the body. The individual segment COM locations were estimated from marker position and anthropometric data as per Dempster et al. (1959). Estimation of the whole-body COM position using Visual3D algorithm has been previously validated (Segers et al., 2007). The MOS was calculated as follows:

$$MOS = BOS - COM_{extr} \tag{1}$$

where *BOS* is the horizontal position of the anterior boundary of BOS, defined as the location of the marker affixed to the 1st metatarsal head of the stepping foot (the distance between the marker and the tip of the big toe was added so that the MOS measure is based on the true size of the BOS); and COM_{extr} is the horizontal position of the extrapolated COM. The extrapolated COM was calculated as follows:

$$COM_{extr} = COM + COM_v / \sqrt{(g/l)} \tag{2}$$

where *COM* is the horizontal (anteroposterior) location of COM, COM_v is the horizontal (anteroposterior) velocity of COM, g is the acceleration due to gravity, and l is the distance between COM and the centre of the ankle joint of the stance limb in sagittal plane (Arampatzis et al., 2008).

Primary dependent measure MOS was reported at step start and at foot contact events. In addition, secondary dependent measures: head velocity at step start and foot contact, step latency, step time, step length, distance between COM and BOS at step start and foot contact, velocity of COM with respect to BOS at step start and foot contact, and average velocity of COM with respect to BOS during the onset to step start epoch, were constructed to further investigate differences noted in MOS measure between PLAT and PULL trials. Head velocity was calculated using the central finite difference of the average of all head marker signals. Step latency was defined as the time between the onset event and step start. Step time was defined as the time between step start and the foot contact event. Step length was calculated as the displacement of the heel marker of the stepping foot. The distance between COM and BOS was calculated by subtracting the signal of the horizontal position of the whole-body COM from the position signal of the marker affixed to the 1^{st} metatarsal head. The COM velocity, with respect to BOS, was calculated by single-differentiating signal of the distance between COM and BOS.

Statistical analysis

Statistical analyses were conducted using JMP (v8.0, SAS Institute, North Carolina). A two-factor (*method* [PLAT/PULL] x *vision* [EO/EC]) mixed effects (*participant* – random effect, *method* and *vision* – fixed effect) repeated measures analysis of variance (rmANOVA) was used to test for differences in the primary and secondary dependent measures between perturbation methods. Catch trials were excluded from the analysis. Contrast analysis with Tukey HSD correction was performed to compare means and test interactions. Values are reported as mean ± SE. The effect size was reported using generalized eta squared (η_G^2) and considered trivial (< 0.02), small (0.2 – 0.12), moderate (0.13 – 0.25), and large (≥ 0.26) (Bakeman, 2005).

3.4. Results

Primary measures

MOS at step start: There was no significant interaction effect (F(1,14) = 1.96; p = 0.183; η_G^2 = 0.01) or main effect of *vision* (F(1,14) = 3.17; p = 0.097; η_G^2 = 0.01); the main effect of *method* (F(1,14) = 12.38; p = 0.003; η_G^2 = 0.27) was significant. MOS was smaller in PLAT trials than in PULL trials (Table 3.1; Figure 3.3A).

MOS at foot contact: There was a significant interaction effect (F(1,14) = 6.60; p = 0.022; η_G^2 = 0.03) observed; the main effects of *method* (F(1,14) = 142.96; p < 0.001; η_G^2 = 0.69) and *vision* (F(1,14) = 27.85; p < 0.001; η_G^2 = 0.09) were also significant. MOS in the PLAT-EO condition was smaller than in the PLAT-EC. MOS in both PLAT-EO and PLAT-EC was lower than in PULL-EO and PULL-EC condition, which were not different from each other (Table 3.1; Figure 3.3B).

Table 3.1. Summary of the results for the primary and secondary measures.

	Plat		Pull		*p*-value and effect size		
Primary measures	**EO**	**EC**	**EO**	**EC**	**Interaction**	**Method**	**Vision**
MOS at step start (m)	0.003 (0.008)	0.012 (0.005)	0.036 (0.007)	0.038 (0.007)	0.183 t	0.003 **L**	0.097 t
MOS at foot contact (m)	0.079 (0.007)	0.106 (0.008)	0.170 (0.007)	0.177 (0.006)	0.022 s	< 0.001 **L**	< 0.001 s
Secondary measures							
Head velocity at start (m/s)	0.142 (0.018)	0.113 (0.015)	0.465 (0.032)	0.448 (0.029)	0.423 t	< 0.001 **L**	0.058 t
Head acceleration at start (m/s^2)	1.774 (0.263)	1.858 (0.273)	2.614 (0.277)	2.682 (0.280)	0.933 t	0.011 **M**	0.357 t
Step latency (s)	0.251 (0.010)	0.226 (0.009)	0.247 (0.012)	0.234 (0.013)	0.474 t	0.917 t	0.004 s
Step time (s)	0.341 (0.013)	0.321 (0.012)	0.359 (0.010)	0.349 (0.011)	0.381 t	0.039 s	0.019 s
Step length (m)	0.347 (0.017)	0.290 (0.015)	0.333 (0.014)	0.320 (0.016)	0.002 s	0.665 t	< 0.001 s
Distance COM to BOS at start (m)	0.017 (0.005)	0.019 (0.004)	0.047 (0.006)	0.046 (0.005)	0.185 t	< 0.001 **L**	0.904 t
Distance COM to BOS at contact (m)	0.220 (0.010)	0.212 (0.011)	0.265 (0.013)	0.263 (0.012)	0.407 t	< 0.001 **M**	0.181 t
COM velocity at start (m/s)	0.247 (0.015)	0.224 (0.013)	0.239 (0.013)	0.231 (0.013)	0.074 t	0.991 t	< 0.001 s
COM velocity at contact (m/s)	0.662 (0.040)	0.546 (0.026)	0.509 (0.023)	0.478 (0.023)	0.002 s	0.012 **M**	< 0.001 s
Ave COM velocity onset to start (m/s)	0.178 (0.007)	0.171 (0.008)	0.106 (0.006)	0.104 (0.006)	0.104 t	< 0.001 **L**	0.009 t

Note: Mean (SE). Probability presented is for the interaction effect between *method* and *vision*, and the main effects of *method* and *vision*. The effect size interpretation is "t" for trivial, "s" for small, "M" for moderate, and "L" for large.

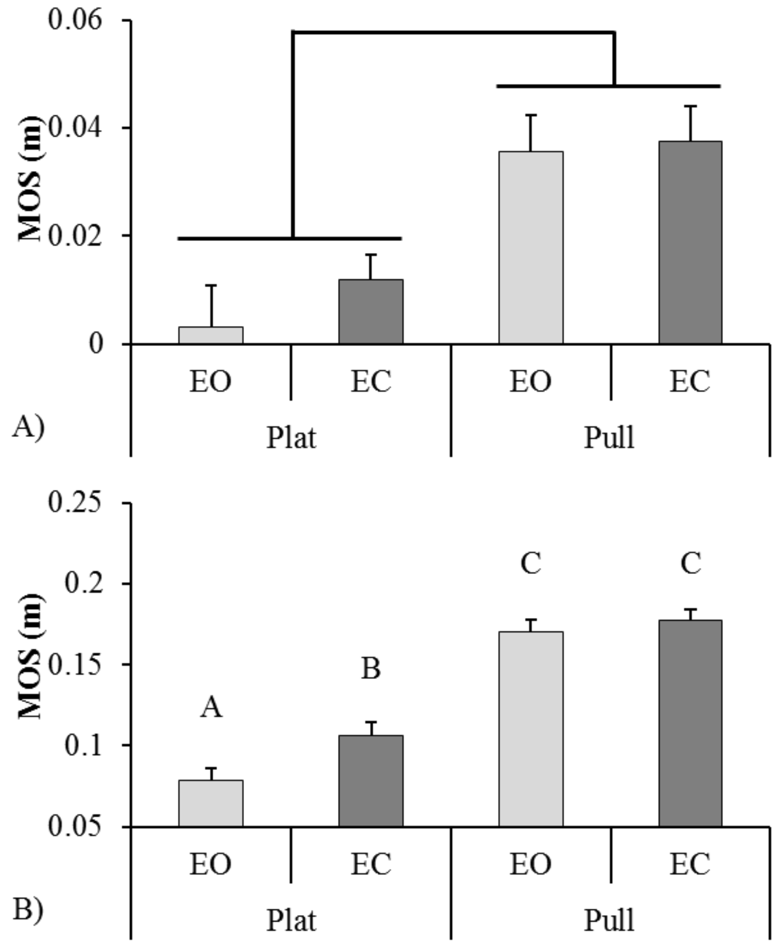

Figure 3.3. A graphical representation of MOS means A) at step start and B) at foot contact events. The error bars are SE. Bracket represents significant main effect of *method*. Letters represent significant interaction between *method* and *vision*, levels not connected by same letter are significantly different. A) MOS was smaller during PLAT trials than during PULL trials at step start. B) At foot contact MOS was larger during PLAT-EC trials than during PLAT-EO trials, but smaller than during both PULL-EO and PULL-EC trials, which were not different from each other.

Secondary measures

Velocity of the head at step start: There was no significant interaction effect (F(1,14) = 0.68; p = 0.423; $\eta_G^2 < 0.01$) or main effect of *vision* (F(1,14) = 4.27; p = 0.058; $\eta_G^2 = 0.02$) observed; the main effect of *method* (F(1,14) = 161.66; p < 0.001; $\eta_G^2 = 0.76$) was significant. The velocity of the head was lower in PLAT trials than in PULL trials (Table 3.1; Figure 3.4A).

Acceleration of the head at step start: There was no significant interaction effect (F(1,14) = 0.01; p = 0.933; $\eta_G^2 < 0.01$) or main effect of *vision* (F(1,14) = 0.89; p = 0.357; $\eta_G^2 < 0.01$) observed; the main effect of *method* (F(1,14) = 8.66; p = 0.011; $\eta_G^2 = 0.16$) was significant. The acceleration of the head was lower in PLAT trials than in PULL trials (Table 3.1; Figure 3.4B).

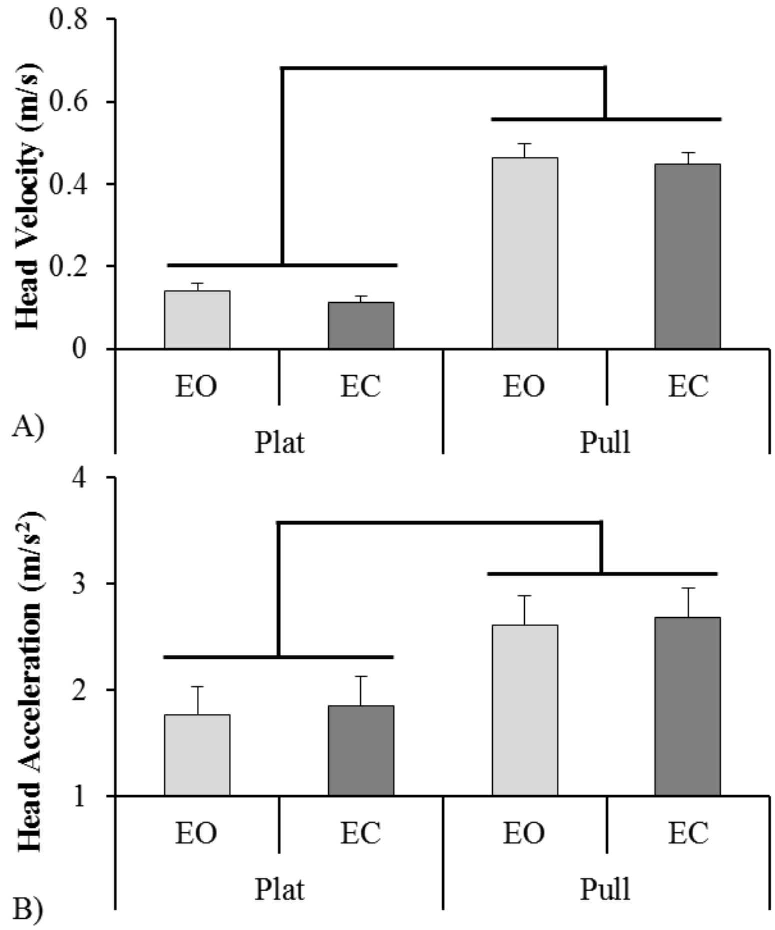

Figure 3.4. A graphical representation of the head A) velocity and B) acceleration means at step start event. The error bars are SE. Brackets represent significant main effect of *method*. Both A) velocity and B) acceleration of the head reported at step start event were smaller during PLAT trials than during PULL trials.

Step latency: There was no significant interaction effect ($F(1,14) = 0.54$; $p = 0.474$; $\eta_G^2 < 0.01$) or main effect of *method* ($F(1,14) = 0.01$; $p = 0.917$; $\eta_G^2 < 0.01$) observed; the main effect of *vision* ($F(1,14) = 11.64$; $p = 0.004$; $\eta_G^2 = 0.04$) was significant. Step initiation latency was lower in EC trials than in EO trials (Table 3.1; Figure 3.5A).

Step time: There was no significant interaction effect ($F(1,14) = 0.82$; $p = 0.381$; $\eta_G^2 < 0.01$); the main effect of *method* ($F(1,14) = 5.20$; $p = 0.039$; $\eta_G^2 = 0.07$) and *vision* ($F(1,14) = 7.00$; $p = 0.019$; $\eta_G^2 = 0.03$) were significant. Step time was lower in PLAT trials than in PULL trials. Step time was lower in EC trials than in EO trials (Table 3.1; Figure 3.5B).

Step length: There was a significant interaction effect ($F(1,14) = 13.62$; $p = 0.002$; $\eta_G^2 = 0.04$) and the main effect of *vision* ($F(1,14) = 37.07$; $p < 0.001$; $\eta_G^2 = 0.08$) was significant; the main effect of *method* ($F(1,14) = 0.20$; $p = 0.665$; $\eta_G^2 < 0.01$) was not significant. Step length was lower in PLAT-EC trials than in PULL-EC trials, which was lower than in both PLAT-EO and PULL-EO trials, which were not different from each other (Table 3.1; Figure 3.5C).

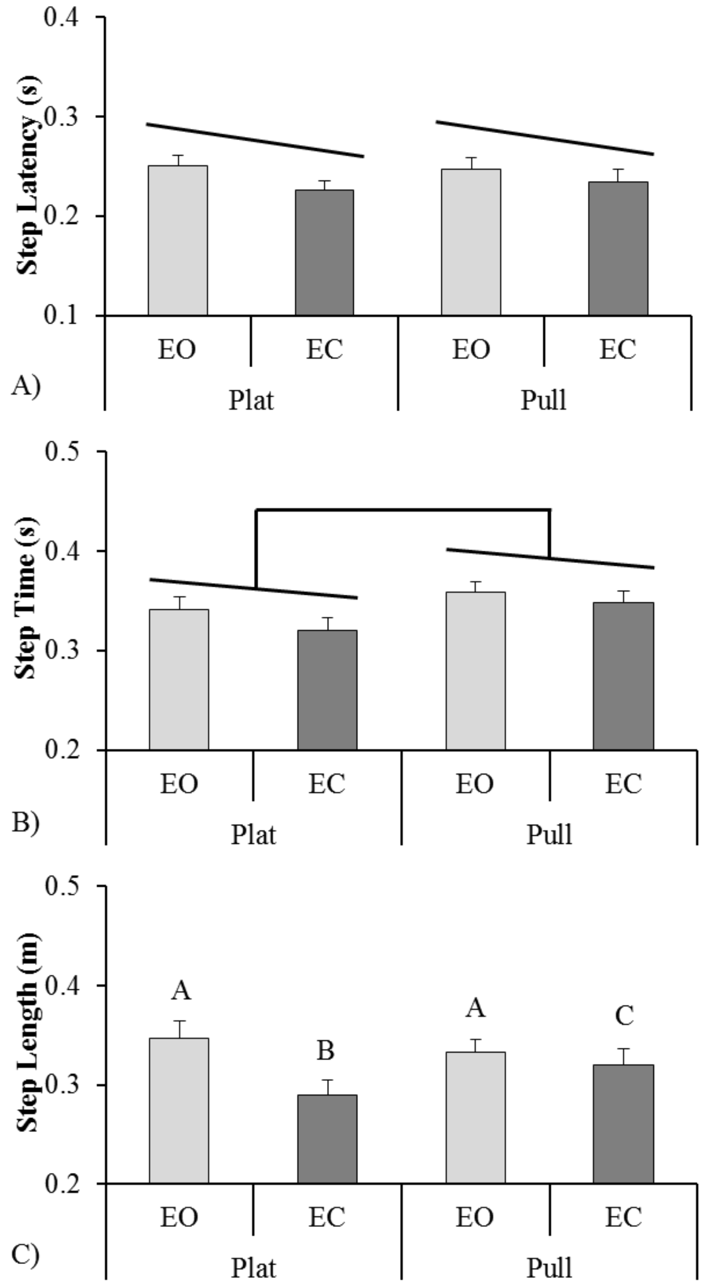

Figure 3.5. A graphical representation of A) step latency, B) step time, and C) step length means. The error bars are SE. Slanted lines represent significant main effect of *vision* and the associated direction of the effect. Bracket represents significant main effect of *method*. Letters represent significant interaction between *method* and *vision*, levels not connected by same letter are significantly different. A) Step latency was smaller in EC trials than in EO trials. B) Step time was smaller in PLAT trials than in PULL trials. Step time was smaller in EC trials than in EO trials. C) Step length was larger in PULL-EC trials than in PLAT-EC trials, but lower than in both PLAT-EO and PULL-EO trials, which were not different from each other.

Distance between COM and BOS at step start: There was no significant interaction effect ($F(1,14) = 1.95$; $p = 0.185$; $\eta_G^2 < 0.01$) or main effect of *vision* ($F(1,14) = 0.02$; $p = 0.904$; $\eta_G^2 < 0.01$) observed; the main effect of *method* ($F(1,14) = 55.75$; $p < 0.001$; $\eta_G^2 = 0.38$) was significant. Distance between COM and BOS was lower in PLAT trials than in PULL trials (Table 3.1; Figures 3.6A).

Distance between COM and BOS at foot contact: There was no significant interaction effect ($F(1,14) = 0.73$; $p = 0.407$; $\eta_G^2 < 0.01$) or main effect of *vision* ($F(1,14) = 1.98$; $p = 0.181$; $\eta_G^2 < 0.01$) observed; the main effect of *method* ($F(1,14) = 30.74$; $p < 0.001$; $\eta_G^2 = 0.24$) was significant. Distance between COM and BOS was lower in PLAT trials than in PULL trials (Table 3.1; Figures 3.6B).

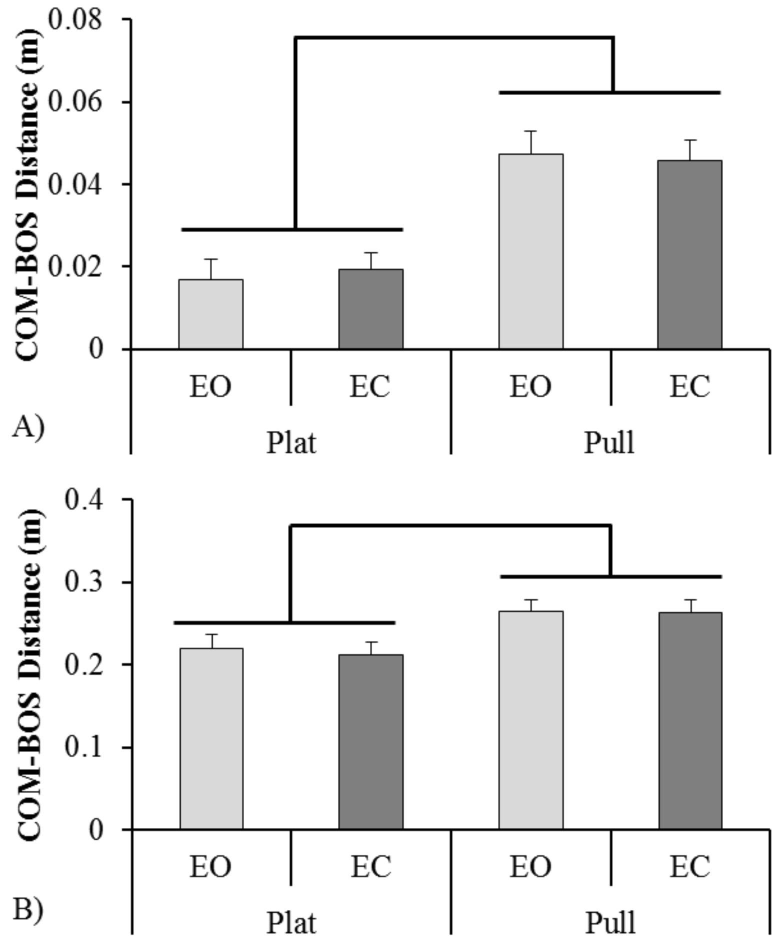

Figure 3.6. A graphical representation of distance between COM and BOS means A) at step start and B) at foot contact. The error bars are SE. Brackets represent significant main effect of *method*. Distance between COM and BOS was lower during PLAT than PULL trials at step start and at foot contact.

COM velocity with respect to BOS at step start: There was no significant interaction effect ($F(1,14) = 3.72$; $p = 0.074$; $\eta_G^2 = 0.01$) or main effect of *method* ($F(1,14) < 0.01$; $p = 0.991$; $\eta_G^2 < 0.01$) observed; the main effect of *vision* ($F(1,14) = 28.89$; $p < 0.001$; $\eta_G^2 = 0.02$) was significant. COM velocity was lower in EC trials than in EO trials (Table 3.1; Figures 3.7A).

COM velocity with respect to BOS at foot contact: There was a significant interaction effect ($F(1,14) = 15.17$; $p = 0.002$; $\eta_G^2 = 0.04$) observed; the main effects of *method* ($F(1,14) = 8.38$; $p = 0.012$; $\eta_G^2 = 0.22$) and *vision* ($F(1,14) = 48.21$; $p < 0.001$; $\eta_G^2 = 0.11$) were also significant. COM velocity in the PLAT-EO condition was higher than in the PLAT-EC. COM velocity in PLAT-EO and PLAT-EC was higher than in PULL-EO and PULL-EC condition, which were not different from each other (Table 3.1; Figures 3.7B).

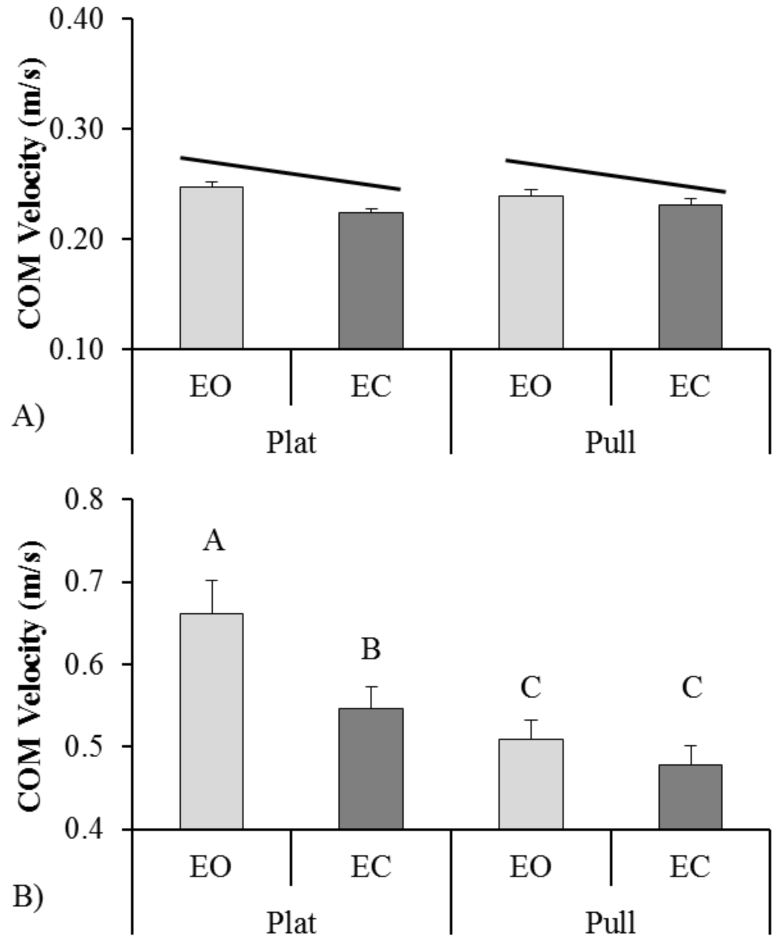

Figure 3.7. A graphical representation of COM velocity with respect to BOS means A) at step start and B) at foot contact. The error bars are SE. Slanted lines represent significant main effect of *vision* and the associated direction of the effect. Letters represent significant interaction between *method* and *vision*, levels not connected by same letter are significantly different. A) At step start COM velocity with respect to BOS was lower in EC trials than in EO trials. B) At foot contact COM velocity was smaller during PLAT-EC than during PLAT-EO, but larger than during both PULL-EO and PULL-EC trials, which were not different from each other.

Average COM velocity with respect to BOS between onset and step start: There was no significant interaction effect ($F(1,14) = 3.02$; $p = 0.104$; $\eta_G^2 < 0.01$); the main effect of *method* ($F(1,14) = 64.23$; $p < 0.001$; $\eta_G^2 = 0.65$) and *vision* ($F(1,14) = 9.29$; $p = 0.009$; $\eta_G^2 = 0.01$) were significant. Average COM to BOS closing velocity was lower in PULL trials than in PLAT trials; and lower in EC trials than in EO trials (Table 3.1; Figures 3.8).

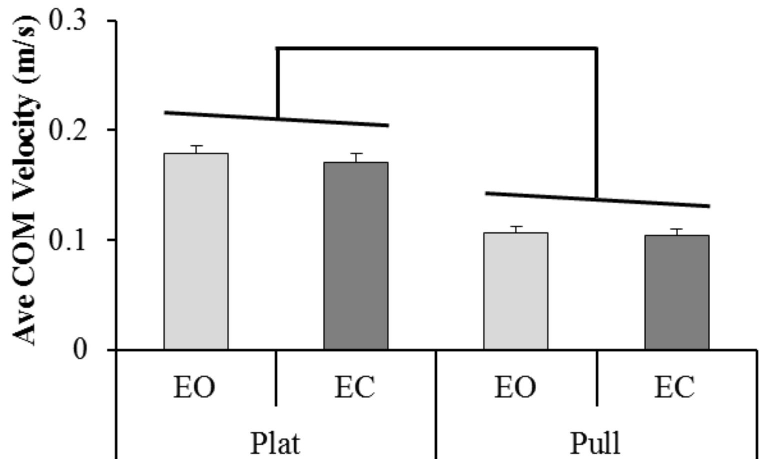

Figure 3.8. A graphical representation of average COM velocity with respect to BOS means between onset and step start events. Bracket represents significant main effect of *method*. Slanted lines represent significant main effect of *vision* and the associated direction of the effect. The error bars are SE. Average COM velocity with respect to BOS was lower in PULL trials than in PLAT trials; and lower in EC trials than in EO trials.

3.5. Discussion

It has been suggested that balance-correcting responses may be method-specific (Mansfield and Maki, 2009). To date, the research addressing this topic is sparse. The current study investigated balance recovery responses induced with PLAT and PULL perturbation methods with the addition of EO and EC sensory conditions. It was hypothesized that: 1) participants would show poorer dynamic balance control, defined as smaller MOS, with platform-translation than with the shoulder-pull method, and that 2) vision would have a larger effect on balance-correcting responses induced with shoulder-pull perturbation as opposed to platform-translation. In the current study, it was found that: 1) MOS was smaller during balance recovery following PLAT as opposed to PULL trials at both step start and at foot contact, thereby confirming the first hypothesis; 2) MOS was larger during the EC than EO vision condition at foot contact but not at step start, which confirmed the second hypothesis in part.

The effect of perturbation method on balance-correcting responses

It was found that during balance recovery, MOS was smaller during PLAT than PULL trials at step start and at foot contact, which suggests that participants were at a dynamic postural instability at step start (effect size of 0.27; large) and even greater dynamic postural instability at foot contact (effect size of 0.69; large) when the PLAT method was used as opposed to the PULL method. Since the MOS was smaller, the results may suggest that the COM was closer to the BOS and/or moving faster toward the

BOS during the PLAT trials, at both step start and foot contact. This relationship could be associated with a greater risk of loss of balance and dynamic postural instability; and thus, merits further investigation.

Larger MOS during PULL could be explained by earlier perturbation threat detection demonstrated during PULL as it was hypothesized that during PULL perturbation, participants would experience larger and faster head movement than during PLAT. Larger and faster head movement would stimulate the vestibular and visual systems alerting participants to the postural threat earlier in PULL than during PLAT trials. The head velocity and acceleration, indeed, were found to be larger during PULL trials than during PLAT trials at step start; both showed large effect size. Therefore, it is reasonable to suggest that perhaps more sensory information (visual and vestibular) about the perturbation was available to participant, and that possibly perturbation initiation was sensed earlier during PULL trials than during PLAT trials. Perhaps, participants, having more sensory information available, initiated a stepping response earlier during PULL trials. Earlier initiation of a stepping response would preclude COM from travelling farther and with larger velocity, which could produce larger MOS if operating under the assumption that step length was unchanged. Early initiation of a balance-correcting response could also explain the larger MOS at foot contact observed during PULL trials. The COM would not have travelled as far and as fast in the shorter period of time that it took participants to recognize a postural threat. Lower velocity of COM travel would also mean lower momentum. A translating mass that has lower momentum requires less force, i.e. muscle torque, to be arrested. Perhaps, larger MOS during PULL trials at step start and at foot contact can be explained with spatiotemporal measures. As previously suggested, smaller step latency at step start could explain a larger MOS during PULL. While at foot contact, larger step length and shorter step time could explain larger MOS during PULL. However, as discussed below, the differences observed in MOS means between PLAT and PULL trials were explained by the physical properties of a human body and the perturbation, and not the neuromuscular control of posture in response to perturbation.

The investigation into step latency showed that there was no difference in step latency between perturbation methods. Even though it is reasonable to suggest that more sensory information (i.e. visual and vestibular information) was available to participants during PULL trials, as the head velocity was larger than during PLAT, participants initiated step at approximately the same time (~240ms) during both perturbations, which is also similar to previously published data (McIlroy and Maki, 1996). The lack of difference between PLAT and PULL means for step time disproves the hypothesis of sensory advantage during PULL trials. Step time, therefore, does not explain difference observed between PLAT and PULL MOS means at step start. Further, the results show that step time was smaller during PLAT trials than during PULL, though the effect size was small. Step length of approximately 0.32m, which is remarkably similar to previously published data (McIlroy and Maki, 1996), was not different between PLAT and PULL trials. A shorter step time combined with equivalent step length could have produced larger MOS at foot contact during PLAT trials. Yet, the contrary was true: MOS was larger for the PULL trials.

An investigation into variables directly involved in MOS calculation, COM location and velocity with respect to the boundary of BOS, which were also likely to have the largest effect on MOS, was conducted. It was found that during PLAT trials, BOS moved closer to COM, while COM remained relatively motionless (Figure 3.9A). The opposite was true for the PULL trials (Figure 3.9B); BOS remained relatively motionless while COM moved closer to the BOS. Interestingly, COM was significantly closer to the boundary of BOS both at step start (Figure 3.6A) and at foot contact (Figure 3.6B) during PLAT trials than PULL trials, with large and moderate effect sizes, respectively. However, an investigation into COM velocity revealed no difference between PLAT and PULL conditions at step start (Figure 3.7A). It was, therefore, important to explain how COM translated significantly closer to BOS during PLAT trials, despite the lack of difference between perturbation methods in the measures of step latency and COM velocity at step start.

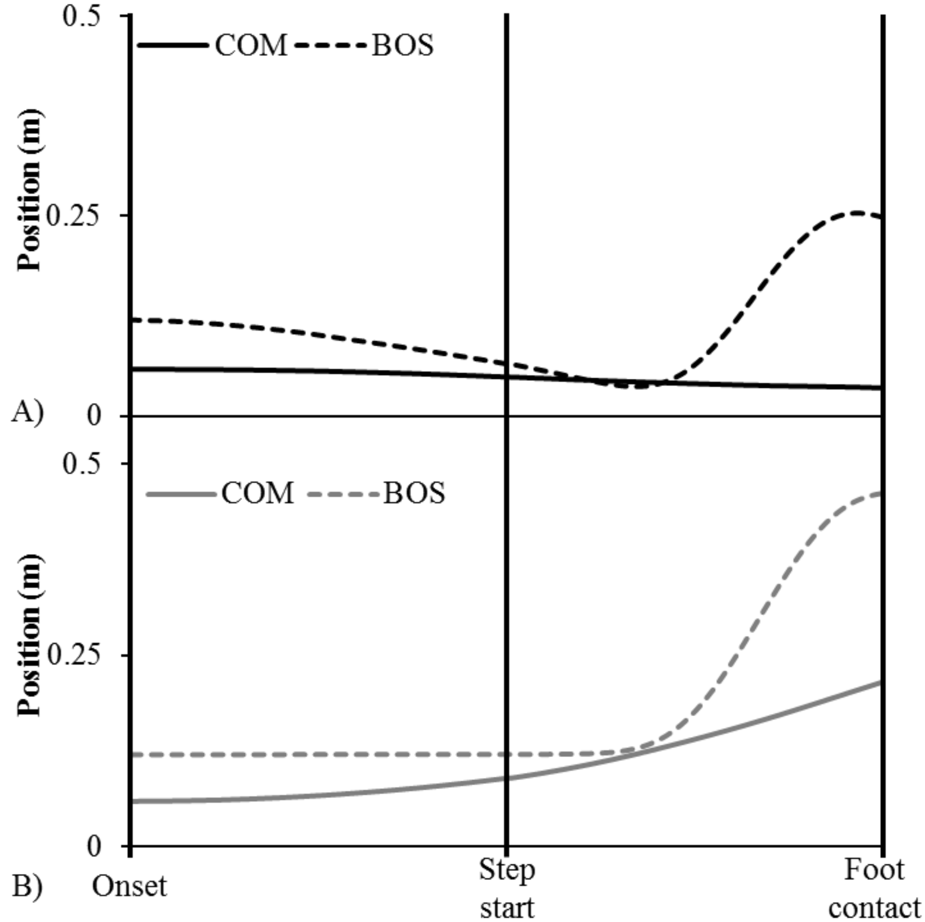

Figure 3.9. A graphical representation of averaged (n = 15) time series of COM and BOS position signals of A) PLAT and B) PULL trials. The signals were time-normalized between onset and step start, and between step start and foot contact events. COM was closer to the anterior boundary of BOS in PLAT than in PULL trials at step start and at foot contact.

The answer to the above question can be observed in the COM velocity signal profile (Figure 3.10). From Figure 3.10, it is apparent that the COM velocity profiles for PLAT and PULL between onset and step start events were qualitatively different than between step start and foot contact, where profiles appeared to be nearly identical throughout. Statistical analysis showed that average COM velocity between onset and step start was significantly larger during PLAT trials than during PULL trials, with a large effect size of 0.65 (Figure 3.8). Between the perturbation initiation (onset event) and step initiation (step start event), the platform and the participant's feet translated farther and at higher average velocity during PLAT trials than participant's upper body (HAT; head, arms, and trunk) during PULL trials, therefore placing the participant at greater dynamic postural instability at the time of step initiation. The effect observed may be explained by the distribution of weight in the human body and the inertial properties of its segments. Two-thirds of the body mass is located in HAT (Winter, 2005). The distribution of mass in HAT is also very different from that in the lower body. The HAT has greater mass distal to the hips than the lower body does, which creates larger moment of inertia in the HAT (Winter, 2005). Larger moment of inertia means higher resistance to change in an object's state of motion. Likely because HAT had a larger moment of inertia than the lower body, it offered greater resistance to perturbation stimulus than did the lower body. Thus, during the time period between perturbation onset and step start (step latency), which was not different between PLAT and PULL, the COM in PULL trials did not translate as far and as fast as it did in PLAT trials. During PLAT trials, the participants' COM displaced closer to the boundary of BOS and at a higher velocity, effectively placing them in a less favorable circumstance for balance recovery. Moreover, it was found that in 14% of the PLAT trials, in contrast to 3% of the PULL trials, participants required a second balance-correcting step which supports the argument of platform-translation perturbations being more destabilizing than cable-pull perturbations. The current findings support the notion by Mansfield and Maki (2009), of surface-translation perturbation being more destabilizing than cable-pull perturbation, and broaden the understanding of perturbation-specific balance-correcting response mechanics.

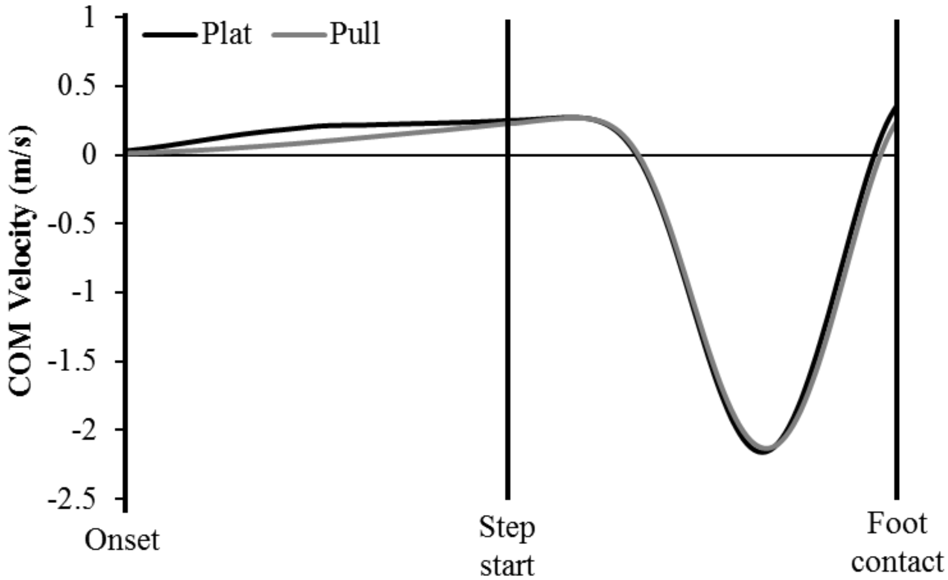

Figure 3.10. A graphical representation of averaged (n = 15) time series of COM velocity with respect to BOS position signals for both PLAT and PULL trials. The signals were time-normalized between onset and step start, and between step start and foot contact events. While COM velocity with respect to BOS was not different at step start, the average velocity between onset and step start was larger during PLAT trials than during PULL trials. COM velocity at foot contact was larger during PLAT than during PULL trials.

The effect of perturbation method and vision on balance-correcting responses

The absence of vision (EC) did not have a significant effect on MOS at step start. Contrary, at foot contact, the analysis revealed a significant interaction between method and vision, although the effect size was small (0.03). The MOS was larger during PLAT trials with EC than with EO. There was no difference between EO and EC during PULL trials. This was an unexpected finding because it was hypothesized that vision would have a greater effect on movement detection when the perturbation stimulus was applied near the head (PULL), as opposed to near the feet (PLAT), since the stimulus applied near the head would likely produce larger head velocity and acceleration, confirmed by the current findings (Figure 3.4), and therefore larger stimulation of visual sensors aiding in postural threat detection. Instead, the opposite was observed. There was no difference between vision conditions observed during PULL trials at step start or at foot contact. Horak et al. (1990) reported that due to redundancy of sensory systems in healthy participants, the lack of sensory contribution from any one of the systems was supplemented by the upregulation of sensory input from the other two systems, which likely explains the absence of a vision effect observed in PULL trials. Further investigation showed that the distance between COM and BOS was not different between the vision conditions and there was no interaction between method and vision conditions at foot contact. The findings observed in MOS measure at foot contact can be explained by the COM velocity at foot contact. At foot contact, the COM velocity was larger during PLAT trials with EO than with EC, while COM velocity during PULL trials with both EO and EC were smaller than during PLAT and not different from each other (Figure

3.7B). It is likely that PLAT was perceived as a more posturally threatening condition, and even more so with added EC condition, which could have led to a more conservative response as characterized by the differences between EO and EC in the PLAT trials.

Furthermore, spatiotemporal measures, including step latency, step time, and step length, showed significant effects of vision even though, yet again, the effect sizes were small. Participants initiated a stepping response later with EO than with EC, while their steps were faster and shorter in both PLAT and PULL trials. Interestingly, vision appeared to have a larger effect on step length in PLAT trials than in PULL trials. Step length was shorter by 16% during EC than EO in PLAT trials; while in PULL trials, step length was shorter by 4%. This is consistent with the hypothesis of a more conservative balance recovery response with EC and supports previous findings in healthy adults (Verniba and Gage, 2014). That participants showed no difference in MOS between vision conditions in PULL trials while showing nearly five-fold larger MOS in PULL than in PLAT could possibly mean that PULL perturbation was not as challenging as PLAT. In PLAT where MOS was smaller, the difference between EO and EC was significant, which is likely due to the greater challenge presented by PLAT compared to PULL perturbations. Therefore, PLAT perturbation, unlike PULL, was able to bring out the differences between sensory conditions. The absence of vision and reduced vestibular input due to lower head velocity during PLAT was perceived as a larger threat by participants, and therefore, participants showed a more conservative response during EC than EO. Still, all statistically significant effect sizes associated with the vision condition were small, and therefore, may not have been meaningful.

3.6. Conclusion

This study compared stepping balance-correcting responses induced with platform-translation and shoulder-pull perturbation methods with EO and EC sensory condition. Margin of stability measured at step initiation (step start) and at foot contact was used as the primary measure to quantify and compare balance-correcting responses. In this study, participants showed smaller margin of stability during platform-translation trials than during shoulder-pull trials at step initiation and at foot contact. The removal of vision did not have an effect on the margin of stability in shoulder-pull trials; however, the absence of vision did have an effect on margin of stability during platform-translation trials. Furthermore, participants showed more conservative balance-correcting responses with vision removed (EC), which was more pronounced for platform-translation trials. The participants showed shorter stepping latency, and faster and shorter steps with vision removed. The effect sizes for all vision condition effects were small. The difference in margin of stability between platform-translation and shoulder-pull trials was likely due to the mechanical effect of the specific type of perturbation used, rather than neuromuscular control associated with the postural perturbation method. The platform-translation perturbation appears to be more challenging and destabilizing than shoulder-pull perturbation, which is also evident from the larger number of trials where participants required a second step to fully recover balance. The current research highlights differences in margin of stability, a marker of dynamic postural stability, and underscores that caution is required when interpreting results of studies utilizing different perturbation methods.

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CHAPTER 4

Study 3 (#EMG): Postural Organization during Balance-correcting Responses Induced with Platform-translation and Shoulder-pull Perturbation Methods

4.1. Summary

Introduction: There is a disagreement in the postural control literature with respect to organization of balance-correcting responses, which may be due to the use of different perturbation methods. The current study examined neuromuscular organization of balance-correcting responses induced with common perturbation methods: platform-translation and shoulder-pull, and under two vision conditions: eyes-open and eyes-closed.

Methods: Fifteen young healthy males participated. Unexpected forward and backward platform-translation and shoulder-pull perturbations were induced with eyes-open and eyes-closed. Participants were asked to behave naturally. Only forward stepping trials were analyzed. EMG of leg, thigh, and trunk anterior and posterior muscles was recorded bilaterally. Muscle activation latencies were calculated with respect to perturbation initiation.

Results: Anterior muscles showed no difference in activation latencies (~210ms) across perturbation methods. Bilateral symmetrical distal-proximal sequential activation of posterior muscles between 70ms and 260ms was observed during platform-translation trials. During shoulder-pull trials, proximal-distal sequential activation of posterior

muscles between 70ms and 130ms was observed in stance side muscles; while latencies were not different between muscles (~80ms) in stepping side. The effect of vision was not significant across anterior or posterior muscles. Posterior muscles showed four to five times larger muscle activity in shoulder-pull than in platform-translation trials throughout the response. The absence of vision increased muscle activity by 40%, which was not different between methods.

Discussion: Though triggering of balance-correcting responses, demonstrated by dissimilar activation sequence of posterior muscles, was different between methods, the response of anterior muscles was not. The larger muscle activity during shoulder-pull suggests a more robust balance-correcting response during shoulder-pull perturbation, as opposed to platform-translation. The absence of vision showed a similar effect on balance-correcting responses between perturbation methods. Though some similarities between methods in balance-correcting responses exist, the method-specific differences in response triggering and muscle activity are clear.

4.2. Introduction

The ability to maintain upright posture following external perturbations is critical to locomotion. Since the human body consists of multiple segments with whole-body fulcrum at the ankles and the centre of mass above the ankles, it is inherently posturally unstable. Therefore, timely postural threat detection and appropriate neuromuscular coordination of postural joint moments is crucial to successful balance recovery following external postural perturbations. It has been shown that neuromuscular responses to perturbation during balance recovery are faster than during volitional movements, which suggests that the responses may be triggered on a spinal level (Allum et al., 1995; Horak and Nashner, 1986). Despite very short latency and rapid execution, the balance-correcting response is complex as it is contingent on the context and the integration and availability of sensory information (Horak and Nashner, 1986; Horak et al., 1994; Inglis et al., 1994). While it has been suggested that vision has little contribution to balance-correcting responses (Carpenter et al., 1999; Colebatch et al., 2016; Horak et al., 1990), there is a history of evidence supporting substantial contribution of both vestibular and somatosensory systems to balance-correcting responses (Allum et al., 1995; Bloem et al., 2000; Horak et al., 1990; Horak et al., 1994). Moreover, the somatosensory system has been recognized as the primary trigger of postural correction responses (Allum and Honegger, 1998; Inglis et al., 1994), and the vestibular system for modulating the magnitude of postural responses (Allum and Honegger, 1998). The somatosensory triggers are manifested by short-latency electromyographic (EMG) muscle stretch responses as short as 30ms for the upper limb and 50ms for the lower limb (Cordo and Nashner, 1982). The understanding of the organization of balance-correcting responses and muscle activation sequence during balance recovery, however, emanates from studies that utilize a variety of perturbation methods.

While there is an agreement within literature that balance-correcting responses are triggered by the somatosensory receptors, there appears to be a discord with regards to which muscle group stretch receptors (muscle spindles) act as the primary triggers for balance correction. Classic literature suggests that lower leg muscles are responsible for early triggering and directional sensitivity (Nashner, 1977). Moreover, distal-proximal muscle activation and stabilization of the joint closest to perturbation has been observed (Horak and Nashner, 1986; Nashner, 1982). Contrary, more recent studies have revealed that proximal muscles such as hip and lower trunk muscles have shown similar or shorter latencies as leg muscles (Bloem et al., 2000; Bloem et al., 2002). It has also been argued that trunk and hip muscles are primary somatosensory triggers of postural responses rather than lower-leg muscles; in addition, due to the proximity of lower trunk and hip muscles to the CNS, the postural threat is detected earlier when trunk and hip muscles are activated, which also act to trigger balance-correcting responses in lower-leg muscles (Bloem et al., 2002). It is important to note, however, that the perturbation method used by the supporters of lower-leg muscle triggers and distal-proximal activation was platform translation, while those who argued for the centrally generated triggers (trunk and hip) and proximal-distal activation used a rotating platform. While it is reasonable to suggest that, for instance, during posterior platform translation the ankles undergo flexion, which mechanically is comparable to toes-up pitch rotation where the ankle flexes as well, the expectation that the two methods, platform translation and pitch rotation, should produce similar neuromechanical responses may be flawed. For example, one reason might be that head movement is likely much smaller, if not absent, in the platform pitch rotation method, as opposed to platform translation. Consequently, the responses observed may be method-specific, which could explain the contradictory findings reported thus far.

Indeed, newly emerging evidence within postural control literature suggests that balance-correcting responses are method-specific (Mansfield and Maki, 2009). That research using a translating and rotating platform is in disagreement may be the consequence of different perturbation methods used. The purpose of the current study was to examine the similarities and differences in the neuromuscular organization of the balance-correcting responses between two of the commonly used perturbation methods: platform-translation and shoulder-pull, and under two sensory conditions: eyes-open and eyes-closed. The research questions were: 1) is there a difference between muscle activation latencies observed during balance-correcting responses induced with platform-translation and shoulder-pull methods with and without vision, and 2) is there a difference between the amount of muscle activity during balance-correcting responses induced with platform-translation and shoulder-pull methods with and without vision. It was hypothesized that: 1) participants would demonstrate longer muscle activation latencies during balance recovery induced with platform-translation than with shoulder-pull, and moreover, participants would demonstrate distal-proximal sequence of muscle activation during platform-translation and proximal-distal sequence of muscle activation during shoulder-pull, while the absence of vision would not have an effect on muscle activation latencies with either of the methods; and 2) participants would demonstrate larger muscle activity during balance recovery with platform-translation than with shoulder-pull methods, and even greater activity with vision absent.

4.3. Methods

Participants

Fifteen healthy males (age 24.3 ± 3.0 years, height 181.2 ± 5.9cm, body weight 82.0 ± 14.0kg, foot length 27.2 ± 1.0cm; mean ± SD) participated. The participants in this study (#EMG) were the same as those who participated in the #MOS study. Participants were not included if they reported a history of neurological or musculoskeletal disorders; or an injury, pain or surgery on their lower body and back in the six months prior to participation. York University research ethics board provided approval of the methods used in this study. All participants provided informed consent prior to participation.

Participant preparation, set-up, and protocol

The marker set-up was the same as that used in the #MOS study. However, in this study (#EMG) an 8-segment kinematic model used: trunk, pelvis, left and right thigh, left and right shank, and left and right foot. Following reflective marker application, the sites of EMG electrode application were shaved and swabbed with alcohol. Two surface EMG electrodes (Ambu® Blue Sensor N Electrodes, King Medical Ltd., Ontario) per shaved site with approximately 3cm spacing were applied bilaterally over each of the anterior muscles: rectus abdominis (RA), rectus femoris (RF), and tibialis anterior (TA); and each of the posterior muscles: erector spinae (ES), biceps femoris (BF), and gastrocnemius medialis (GM). The EMG electrodes were applied over the muscle belly in parallel with the muscle fibre orientation.

Participants were barefoot for the duration of the experiment. All trials were initiated with participants standing on a custom-made platform used to produce support-surface perturbations. During platform-translation trials (PLAT), the platform was affixed to the perturbation device via cables and pulleys. The platform was unexpectedly translated either forward or backward. During shoulder-pull trials (PULL), the platform was locked in place such that there was no support surface movement. Participants wore a shoulder harness which was affixed to the perturbation device via cables and pulleys. Participants were unexpectedly pulled forward or backward via the shoulder harness.

Perturbations were induced by release of an electromagnet, which allowed free weights to fall a controlled distance and thereby exerting force via the cables on the translating platform in PLAT trials or the participant's upper body in PULL trials. The intensity of the perturbation stimulus was set individually for each participant as a combination of two perturbation parameters: pull force and pull distance. The pull force was equivalent to 8.75% of the participant's body weight (BW), while the pull distance was equivalent to 105% of the participant's base of support (BOS) length. The BOS length was defined as the participant's foot length. The perturbation parameters were previously determined to consistently induce a stepping response using both PLAT and PULL paradigms (#Thresholds). Participants were instructed as follows: "Behave as naturally as possible. If you don't have to take a step, don't take a step. If you feel the need to take step to avoid falling, do take a step. Do what is natural to avoid falling." The same instructions were given to all participants in both PLAT and PULL trials.

Each participant completed 8 trials per visual condition (eyes-open: EO/eyes-closed: EC), 16 trials per direction of perturbation which elicited forward (FWD) and backward (BWD) stepping response, and 32 trials per perturbation method (PLAT/PULL) for a total of 64 trials per participant (Figure 4.1). On average, participants performed one trial per minute, and rest/recovery breaks were provided between perturbation paradigm blocks to attenuate the effects of fatigue. For the vision condition trials, participants were asked to maintain their eyes closed, in addition a blindfold was provided. The blindfold was placed over the eyes prior to each EC trial; upon completion of an EC trial, participants were allowed to remove the blindfold and open their eyes in order to reposition themselves on the platform for the following trial. The order of trials (FWD/BWD with EO/EC) within each perturbation method block was randomized. The trials which resulted in a posterior step, i.e. the trials where the platform was moved forward and the trials where participants were pulled backward via a shoulder harness, were treated as catch trials and were not used in the analysis. Participants reported that they were unable to predict the direction and timing of perturbation. The trials were blocked by perturbation paradigm. The order of perturbation paradigm blocks was counter balanced across participants.

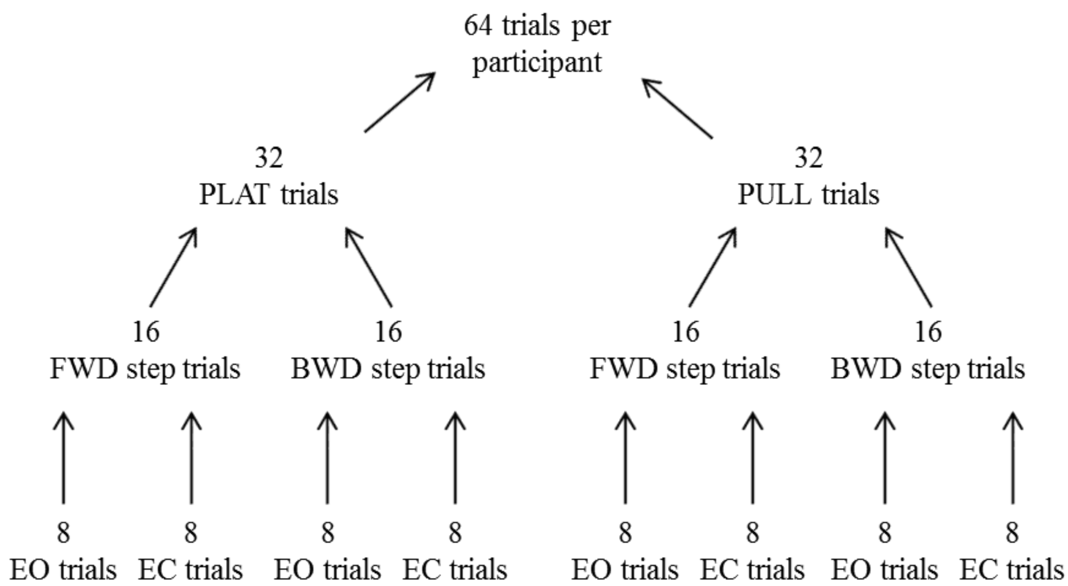

Figure 4.1. A diagram depicting the arrangement of trials and conditions for each participant. The order of PLAT and PULL blocks were counterbalanced between participants, all trials within PLAT and PULL blocks were randomized.

Data collection and processing

Marker movement was recorded at 100Hz using a 7-camera motion capture system (MX40, Vicon, Colorado). Marker position signals were filtered offline using a digital Butterworth 4^{th} order low-pass filter with an 8Hz cut-off. The cut-off frequency was determined using a residual analysis approach (Winter, 2005). Surface EMG signals were collected with TeleMyo 2400T/R G2 system (Noraxon USA Inc., Arizona): Input Impedance > 100 MegOhm, CMR > 100 dB, Unit Sampling Rate – 3000Hz, 1st order high-pass analog filter with a 10Hz cut-off. EMG signals were recorded using the motion capture system software (Nexus v1.6, Vicon, Colorado) at a 1000Hz sampling frequency. The EMG signals were then high-pass filtered using a digital 4th order Butterworth dual-pass filter with a 40Hz cut-off frequency to attenuate contamination of the EMG signal with a heart depolarization signal (Drake and Callaghan, 2006). The EMG signals were then full-wave rectified. Finally, the EMG signals were low-pass filtered using a digital 4th order Butterworth dual-pass filter with a 100Hz cut-off frequency (Gage et al., 2007). All digital processing was conducted offline using Visual3D software (v4.84.0, C-Motion Inc., Ontario); Visual3D software was successfully utilized in previously published work (Verniba et al., 2015; Wang et al., 2013).

Measures of interest

All measures of interest were calculated using Visual3D software. Primary dependent measures, muscle activation latency and muscle activity, were computed. Muscle activation latency was defined as the time between perturbation initiation and the onset of muscle activity. The onset of muscle activity was determined by selecting the first point at which the EMG profile in each trial exceeded a level of three standard deviations above the baseline EMG signal. The EMG baseline was calculated over 100ms prior to the initiation of perturbation. The initiation of the perturbation was defined as posterior heel marker movement in PLAT trials and anterior C7 marker movement in PULL trials, induced by movement of the platform on which the foot was placed or the harness which was fixed securely to the trunk (#MOS). Further, to satisfy the criteria for the onset of muscle activity, the EMG signal had to remain above three-standard-deviation threshold longer than 50ms. All EMG signals were visually inspected to ensure the accuracy of muscle activity onset identification.

Muscle activity was quantified across each of the following six epochs: 1 (-100–0 ms; pre-perturbation background activity), 2 (30–80 ms; short-latency stretch response), 3 (80–120 ms; medium-latency stretch response), 4 (120–220 ms; primary balance-correcting response), 5 (240–340 ms; secondary balance-correcting response), and 6 (350–700 ms; stabilizing phase); which are consistent with previously published work (Gage et al., 2007). Muscle activity was calculated by integrating the EMG signal (iEMG) using the trapezoidal rule. To compare muscle activity between epochs, the average EMG amplitude was calculated across each epoch by normalizing iEMG to the length of each epoch.

Spatiotemporal kinematic measures: step initiation latency, step time, time from perturbation onset to foot contact, and step length were calculated. Trunk, hip, knee, and ankle flexion angle signals were calculated in order to understand the movement that was occurring during the responses. All joint angle signals were referenced to pre-perturbation standing posture, i.e. the average of the joint angle signals that was calculated over 500ms for each trial prior to the onset of the perturbation and subtracted from the corresponding joint angle signal.

Statistical analysis

All statistical analyses were conducted using JMP (v8.0, SAS Institute, North Carolina). A two-factor (*method* [PLAT/PULL] x *vision* [EO/EC]) mixed effects (*participant* – random effect, *method* and *vision* – fixed effect) repeated measures analysis of variance (rmANOVA) was used to test for differences in the measures of interest: muscle activation latency, muscle activity, and spatiotemporal measures (step initiation latency, step time, time from perturbation onset to foot contact, and step length). Muscle activation latencies were analyzed separately for each muscle within each side (stance side: NSS; stepping side: SS). Muscle activity was analyzed separately for each epoch within each muscle within each side. As described in the results, below, there were no significant interaction effects and no main effect of *vision* found in the measure of muscle activation latency, thus muscle activation latency data were collapsed across *vision* conditions.

The collapsing of muscle activation latency data across the *vision* condition allowed for the investigation of the sequence of muscle activation within anterior and posterior NSS and SS muscles using a two-factor *method* x *muscle* rmANOVA model. Thus, muscle activation sequence was analysed using a two-factor (*method* [PLAT/PULL] x *muscle* [RA/RF/TA]) for the anterior muscles and two-factor (*method* [PLAT/PULL] x *muscle* [ES/BF/GM]) for the posterior muscles mixed effects (*participant* – random effect, *method* and *muscle* – fixed effect) rmANOVA that was conducted separately for NSS and SS muscles.

Preliminary analysis revealed that muscle activation latencies and muscle activity data were not normally distributed, so these data were log-transformed prior to conducting the ANOVA. Significant interactions and main effects were analyzed with a Tukey HSD post hoc test. Alpha values were adjusted to correct for multiple comparisons using the Holm-Bonferroni sequentially rejective procedure (Holm, 1979).

The Holm-Bonferroni sequentially rejective procedure is an approach that controls the probability of detecting one or more Type I errors by adjusting the rejection criteria of each of the individual hypotheses or comparisons. First, all *p*-values within tested hypothesis are arranged from smallest to largest. Then alpha is calculated for each respective value. The first (smallest) alpha in the sequence is calculated by dividing the typical alpha, which in the current study is 0.05, by the total number of comparisons. The rest of the alpha values are calculated by dividing 0.05 by the number of remaining comparisons within the sequence. As such, with every comparison made the alpha become less conservative as the number of remaining comparisons is reduced, until the last alpha in the sequence is equal to 0.05. Lastly, starting with the first (smallest) *p*-value within the sequence, each *p*-value is compared against the respective adjusted alpha; those *p*-values that are smaller than the respective adjusted alpha are considered significant.

4.4. Results

Muscle activation latency: method x vision model

The results of the rmANOVA *method* x *vision* model revealed that following adjustment for multiple comparisons, none of the interactions or main effects of *vision* were significant. However, the main effect of *method* was found to be significant for both NSS ES and SS ES muscles as well as both NSS and SS BF muscles (Table 4.1). ES and BF muscles showed significantly shorter muscle activation latencies during PULL trials than during PLAT trials.

Table 4.1. Muscle activation latency (*method* x *vision*) results with multiple comparisons adjustment.

	Interaction				***Method***				***Vision***			
Alpha	**Side**	**Muscle**	**F**	***p*-value**	**Side**	**Muscle**	**F**	***p*-value**	**Side**	**Muscle**	**F**	***p*-value**
0.004	SS	RA	3.89	0.068	**NSS**	**ES**	**121.64**	**<0.001**	SS	TA	6.37	0.024
0.005	NSS	ES	2.09	0.162	**SS**	**ES**	**49.65**	**<0.001**	NSS	TA	4.30	0.057
0.005	SS	ES	2.09	0.165	**SS**	**BF**	**37.94**	**<0.001**	SS	RF	2.96	0.107
0.006	NSS	BF	1.58	0.224	**NSS**	**BF**	**28.05**	**<0.001**	SS	RA	2.60	0.128
0.006	SS	BF	1.57	0.228	SS	RA	10.62	0.006	SS	ES	0.50	0.487
0.007	NSS	RA	1.33	0.266	NSS	TA	7.27	0.017	NSS	ES	0.35	0.563
0.008	NSS	RF	1.13	0.300	NSS	RA	6.90	0.019	NSS	RA	0.29	0.597
0.010	NSS	GM	0.82	0.377	SS	TA	5.02	0.042	NSS	BF	0.22	0.644
0.013	SS	RF	0.25	0.628	SS	RF	4.12	0.062	SS	BF	0.19	0.672
0.017	SS	TA	0.06	0.816	NSS	GM	2.46	0.138	NSS	RF	0.04	0.849
0.025	NSS	TA	<0.01	0.962	SS	GM	0.47	0.502	NSS	GM	<0.01	0.973
0.050	SS	GM	<0.01	0.999	NSS	RF	0.08	0.777	SS	GM	<0.01	0.985

Note: The degrees of freedom were 1 and 14 for all interactions and main effects. The *p*-values for interactions and main effects were sorted from smallest to largest. The *p*-values which were smaller than the alpha for the associated position in sequence were considered statistically significant; significant effects are bolded.

Muscle activation latency (the order of muscle activation): method x muscle model

The results of the rmANOVA *method* x *muscle* model revealed that following adjustment for multiple comparisons, both anterior and posterior NSS and SS muscle *method* x *muscle* interactions were significant. The main effects of *method* were significant for posterior NSS and SS muscle comparisons. The main effect of *muscle* was significant for posterior SS muscles only (Table 4.2).

Table 4.2. Muscle activation latency (*method* x *muscle*) results with multiple comparisons adjustment.

	Interaction				***Method***				***Muscle***			
Alpha	**Side**	**Muscle**	**F**	***p*-value**	**Side**	**Muscle**	**F**	***p*-value**	**Side**	**Muscle**	**F**	***p*-value**
0.013	**NSS**	**Post**	**49.04**	**<0.001**	**SS**	**Post**	**59.39**	**<0.001**	**SS**	**Post**	**17.80**	**<0.001**
0.017	**SS**	**Post**	**26.44**	**<0.001**	**NSS**	**Post**	**49.88**	**<0.001**	SS	Ant	3.03	0.064
0.025	**SS**	**Ant**	**11.91**	**<0.001**	SS	Ant	0.43	0.524	NSS	Post	2.05	0.147
0.050	**NSS**	**Ant**	**5.95**	**0.007**	NSS	Ant	<0.01	0.946	NSS	Ant	0.74	0.484

Note: The degrees of freedom were 2 and 14 for interactions and main effect of *muscle;* for main effect of *method* the degrees of freedom were 1 and 14. The *p*-values for interactions and main effects were sorted from smallest to largest. The *p*-values which were smaller than the alpha for the associated position in sequence were considered statistically significant; significant interactions and main effects are bolded.

Anterior NSS muscles: During PLAT trials, RA, RF, and TA activation latencies were not different; during PULL trials, RA showed longer latencies than TA, both RA and TA were not different from RF (Figure 4.2). Anterior SS muscles: During PLAT trials, RA, RF, and TA activation latencies were not different; during PULL trials, RF showed longer latencies than TA, both RF and TA were not different from RA (Figure 4.2).

Posterior NSS muscles: During PLAT trials, muscle activation latencies increased from GM to BF to ES; the order of the onset of muscle activity was reversed in the PULL trials where muscle activation latencies increased from ES to BF to GM (Figure 4.2). Posterior SS muscles: A similar sequence of activation to that observed for posterior NSS muscles during PLAT trials was observed, where muscle activation latencies increased from GM to BF to ES; during PULL trials, however, there was no difference in muscle activation latencies observed (Figure 4.2).

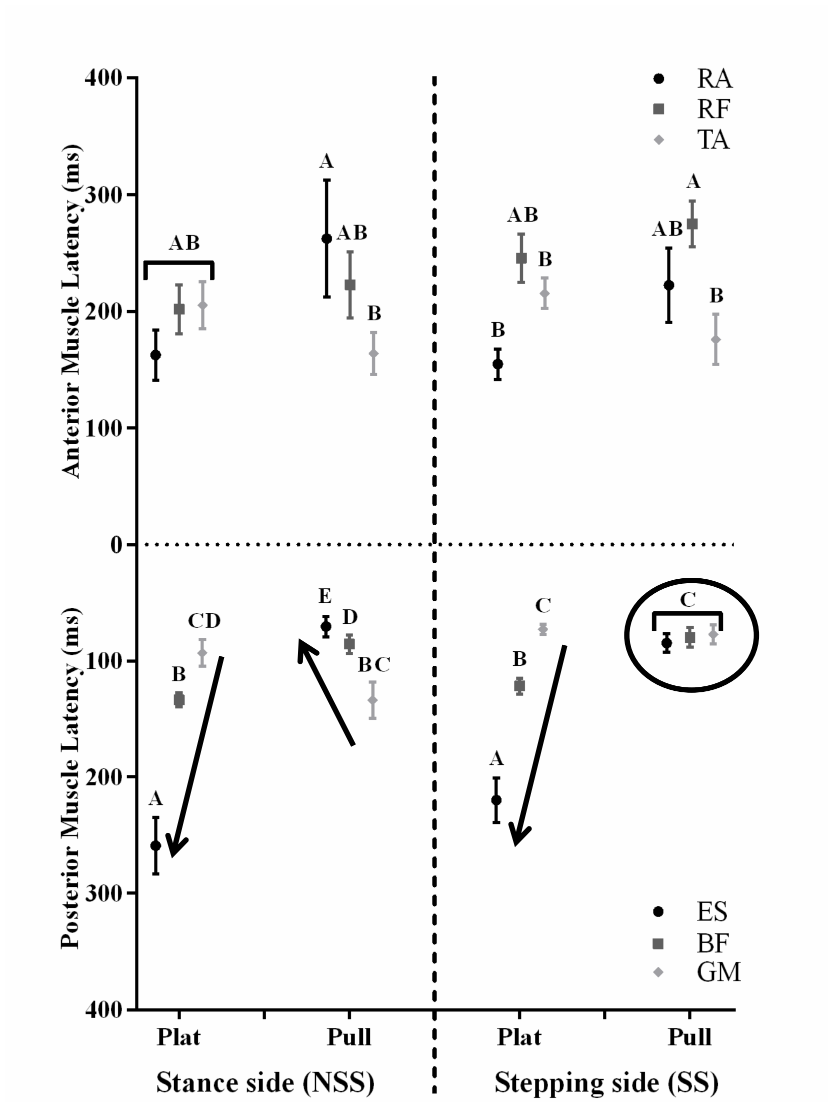

Figure 4.2. A graphical representation of muscle activation latency means for anterior (RA, RF, and TA) and posterior (ES, BF, and GM) muscles, stance (NSS) and stepping sides (SS). The error bars are SE. Letters represent significant interaction between *method* and *muscle*; levels not connected by the same letter are significantly different. Anterior NSS muscles showed no difference in activation latencies during PLAT trials; during PULL trials, RA showed longer latencies than TA, both RA and TA were not different from RF. Anterior SS muscles showed no difference during PLAT trials; during PULL trials, RF showed longer latencies than TA, both RF and TA were not different from RA. Posterior NSS muscles showed increase in latencies from GM to BF to ES (downward arrow) during PLAT trials; during PULL trials, latencies increased from ES to BF to GM (upward arrow). Posterior SS muscles showed increase in latencies from GM to BF to ES (downward arrow) during PLAT trials; during PULL trials, latencies were not different between the muscles (circle).

Muscle activity

The NSS muscle activity: The results of the rmANOVA *method* x *vision* model revealed that following adjustment for multiple comparisons, none of the interactions were significant. The main effect of *method* was found to be significant for TA during the 1^{st} epoch; ES and BF during the 2^{nd} epoch; RA, ES, BF, and GM during the 3^{rd} epoch; and BF during the 4^{th} and 5^{th} epochs. During the 1^{st} epoch, TA activity was larger in PLAT trials than in PULL. During the 2^{nd} epoch, ES and BF activity was larger in PULL than in PLAT trials. During the 3^{rd} epoch, RA and GM activity was larger in PLAT trials, while ES and BF activity was larger in PULL than in PLAT trials. During the 4^{th} and 5^{th} epochs, BF activity was larger in PULL than in PLAT trials. The main effect of *vision* was significant for RF during the 2^{nd} epoch; BF and GM during the 4^{th} epoch; and RA, RF, and ES during the 5^{th} epoch. The post hoc analyses revealed that muscle activity was larger during EC than during EO trials across all significant main effects of *vision* (Table 4.3; Figure 4.3).

Table 4.3. Muscle activity statistical results for Stance Side (NSS) with multiple comparisons adjustment.

	Interaction				*Method*				*Vision*			
Alpha	**Epoch**	**Muscle**	**F**	***p*-value**	**Epoch**	**Muscle**	**F**	***p*-value**	**Epoch**	**Muscle**	**F**	***p*-value**
0.0014	5	RF	12.70	0.0031	**3**	**ES**	**109.2**	**<.0001**	**4**	**GM**	**55.60**	**<.0001**
0.0014	4	TA	8.99	0.0095	**2**	**ES**	**89.30**	**<.0001**	**5**	**RF**	**41.48**	**<.0001**
0.0015	4	RF	7.42	0.0164	**3**	**BF**	**46.84**	**<.0001**	**4**	**BF**	**29.72**	**<.0001**
0.0015	3	ES	6.42	0.0176	**2**	**BF**	**36.14**	**<.0001**	**5**	**RA**	**22.13**	**0.0003**
0.0016	5	RA	5.86	0.0296	**5**	**BF**	**26.91**	**<.0001**	**5**	**ES**	**14.86**	**0.0012**
0.0016	1	ES	3.95	0.0635	**4**	**BF**	**22.65**	**0.0003**	**2**	**RF**	**15.38**	**0.0015**
0.0017	6	RA	3.98	0.0658	**3**	**GM**	**19.89**	**0.0005**	6	TA	13.73	0.0023
0.0017	3	BF	3.52	0.0780	**3**	**RA**	**19.61**	**0.0006**	4	RF	13.16	0.0027
0.0018	4	ES	2.66	0.1208	**1**	**TA**	**16.60**	**0.0011**	2	TA	11.58	0.0042
0.0019	4	RA	2.71	0.1218	1	GM	10.66	0.0055	6	RF	10.37	0.0061
0.0019	5	ES	2.40	0.1397	4	ES	10.16	0.0062	3	RF	9.30	0.0086
0.0020	1	RF	2.32	0.1501	6	RF	10.26	0.0064	5	TA	8.70	0.0105
0.0021	5	TA	2.24	0.1566	4	RA	10.10	0.0067	4	TA	8.26	0.0122
0.0022	5	BF	2.11	0.1654	1	BF	7.86	0.0129	6	RA	7.21	0.0177
0.0023	6	BF	1.71	0.2079	5	TA	7.22	0.0177	1	BF	6.24	0.0217
0.0024	2	BF	1.06	0.3178	6	RA	7.21	0.0178	1	RF	6.54	0.0227
0.0025	2	RF	1.00	0.3346	4	GM	7.17	0.0178	2	BF	5.40	0.0319
0.0026	5	GM	0.92	0.3521	6	TA	5.84	0.0299	3	TA	5.41	0.0354
0.0028	1	BF	0.85	0.3675	5	RF	5.22	0.0384	4	RA	5.11	0.0402
0.0029	1	RA	0.85	0.3727	4	RF	4.33	0.0563	2	RA	4.88	0.0443
0.0031	6	ES	0.78	0.3892	5	RA	4.14	0.0613	1	TA	4.71	0.0475
0.0033	4	BF	0.72	0.4079	2	TA	3.35	0.0884	6	GM	3.29	0.0893
0.0036	6	TA	0.73	0.4081	5	ES	2.11	0.1676	3	GM	2.97	0.1045
0.0038	1	TA	0.65	0.4336	1	RA	1.28	0.2768	1	GM	2.56	0.1304
0.0042	3	TA	0.61	0.4470	6	GM	1.20	0.2911	6	ES	2.19	0.1569
0.0045	4	GM	0.55	0.4691	3	TA	1.10	0.3121	2	ES	1.98	0.1754
0.0050	1	GM	0.44	0.5189	2	RA	0.89	0.3607	2	GM	1.88	0.1911
0.0056	6	RF	0.40	0.5376	1	RF	0.84	0.3738	1	ES	1.73	0.2040
0.0063	2	RA	0.39	0.5414	2	RF	0.63	0.4416	1	RA	1.29	0.2744
0.0071	3	RA	0.19	0.6723	2	GM	0.38	0.5455	4	ES	0.88	0.3628
0.0083	2	GM	0.16	0.6920	4	TA	0.11	0.7416	3	BF	0.54	0.4742
0.0100	6	GM	0.11	0.7464	6	ES	0.11	0.7499	6	BF	0.45	0.5145
0.0125	3	RF	0.09	0.7654	3	RF	0.09	0.7706	5	GM	0.32	0.5786
0.0167	3	GM	0.03	0.8662	5	GM	0.09	0.7741	3	ES	0.31	0.5825
0.0250	2	ES	0.01	0.9093	6	BF	0.07	0.7918	5	BF	0.23	0.6412
0.0500	2	TA	<.01	0.9781	1	ES	<.01	0.9455	3	RA	0.21	0.6510

Note: The degrees of freedom were 1 and 14 for all interactions and main effects. The *p*-values for interactions and main effects were sorted from smallest to largest. The *p*-values which were smaller than the alpha for the associated position in sequence were considered statistically significant; significant effects are bolded.

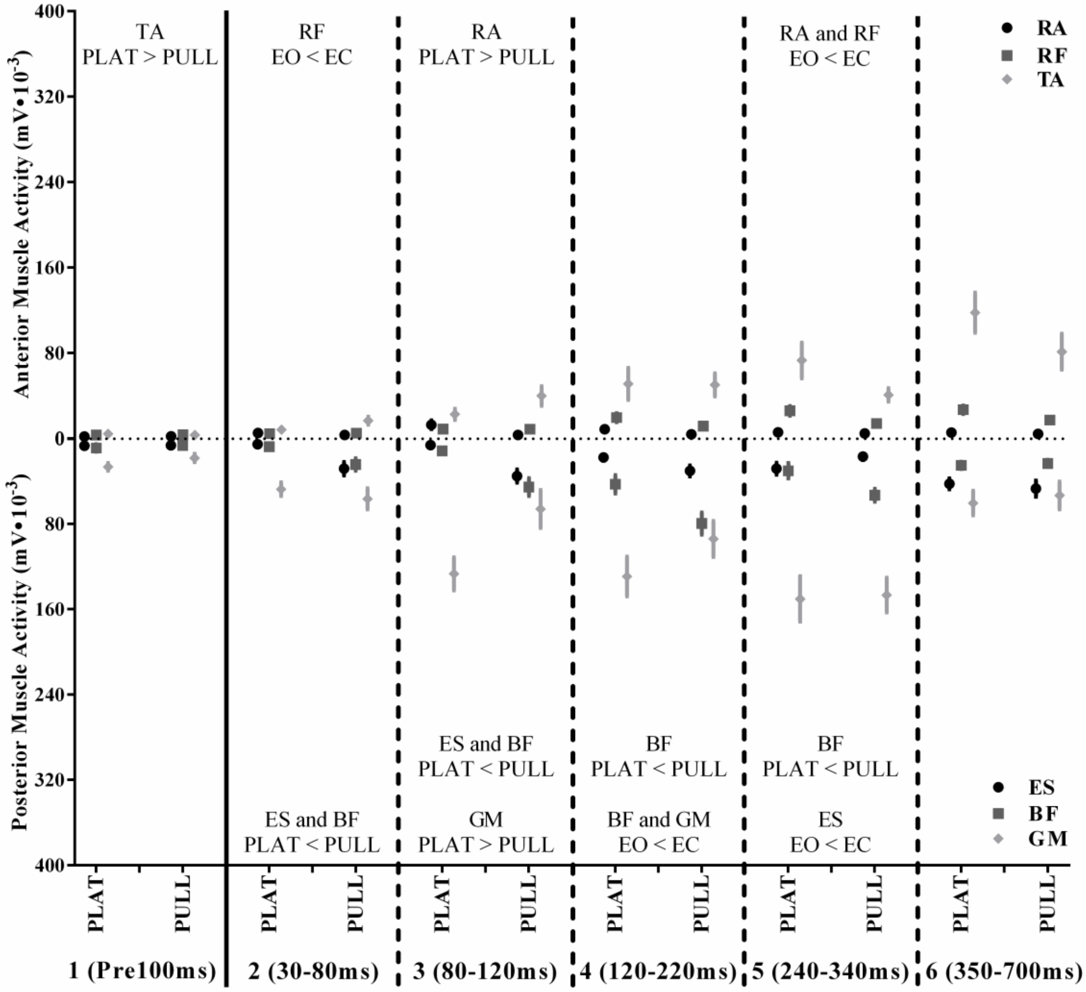

Figure 4.3. A graphical representation of Stance Side (NSS) anterior and posterior muscle activity means. The error bars are SE. During the 1st epoch, TA activity was larger during PLAT trials than during PULL. During the 2nd epoch, ES and BF activity was larger during PULL than during PLAT trials. During the 3rd epoch, RA and GM activity was larger during PLAT trials, while ES and BF activity was larger during PULL than PLAT trials. During the 4th and 5th epochs, BF activity was larger during PULL than during PLAT trials. The post hoc analyses revealed that muscle activity was larger during EC than during EO trials across all significant main effects of *vision*.

The SS muscle activity: The analysis revealed that following adjustment for multiple comparisons, there was a significant interaction for ES during the 4^{th} epoch. The ES activity during the 4^{th} epoch was smaller during EO than EC in PLAT, which was smaller than both EO and EC activity during PULL, and not different from each other. The main effect of *method* was significant for GM during the 1^{st} epoch; and ES and BF during the 2^{nd} and 3^{rd} epochs. During the 1^{st} epoch, GM activity was larger in PLAT than in PULL trials. During the 2^{nd} and 3^{rd} epochs, both ES and BF activity was larger in PULL than in PLAT trials. The main effect of *vision* was significant for RF during the 5^{th} epoch; RF showed larger muscle activity during EC than during EO (Table 4.4; Figure 4.4).

Table 4.4. Muscle activity statistical results for Stepping Side (SS) with multiple comparisons adjustment.

	Interaction				***Method***				***Vision***			
Alpha	**Epoch**	**Muscle**	**F**	***p*-value**	**Epoch**	**Muscle**	**F**	***p*-value**	**Epoch**	**Muscle**	**F**	***p*-value**
0.0014	**4**	**ES**	**16.64**	**0.0011**	**3**	**ES**	**65.22**	**<.0001**	**5**	**RF**	**17.77**	**0.0009**
0.0014	5	RF	15.40	0.0015	**2**	**ES**	**51.29**	**<.0001**	4	TA	15.05	0.0017
0.0015	3	BF	8.53	0.0095	**3**	**BF**	**38.99**	**<.0001**	2	GM	13.98	0.0022
0.0015	5	ES	8.71	0.0105	**2**	**BF**	**35.60**	**<.0001**	5	TA	12.16	0.0036
0.0016	5	TA	6.47	0.0233	**1**	**GM**	**33.83**	**<.0001**	1	TA	12.05	0.0037
0.0016	1	BF	5.45	0.0328	6	TA	13.54	0.0025	6	BF	8.87	0.0083
0.0017	3	TA	4.57	0.0507	3	RA	11.67	0.0042	6	RA	8.57	0.0109
0.0017	4	TA	4.02	0.0646	4	RA	8.66	0.0107	4	RA	8.57	0.0110
0.0018	6	RF	3.90	0.0681	2	TA	8.63	0.0108	4	RF	8.00	0.0134
0.0019	2	ES	3.86	0.0694	5	RA	7.79	0.0144	2	TA	7.46	0.0161
0.0019	3	ES	3.26	0.0925	4	ES	7.09	0.0186	4	ES	6.97	0.0194
0.0020	4	RA	2.47	0.1387	2	GM	6.54	0.0228	3	RF	6.49	0.0232
0.0021	2	BF	2.31	0.1437	3	TA	6.13	0.0267	5	ES	5.53	0.0339
0.0022	3	GM	2.35	0.1475	1	BF	4.43	0.0532	5	RA	4.71	0.0477
0.0023	1	GM	1.78	0.2032	6	GM	4.40	0.0546	2	RF	4.39	0.0548
0.0024	4	BF	1.71	0.2086	6	RA	3.58	0.0792	1	RF	3.69	0.0755
0.0025	3	RF	1.36	0.2631	3	GM	3.50	0.0822	6	GM	3.45	0.0842
0.0026	6	ES	1.29	0.2754	5	TA	3.19	0.0956	5	BF	3.10	0.0983
0.0028	4	GM	0.94	0.3498	4	RF	2.43	0.1412	3	GM	2.58	0.1303
0.0029	1	RF	0.82	0.3818	1	TA	2.37	0.1459	6	ES	2.26	0.1548
0.0031	1	ES	0.46	0.5095	5	BF	2.00	0.1788	4	BF	2.17	0.1605
0.0033	6	BF	0.36	0.5559	5	GM	1.69	0.2149	1	GM	2.10	0.1688
0.0036	5	RA	0.34	0.5699	6	RF	1.28	0.2763	3	TA	1.57	0.2308
0.0038	6	RA	0.33	0.5730	1	RA	1.11	0.3103	3	RA	1.39	0.2585
0.0042	2	RF	0.33	0.5762	6	ES	0.71	0.4127	5	GM	1.05	0.3233
0.0045	1	TA	0.17	0.6842	2	RF	0.64	0.4387	2	BF	0.96	0.3426
0.0050	5	GM	0.14	0.7094	2	RA	0.46	0.5093	1	ES	0.96	0.3450
0.0056	6	TA	0.08	0.7782	1	ES	0.45	0.5115	2	RA	0.94	0.3495
0.0063	3	RA	0.06	0.8097	4	BF	0.38	0.5455	3	BF	0.90	0.3565
0.0071	1	RA	0.05	0.8193	5	RF	0.30	0.5945	6	TA	0.86	0.3686
0.0083	2	GM	0.03	0.8706	5	ES	0.11	0.7489	2	ES	0.86	0.3686
0.0100	5	BF	0.02	0.8902	3	RF	0.07	0.7981	1	BF	0.55	0.4717
0.0125	2	TA	0.01	0.9171	1	RF	0.02	0.8836	4	GM	0.39	0.5424
0.0167	4	RF	0.01	0.9262	4	TA	<.01	0.9514	3	ES	0.34	0.5664
0.0250	6	GM	<.01	0.9645	4	GM	<.01	0.9687	1	RA	0.13	0.7207
0.0500	2	RA	<.01	0.9731	6	BF	<.01	0.9977	6	RF	0.03	0.8695

Note: The degrees of freedom were 1 and 14 for all interactions and main effects. The *p*-values for interactions and main effects were sorted from smallest to largest. The *p*-values which were smaller than the alpha for the associated position in sequence were considered statistically significant; significant interactons and main effects are bolded.

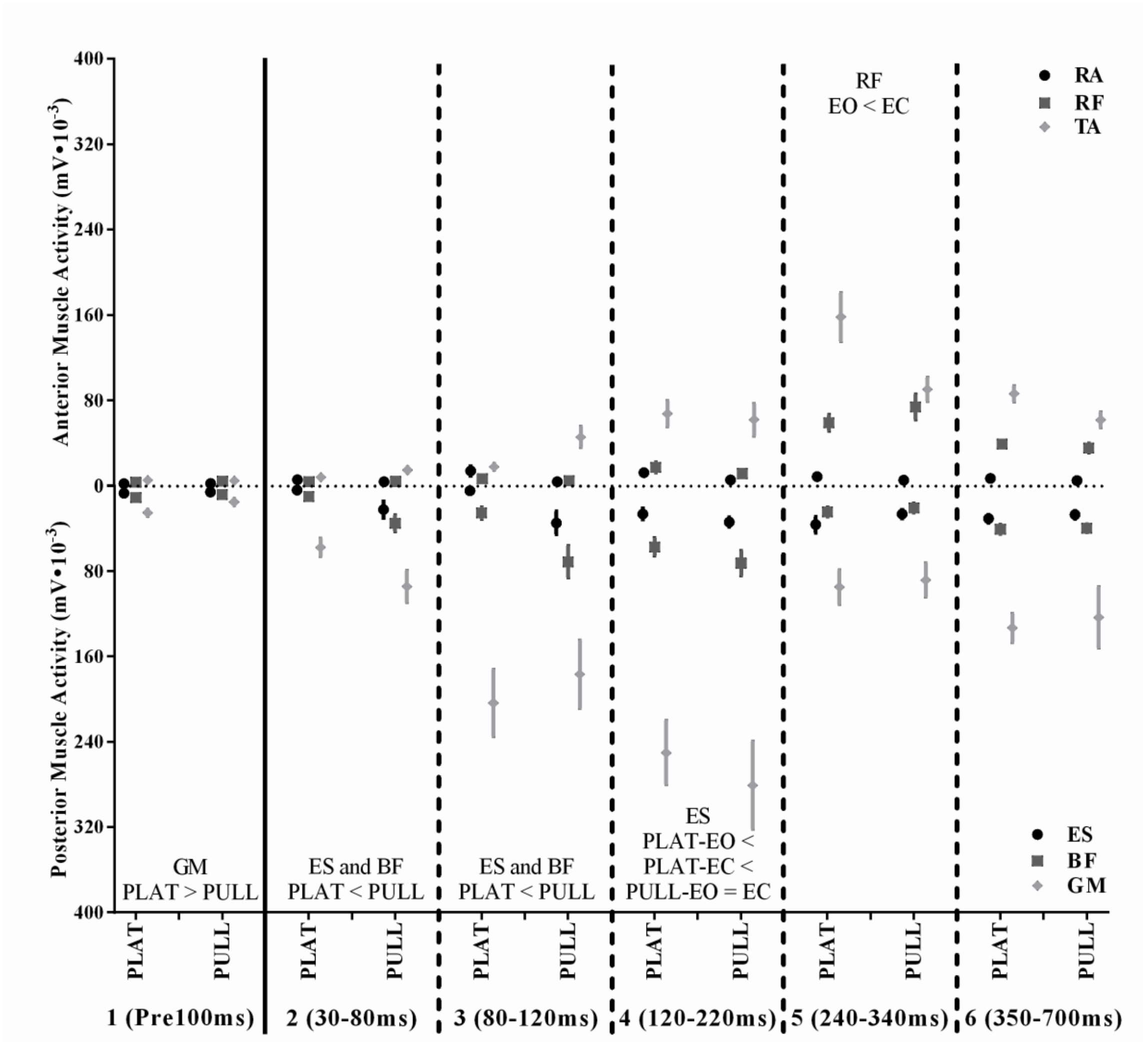

Figure 4.4. A graphical representation of Stepping Side (SS) anterior and posterior muscle activity means. The error bars are SE. During the 1st epoch, GM activity was larger during PLAT than during PULL trials. During 2nd and 3rd epochs, both ES and BF activity was larger during PULL trials than during PLAT trials. During the 4th epoch, the ES activity was smaller in EC than EO trials in PLAT, which was smaller than both EO and EC activity during PULL, and not different from each other. RF showed larger muscle activity in EC than EO during the 5th epoch.

Spatiotemporal and kinematic measures

Step initiation latency: There was no significant interaction ($F(1,14) = 0.54$; $p = 0.474$) or main effect of *method* ($F(1,14) = 0.01$; $p = 0.917$) observed; the main effect of *vision* ($F(1,14) = 11.64$; $p = 0.004$) was significant. Step initiation latency was lower in EC trials than in EO trials (Figure 4.5A).

Step time: There was a significant interaction effect between *method* and *vision* ($F(1,14) = 8.38$; $p = 0.012$); the main effect of *method* ($F(1,14) = 5.68$; $p = 0.032$) and *vision* ($F(1,14) = 26.9$; $p = 0.001$) were also significant. Step time was lowest in PLAT-EC trials than in PLAT-EO, PULL-EO, and PULL-EC trials, which were not different from each other (Figure 4.5A).

Time from perturbation onset to foot contact: There was a significant interaction effect between *method* and *vision* ($F(1,14) = 8.35$; $p = 0.011$) and main effect of *vision* ($F(1,14) = 21.7$; $p < 0.001$) observed; the main effect of *method* ($F(1,14) = 0.07$; $p = 0.792$) was not significant. Time from perturbation onset to foot contact was lowest in PLAT-EC trials than in PLAT-EO, PULL-EO, and PULL-EC trials, which were not different from each other (Figure 4.5A).

Step length: There was a significant interaction effect between *method* and *vision* ($F(1,14) = 13.62$; $p = 0.002$) and the main effect of *vision* ($F(1,14) = 37.07$; $p < 0.001$) was significant; the main effect of *method* ($F(1,14) = 0.20$; $p = 0.665$) was not significant. Step length was lower in PLAT-EC trials than in PULL-EC trials, which was lower than in both PLAT-EO and PULL-EO trials, which were not different from each other (Figure 4.5B).

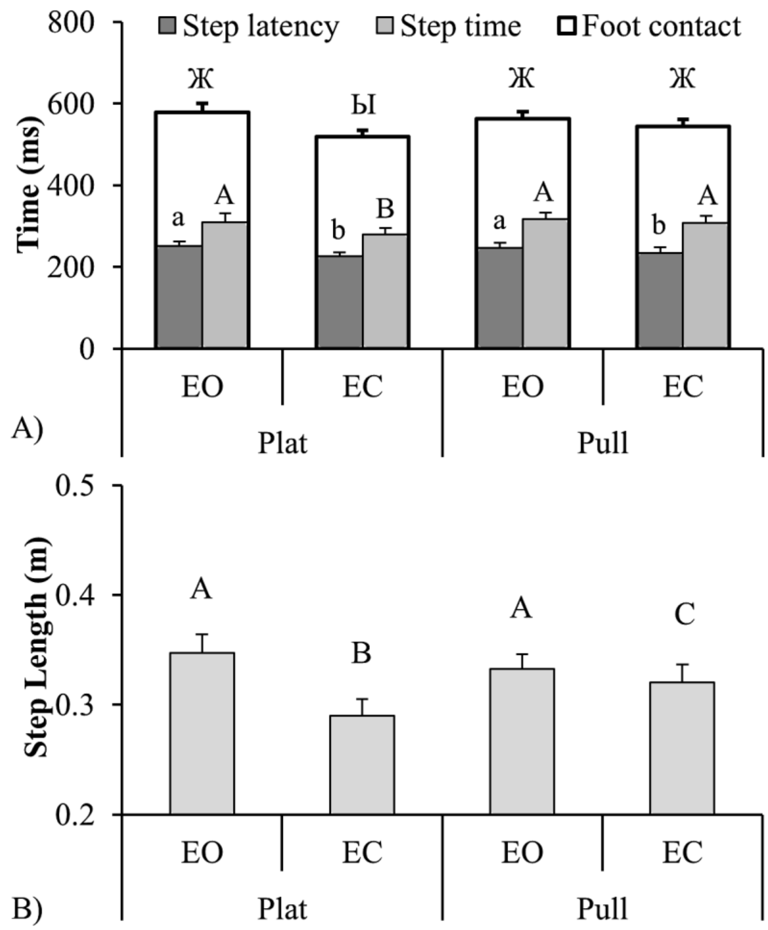

Figure 4.5. A graphical representation of A) step initiation latency, step time, and time from perturbation initiation to foot contact; and B) step length means. The error bars are SE. Levels not connected by the same letter or symbol are significantly different. *NOTE:* The effects in Figure A are presented for each measure separately. A) Step initiation latency was shorter in EC trials than in EO trials. Step time and time from perturbation initiation were shorter during PLAT-EC trials than during PLAT-EO, PULL-EO, and PULL-EC which were not different from each other. B) Step length was larger in PULL-EC trials than in PLAT-EC trials, but lower than in both PLAT-EO and PULL-EO trials, which were not different from each other.

Figure 4.6 depicts representative kinematic data time series based on two participants. The thin solid grey line represents step initiation, while the thick dashed grey line represents foot contact following the stepping balance-correcting response.

4.5. Discussion

The purpose of the current study was to examine the similarities and differences in the organization of balance-correcting responses between two commonly used perturbation methods: platform-translation and shoulder-pull, and under two sensory conditions: eyes-open and eyes-closed. It was hypothesized that: 1) participants would demonstrate longer muscle activation latencies during balance recovery induced with platform-translation than with shoulder-pull, and moreover, participants would demonstrate distal-proximal sequence of muscle activation during platform-translation and proximal-distal sequence of muscle activation during shoulder-pull, while the absence of vision would not have an effect on muscle activation latencies with either of the methods; and 2) participants would demonstrate larger muscle activity during balance recovery with platform-translation than with shoulder-pull methods, and even greater activity with vision absent.

The similarities in postural organization

Anterior muscle activation latencies and muscle activity were not different across perturbation methods; nor were most of the spatiotemporal measures. Anterior muscles became active during the 4^{th} and 5^{th} epochs, between 160ms and 275ms. However, contrary to the current hypothesis and despite the use of different perturbation methods, anterior muscles showed no differences in activation latencies between methods during balance-correcting response for either NSS or SS muscles (Figure 4.2). Though there was a statistical difference observed between NSS RA and NSS TA activation latencies in PULL trials, the differences in the latency means were likely unremarkable due to large variability. There was no effect of vision in any of the anterior or posterior muscle activation latencies, which supports previous findings (Carpenter et al., 1999; Colebatch et al., 2016; Horak et al., 1990).

The majority of anterior NSS and all anterior SS muscles showed no difference in muscle activity between PLAT and PULL throughout the balance-correcting response. Among posterior muscles, GM largely showed no difference in muscle activity between methods. It is likely that anterior muscles were not taxed during forward stepping movements, since the participant's body was forced forward by the perturbation. As discussed further, unlike anterior muscles, the posterior muscles needed to be modulated in order to arrest movement and control posture and balance.

The step initiation latency (~240ms) was not statistically different between PLAT and PULL conditions, and was similar to previously published findings (McIlroy and

Maki, 1996). Many of the kinematic measures did not appear to be different between the two perturbation methods as can be observed in Figure 4.6, which describes general movement pattern, based two representative participants. As anticipated, there was no change in heel position or in the angle of any the joints across conditions or sides prior to initiation of perturbation throughout the 1st epoch (-100-0 ms), which simply demonstrates that participants did not move prior to the onset of the perturbation. Following the onset of the perturbation, the heel trajectory was similar between methods during balance recovery. Further, NSS and SS knee joint angles showed little difference between PLAT and PULL trials. The NSS and SS hip and knee joints showed similar movement initiation onset between PLAT and PULL trials, which was manifested by the deflection of the angle signal from the pre-perturbation values. Moreover, the movement onset appeared to be no different between NSS and SS hip and knee joints.

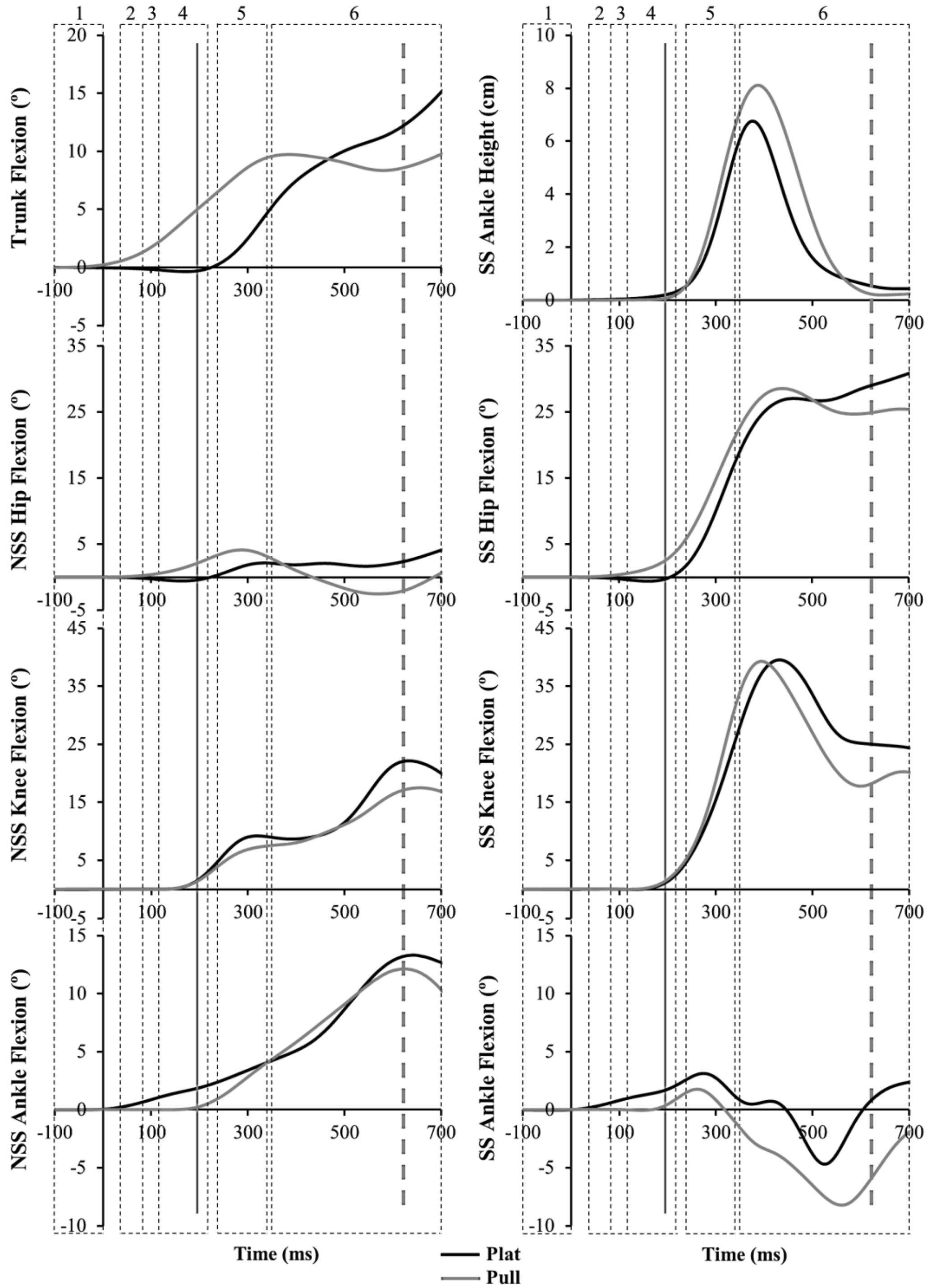

Figure 4.6. A graphical depiction of a representative sample of averaged (n = 2) time series for the trunk, hip, knee, ankle sagittal angle, and vertical position of SS ankle marker during PLAT and PULL trials. Flexion is positive. Thin grey solid line represents step initiation while thick grey dashed line represents foot contact.

The differences in balance recovery organization

The anterior muscles demonstrated difference in activation latencies between SS RF and SS TA in PULL trials, but not in PLAT trials. Longer activation latencies of SS RF at ~275ms when compared to SS TA at ~175ms, which were longer than those demonstrated by NSS RF at ~220ms, likely served to aid the swing of the SS limb as well as, perhaps, to stiffen the knee joint in preparation for the foot contact (Figure 4.2). Unlike anterior muscles, posterior muscles demonstrated dramatic differences between PLAT and PULL trials in muscle activation latencies.

In PLAT trials, posterior muscles became active between 70ms and 260ms (average: ~150ms), while in PULL trials, posterior muscle became active between 70ms and 130ms (average: ~90ms). This finding suggests that perhaps the balance-correcting response triggering may have been slower in PLAT trials than in PULL, though this can be argued. As discussed further below, more muscles were activated earlier in PULL trials, which created shorter average onset latency. The shortest activation during PLAT trials was demonstrated by the NSS and SS GM muscles at ~83ms, which is consistent with that previously reported (Horak et al., 1994); while in PULL trials, the shortest activation was demonstrated by the NSS and SS ES muscles, with an average latency of ~77ms. The difference between methods, with respect to the earliest response, was therefore minimal and unremarkable. While the quickness of response appeared to have been no different between methods, the critical differences in muscle activation were observed in the order with which posterior muscle became active.

The sequence of posterior muscle activation in PLAT trials was similar between NSS and SS muscles. However, more importantly, posterior muscles have demonstrated distal-proximal (GM, BF, and ES) sequence of activation, which is in line with the current hypothesis. The balance-correcting response was likely triggered by the ankle joint mechanoreceptors, since the ankle joint was the earliest among other joints to demonstrate movement onset in PLAT trials (Figure 4.6), which is also consistent with previously published findings (Horak et al., 1994). In PULL trials, consistent with the suggested hypothesis, muscles showed proximal-distal sequence of activation, but only in the NSS posterior muscles. Interestingly, in PULL trials, the activation sequence was different between the limbs unlike in PLAT trials. All three posterior SS muscles (ES, BF, and GM) became active very early during the 3rd epoch, but unlike posterior NSS muscles, their activation latencies (average: ~80ms) were not statistically different (Figure 4.2). This finding is in contrast to the suggested hypothesis of a proximal-distal muscle activation sequence. However, a similar effect was observed by Horak et al. (1994) when the participant’s head was accelerated, i.e. trunk, thigh, and leg posterior muscles showed simultaneous activation, suggesting a vestibular mechanism of balance-correcting response triggering. Moreover, as suggested by Allum and Honegger (1998), trunk muscles have the ability to trigger whole-body balance-correcting responses. Thus, it is conceivable that the movement of the upper body and the stretch of ES muscles triggered activation in all three posterior SS muscles during PULL trials. Perhaps, the purpose of activation of all three posterior SS muscles was to stiffen the stepping limb in order to produce a feet-in-place balance-correcting response, which participants were unable to utilize as the perturbation stimulus was sufficiently large to induce a stepping response.

The pre-perturbation (1^{st} epoch, -100-0 ms) NSS TA and SS GM muscle activity was found to be significantly larger in PLAT than in PULL trials. Previous research has demonstrated that during the step initiation, the stepping limb is unloaded through deactivation of posterior muscles and activation of anterior muscles of the stance limb (Crenna and Frigo, 1991). The activation of anterior stance limb muscles produces a shift in the body's centre of pressure toward the stance foot and unloads the stepping limb in preparation for the swing (Jian et al., 1993). Though the centre of pressure was not measured in the current study, it is conceivable that the function of larger muscle activity, shown by NSS TA and SS GM in PLAT trials during the 1^{st} epoch (pre-perturbation), was to unload the stepping limb in preparation for a step. The larger muscle activity in PLAT trials may have to do with higher anxiety experienced by participants in the PLAT trials as opposed to PULL trials. Previously, pre-perturbation anxiety has been shown to increase muscle tone, which was manifested by larger EMG amplitude (Cleworth et al., 2016). However, while it is insinuated that participants may have experienced higher anxiety during PLAT trials, as opposed to PULL, the anxiety was not measured in the current study.

Throughout the balance-correcting response, the most notable differences between methods were observed in posterior muscle activity but not in anterior muscle activity. Both NSS and SS ES muscles showed larger activity in PULL than in PLAT trials during the 2nd and 3rd epochs; while during the 4th epoch, only the SS ES showed larger activity (Figure 4.3 and 4.4). The NSS BF showed larger activity in PULL than in PLAT trials during the 2nd, 3rd, 4th, and 5th epochs (Figure 4.3); while the SS BF muscle showed larger activity during the 2nd and 3rd epochs (Figure 4.4). Shorter activation latencies and larger muscle activity observed in ES and BF muscles in PULL trials, but not in PLAT trials, suggests that initially participants may have attempted to stiffen the lumbopelvic region in order to limit the displacement of centre of mass or perhaps reduce the chances of postural collapse upon contact of the SS foot. Importantly, nearly all significant effects observed in the measure of muscle activity were associated with larger muscle activity during PULL trials, as opposed to PLAT trials, which is contrary to the suggested hypothesis.

Only a few muscles showed larger activity during balance recovery in PLAT than in PULL trials. The NSS GM and NSS RA showed larger activity early on during the balance-correcting response (3rd epoch) in PLAT trials. Larger activity of the NSS RA muscle during PLAT trials as early as the 3rd epoch, which is analogous to larger activity demonstrated by ES muscles during PULL trials, likely resulted from the stretch reflex evoked by the posterior movement of the platform which forced the trunk and hips into extension. This effect is evident from Figure 4.5 and is consistent with previously published findings (Horak et al., 1994). Importantly, larger activity of the majority of posterior muscles throughout the perturbation observed in PULL trials suggests a more robust response to PULL perturbation, as opposed to PLAT trials, which could in part explain the reason for platform-translation being more destabilizing than cable-pull perturbations as suggested in this series of studies (#MOS) and the previous research (Mansfield and Maki, 2009).

The absence of vision did not affect muscle activation latencies; however, it did have a significant effect on muscle activity. The majority of significant main effects of vision were observed in the NSS muscles, but not in the SS muscles (Table 4.3). During EC trials, the muscle activity was ~40% larger in select muscles, which are listed in Figures 4.3 and 4.4. Consistent with the suggested hypothesis and previous research (Gorgy et al., 2007), the absence of vision (EC) had a scaling effect on muscle activity across all the muscles which showed a significant effect. Furthermore, the co-contraction of the anterior and posterior muscles during EC trials likely resulted in stiffening of the joints, which may have led to a delay of ~20ms in step initiation latency between EO and EC trials (Figure 4.5A).

The time from perturbation onset to foot contact (~550ms) was similar to that previously reported (King et al., 2005; McIlroy and Maki, 1996) and, importantly, was not different between methods, which, again, is supported by previous findings (Mansfield and Maki, 2009). There was, however, an interaction effect observed between method and vision condition means, which was caused by differences in method and vision condition means observed in the measure of step time. The step time during PLAT-EC trials was significantly shorter than during PULL-EC and both PLAT-EO and PULL-EO trials (Figure 4.5A). Thus, although there was a delay with respect to step initiation latency in EC trials, participants appeared to have moved the stepping foot much faster during EC trials, specifically during PLAT-EC. Moreover, the step length, though not different between methods, was reduced when with vision removed (EC); and the difference between step length with vision absent was more pronounced in PLAT than in PULL trials (Figure 4.5B). The smaller step has been previously observed during balance recovery with vision absent and was interpreted as an indication of a more conservative balance-correcting response (Verniba and Gage, 2014), and is consistent with the current findings.

4.6. Conclusion

It is suggested in the current paper that the disagreement regarding the organization of balance-correcting responses exists due to the response dissociation from the type of perturbation used to elicit those responses. The current research identified some similarities in postural responses induced with platform-translation and shoulder-pull perturbation methods. However, there were also critical differences identified, specifically with respect to the neuromuscular organization of balance-correcting responses. This study has also investigated whether the absence of vision would have a dissimilar effect on balance-correcting responses induced with different perturbation methods. Though the absence of vision may have enhanced the sense of postural threat, which was manifested as larger muscle activity during balance recovery, it affected balance-correcting responses similarly across the perturbation methods.

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CHAPTER 5

Study 4 (#YA/OA): Stepping Boundary in Younger and Older Adults Perturbed with a Shoulder-pull Method

5.1. Summary

Introduction: Successful balance recovery is predicated on the type and characteristics of perturbation. The effect of age on relationship between perturbation characteristics has been investigated using platform-translation and wait-pull perturbations. No research has examined such relationship using shoulder-pull perturbation. The purpose of the current research was to investigate the displacement-force relationship required to elicit forward stepping responses with anterior shoulder-pull perturbations in younger (YA) and older (OA) adults.

Methods: Sixteen younger (25.8±3.1 years) and sixteen older (70.3±5.4 years) males participated. Unexpected perturbations were administered by the release of free weights which dropped a controlled height exerting an anterior pull on participants via a shoulder harness. Drop height (displacement) and amount (force) of weight were varied. The displacement-force combinations that elicited stepping responses were plotted for each participant. Minimum perturbation stimulus intensity (mechanical work) required to induce stepping responses was calculated and normalized to the participant's weight and foot length.

Results: The stepping boundary profiles (displacement-force) were not different between YA and OA groups. Further, there was no difference between YA and OA groups in normalized perturbation intensity means, 0.60±0.04 J/kg/m and 0.72±0.08 J/kg/m, respectively. However, visual inspection revealed two displacement-force relationship patterns in YA and three in OA. Four YA participants (YA-as-OA group) demonstrated displacement-force relationship similar to that of majority of OA participants (OA-Maj; *n* = 8). Four OA participants demonstrated relationship similar to that of the majority of YA participants (YA-Maj; *n* = 12). Four OA participants (OA-High) demonstrated distinctly different profiles than the rest of the groups.

Discussion: Unlike previous research, the current study showed no difference between YA and OA participants, which could have been due to the use of a different perturbation method. Further, this study highlights the performance cross-over between YA and OA participants, i.e. some OA participants behaved similarly to YA, and vice versa.

5.2. Introduction

The ability to recover balance following a postural perturbation is vital to mobility. Dynamic postural control can be compromised in special populations, such as older adults, resulting in poor balance recovery ability and falls. It has been shown that approximately 30% of adults over the age of 65 and 40% over the age of 75 fall each year. The percentage of fallers increases to 50% for older adults living in long-term care, and to 75% for those older adults who have fallen in the past (Rubenstein, 2006). Falls have been identified as a major cause of pain, disability, and early mortality (Murray and Lopez, 1997). Many reasons have been identified to suggest why older adults are at a greater risk of falling compared to younger adults, including: changes in peripheral sensory information, vision, and in central integration of sensory information with motor planning (Lord et al., 1994); reduction in muscle strength and power (Thelen et al., 1997; Wojcik et al., 2001); and psychosocial factors such as fear of falling (Vellas et al., 1997). Postural perturbation studies have been instrumental in investigating dynamic postural control in younger and older adults.

It has been shown that during postural perturbation, participants recover balance using a feet-in-place response or change-in-support response, such as a step (Maki and McIlroy, 2006). It has been demonstrated that older adults are more likely to use a change-in-support, or stepping, response than younger adults are and at lower thresholds of instability (Jensen et al., 2001; Mille et al., 2003). Further, it has been shown that the nature of response, feet-in-place or change-in-support, is largely determined by the magnitude of applied perturbation stimulus which is often composed of multiple components such as acceleration and the distance over which the perturbation is applied (e.g. Rogers et al., 2003). Previous research has investigated the relationship between perturbation characteristics that produce balance-correcting responses in younger adults (Verniba, 2016a). To date, the investigation of the relationship between perturbation characteristics which determine the type of correcting response is generally limited to platform-translation (e.g. Jensen et al., 2001) and waist-pull studies (e.g. Mille et al., 2003). Moreover, there is limited research that examined such relationship in older adults (e.g. Pai et al., 1998; Rogers et al., 2003). The purpose of the current study was to investigate differences between younger and older adults in the displacement-force relationship required to elicit forward stepping responses with an anterior shoulder-pull perturbation. The research questions were: 1) what perturbation intensity, defined as the combination of displacement and applied force, is required to elicit a forward stepping response in younger and older adults, and 2) is there a difference in the required perturbation intensity between groups? It was hypothesized that older adults would demonstrated higher sensitivity to perturbation characteristics and use stepping balance-correcting responses when exposed to smaller perturbation stimuli than younger adults.

5.3. Methods

Participants

Sixteen healthy older males (age 70.3 ± 5.4yrs, height 174 ± 6.6cm, body weight 74.3 ± 11.6kg, foot length 26.8 ± 1.3cm; mean ± SD) were recruited. Fourteen of the younger males previously participated in the #Thresholds study; two more participants were recruited to increase the size of the younger adult group to match that of the older adult group. Thus, sixteen healthy younger males (age 25.8 ± 3.1yrs, height 181 ± 5.2cm, body weight (BW) 79.6 ± 10.4kg, foot length 27.5 ± 0.9cm) participated. Participants were excluded if they reported a history of neurological or musculoskeletal disorders; or an injury, pain or surgery on their lower body and back in the six months prior to participation. York University research ethics board provided approval of the methods used in this study. All participants provided informed consent prior to participation.

Set-up and protocol

Participants were barefoot for the duration of the experiment. Base of support (BOS) length was defined as the participant’s foot length. All trials were initiated with participants standing on a stable support surface. Participants wore a shoulder harness, which was affixed to a custom-made perturbation device via a system of cables and pulleys. A detailed description of the perturbation system and its characteristics can be found elsewhere (Verniba, 2016a). The cables were linked to the shoulder harness at the approximate level of the manubriosternal joint from the front and T3/T4 vertebrae from the back. Unexpected anterior shoulder-pull perturbations were induced by the release of an electromagnet, which allowed free weights to fall a controlled distance and thereby exerting a pull on the participant. All perturbations were in the anterior direction. The timing of the weight release was randomly varied to prevent participants from predicting when the perturbation would occur. Participants were instructed as follows: "Behave as naturally as possible. If you don't have to take a step, don't take a step. If you feel the need to take a step to avoid falling, do take a step. Do what is natural to avoid falling."

The perturbation stimulus intensity was defined as the combination of applied force (%BW) and displacement (%BOS). The force was applied between 2.0%BW and 12%BW (4.45N increments), while the displacement was applied between 50%BOS and 155%BOS (2.2cm increments). The order of trials was blocked by applied force. The entire range of displacement iterations was applied within each force block, i.e. the testing began with the lowest weight and the lowest displacement. The displacement was iteratively increased from trial to trial until it reached the maximum value for each participant, while the force remained unchanged. The force was increased once the entire range of displacement per force block was tested; the testing then resumed with the lowest displacement and the process was repeated until the entire range of displacement was applied. The process was repeated until the combination of the highest force and highest displacement was tested. This protocol was used with half of the younger and half of the older participants. For the remaining participants, the protocol was repeated in a reverse order. To offset the effects of fatigue, participants received rest breaks between the force trial blocks. On average, participants performed two trials per minute.

Measures of interest

A complete step was defined as an anteriorly directed foot movement displacement that was larger than 100%BOS. Trials in which the displacement and force characteristics elicited a stepping response were coded as 1 by the primary investigator (DV). To assess the balance-correcting responses, the primary investigator stood next to the participant with the line of sight perpendicular to the line of step and in line with the adhesive tape (origin reference) placed in front of participant's toes. A trial that resulted in a feet-in-place response or a step that was smaller than 100%BOS was coded as 0. The response codes were placed in the 16x14 cell pull displacement-force matrix (Figure 5.1). The stepping boundary, which is the combination of the lowest force and associated displacement that produced the result of 1 (step), for each participant was mapped on a scatter plot to allow visual examination of individual differences in stepping threshold. The means were calculated for each displacement-force combination that produced a stepping response across all individual matrices within each group.

Perturbation stimulus intensity, defined as the amount of mechanical work done on a participant in order to induce a stepping balance-correcting response, was calculated. The mechanical work was calculated by multiplying the lowest displacement by the associated force that in combination produced a complete-step response. The mechanical work was normalized by participant BW and BOS length.

Force

Displacement (cm)	Displacement (BOS %)	13.4 N / 2.0 BW %	17.8 N / 2.7 BW %	22.3 N / 3.4 BW %	26.8 N / 4.1 BW %	31.2 N / 4.8 BW %	35.7 N / 5.4 BW %	40.1 N / 6.1 BW %	44.6 N / 6.8 BW %	49.1 N / 7.5 BW %	53.5 N / 8.2 BW %	58.0 N / 8.8 BW %	62.4 N / 9.5 BW %	66.9 N / 10.2 BW %	71.3 N / 10.9 BW %	75.8 N / 11.6 BW %	80.3 N / 12.2 BW %
13.2	51	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	1
15.4	59	0	0	0	0	0	0	0	0	0	0	0	0	0	0	1	1
17.6	68	0	0	0	0	0	0	0	0	0	0	0	0	0	1	1	1
19.8	76	0	0	0	0	0	0	0	0	0	0	0	0	0	1	1	1
22.0	85	0	0	0	0	0	0	0	0	0	0	0	0	0	1	1	1
24.2	93	0	0	0	0	0	0	0	0	0	0	0	0	0	1	1	1
26.4	102	0	0	0	0	0	0	0	0	0	0	0	0	1	1	1	1
28.6	110	0	0	0	0	0	0	0	0	0	0	0	0	1	1	1	1
30.8	118	0	0	0	0	0	0	0	0	0	0	0	0	1	1	1	1
33.0	127	0	0	0	0	0	0	0	0	0	0	0	1	1	1	1	1
35.2	135	0	0	0	0	0	0	0	0	0	0	1	1	1	1	1	1
37.4	144	0	0	0	0	0	0	0	0	0	0	1	1	1	1	1	1
39.6	152	0	0	0	0	0	0	0	0	0	0	1	1	1	1	1	1
41.8	161	0	0	0	0	0	0	0	0	0	0	1	1	1	1	1	1

Figure 5.1. A sample displacement-force matrix used to record perturbation responses for a participant with the weight of 67kg and BOS length of 26cm. Each blank cell was filled with either 0 (no step) or 1 (step).

Statistical analysis

All statistical analyses were conducted using JMP (v8.0, SAS Institute, North Carolina). The linear regression function was fitted to the complete-step threshold scatterplot data in order to establish stepping boundary profiles, which could then be compared between the groups. The linear function fit was considered significant at $p < 0.05$.

A one-way (*group* [YA/OA]) mixed effects (*participant* – random effect, *group* – fixed effect) repeated measures analysis of variance (rmANOVA) with *participant* nested within *group* was used to test for differences between participant groups in the amount of work required to induce a complete-step balance-correcting response. The statistical significance level was set at $p < 0.05$. Contrast analysis with Tukey HSD correction was performed to compare means. The effect size was reported using generalized eta squared (η_G^2) and considered trivial (< 0.02), small (0.2 – 0.12), moderate (0.13 – 0.25), and large (≥ 0.26) (Bakeman, 2005).

Follow-up exploratory analysis

Upon exploratory inspection of individual stepping boundary profiles and distribution of stimulus intensity values (mechanical work), it became clear that within the YA group there were participants who demonstrated stepping boundary profiles and perturbation stimulus intensity values remarkably similar to that demonstrated by the majority of OA, and vice versa. Thus, a follow-up exploratory appraisal was performed, where participants were grouped, within their respective age groups, according to the similarity of their stepping boundary profiles and perturbation stimulus intensity values.

5.4. Results

The combination of lowest displacement and associated force, which induced complete-step responses, demonstrated inverse linear relationships in both the YA and OA groups with thresholds at 130%BOS and 90%BOS, respectively. In the YA group, a significant (R^2 = 0.994, df = 9, p < 0.001) linear inverse relationship below 130%BOS was observed; the linear inverse relationship was not significant (R^2 = 0.906, df = 1, p = 0.139) above 130%BOS threshold. In the OA group, a significant (R^2 = 0.938, df = 4, p < 0.001) linear inverse relationship below 90%BOS was observed; the linear inverse relationship was significant (R^2 = 0.976, df = 6, p < 0.001) above 90%BOS threshold as well (Figure 5.2).

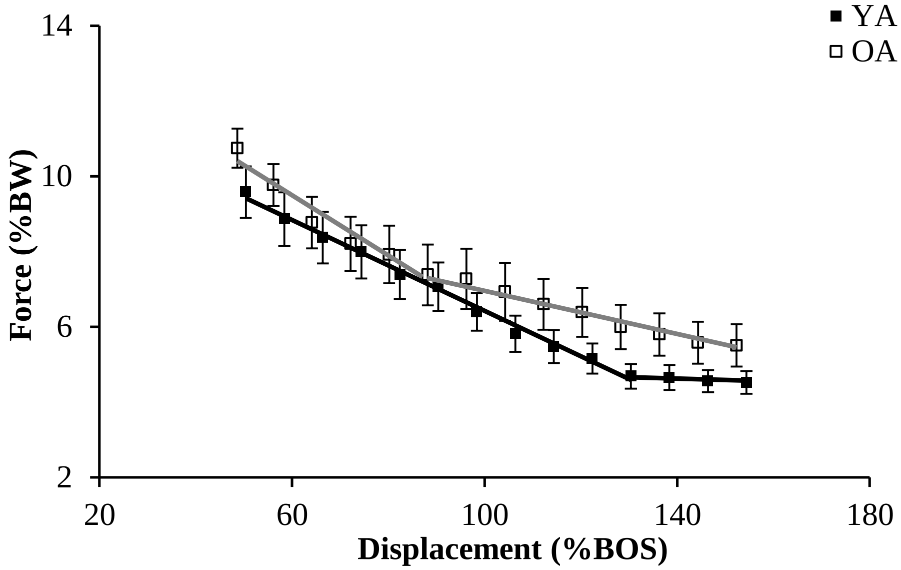

Figure 5.2. A graphical representation of perturbation characteristics required to produce stepping balance-correcting response in YA and OA groups with fitted linear functions. Error bars are SE. Both YA and OA demonstrated linear inverse displacement-force stepping boundary relationships with thresholds at 130%BOS and 90%BOS, respectively. The YA regression function was significant ($p < 0.01$) below the 130%BOS threshold, while OA regression function was significant ($p < 0.01$) above and below the 90%BOS displacement thresholds.

The statistical analysis showed no significant effect of *group* ($F(1,30) = 2.16$; $p = 0.152$; $\eta_G^2 = 0.07$). The normalized work means were not different between YA (0.60 ± 0.04 J/kg/m) and OA (0.72 ± 0.08 J/kg/m) groups.

Results of the exploratory follow-up analysis

Based on the follow-up qualitative analysis of all individual stepping boundary profiles placed on the common scatterplot, two patterns for YA and three patterns for OA were identified. Further, a box-and-whisker plot was constructed using perturbation stimulus intensity (mechanical work) results, which suggested participant placement within their respective YA and OA subgroups (Figure 5.3). Two YA subgroups were created with twelve participants placed in the “YA-Maj” and four in the “YA-as-OA” subgroup. Three OA subgroups were created with four participants placed in the “OA-as-YA”, eight in the “OA-Maj”, and four in the “OA-High” subgroup (Figure 5.4).

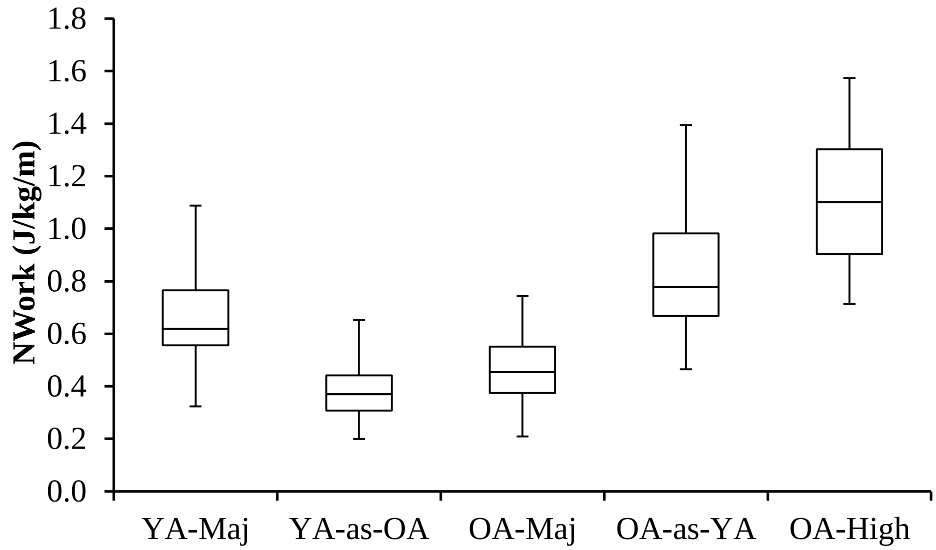

Figure 5.3. A box-and-whisker plot for normalized stimulus intensity (mechanical work) results for each individual within YA and OA subgroups. The normalized work value distribution is similar between OA-Maj and YA-as-OA and between OA-as-YA and YA-Maj.

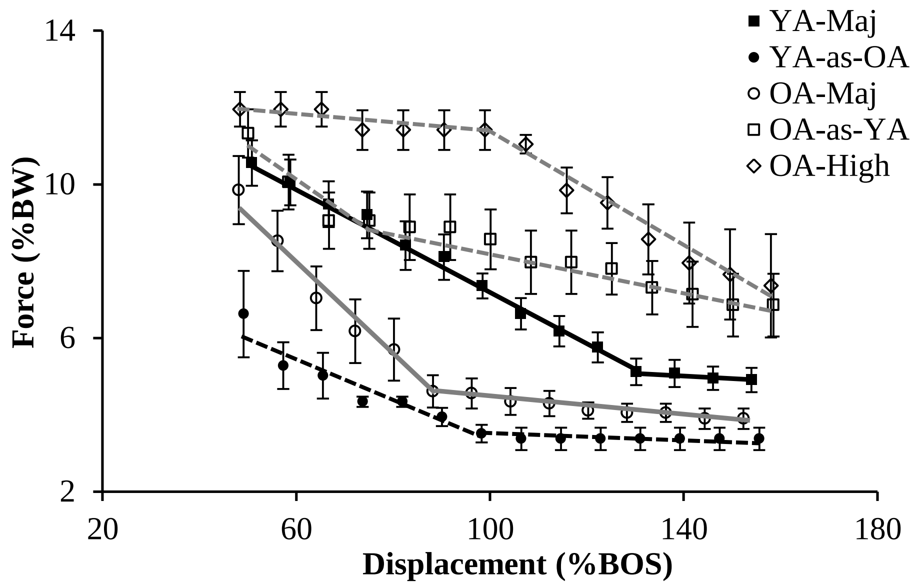

Figure 5.4. A graphical representation of group displacement-force stepping boundary means for YA (black trend lines) and OA (grey trend lines) subgroups. The error bars are SE. The figures present means for two YA groups: YA-Maj (*n* = 12) and YA-as-OA (*n* = 4); and three OA groups: OA-Maj (*n* = 8), OA-as-YA (*n* = 4), and OA-High (*n* = 4). The YA-Maj subgroup demonstrated a linear inverse relationship with threshold at 130%BOS; the displacement-force relationship of the YA-Maj subgroup showed a shallow slope of -0.01 above 130%BOS and steeper slope of -0.07 below 130%BOS. The YA-as-OA subgroup demonstrated a linear inverse displacement-force relationship with threshold at 100%BOS; the displacement-force relationship of the YA-as-OA subgroup showed a slope of -0.06 below 100%BOS and the slope of 0 above 100%BOS. The OA-Maj subgroup demonstrated a linear inverse displacement-force relationship with threshold at 90%BOS; the displacement-force relationship of the OA-Maj subgroup showed a shallow slope of -0.01 above 90%BOS and a steeper slope of -0.13 below 90%BOS. The OA-High subgroup demonstrated a linear inverse displacement-force relationship with threshold at approximately 100%BOS; the displacement-force relationship of the OA-High subgroup demonstrated a shallow slope of -0.01 below 100%BOS threshold and a steeper slope of -0.07 above 100%BOS.

5.5. Discussion

The purpose of the current study was to investigate differences between younger and older adults in the perturbation characteristic (displacement-force) relationship required to elicit forward stepping responses with an anterior shoulder-pull perturbation paradigm. It was hypothesized that older adults would demonstrated higher sensitivity to perturbation characteristics and use stepping balance-correcting responses when exposed to smaller perturbation stimuli than younger adults. The current hypotheses were rejected. Qualitatively, the perturbation characteristic (displacement-force) relationship was similar between older and younger adults (Figure 5.2). Furthermore, the perturbation stimulus intensity (mechanical work) required to produce stepping balance-correcting response was not statistically different between younger and older adult groups. The current findings are in contrast to previously published results (Jensen et al., 2001; Mille et al., 2003). However, that the current results suggest no difference between younger and older adults may be the function of perturbation method used. While the previous results were obtained using platform-translation (Jensen et al., 2001) and waist-pull (Mille et al., 2003) perturbation methods, the current findings were obtained using the shoulder-pull method. It has been previously suggested that balance-correcting responses are method-specific and that the shoulder-pull method might be easier for participants to accommodate compared to the platform-translation perturbation (Mansfield and Maki, 2009; Verniba, 2016b). Thus, the differences between previous and current findings may be attributed to differences in perturbation methods used. Interestingly, older adults, when compared to younger adults, demonstrated larger variability in perturbation characteristics and perturbation stimulus intensity (mechanical work) required to induce stepping responses. Upon further inspection of individual stepping boundary profiles, it became clear that within the OA group there were participants who adopted what appeared to be different balance-correcting strategies in response to the perturbation; for instance, some OA participants responded similarly to YA participants. Thus, an exploratory follow-up investigation was performed.

Exploratory follow-up investigation

The appropriateness of participant placement within one of the three OA and two YA subgroups was confirmed with a qualitative analysis of the distribution of perturbation stimulus intensity results (normalized mechanical work), the measure that combined both displacement and force perturbation characteristics (Figure 5.3). The qualitative analysis suggested that there may be two YA subgroups that may have been different from one another; the three OA subgroups may also have been different from one another. Importantly, it appears as though the OA-Maj subgroup was not different from the YA-as-OA, and that the OA-as-YA subgroup was not different from the YA-Maj subgroup; meanwhile, the OA-High subgroup appears to have been different from the other four subgroups. A qualitative comparison between the YA-Maj and OA-Maj subgroup stepping boundary plots revealed that the plots were similar, i.e. both displacement-force relationship plots showed a linear inverse relationship above and below group-specific displacement thresholds, which were 130%BOS for YA-Maj and

90%BOS for OA-Maj, respectively (Figure 5.4). The OA-Maj subgroup stepping boundary was lower with respect to the YA-Maj subgroup, which suggests that for a given displacement, a lower perturbation force was required to induce a stepping response in OA-Maj participants than in YA-Maj participants. This finding is consistent with previous research which demonstrated that older adults are more likely to step when presented with lower perturbation stimulus in comparison to younger adults (Pai et al., 1998). Moreover, below the respective displacement thresholds, 130%BOS for YA-Maj and 90%BOS for OA-Maj, the sensitivity to perturbation force in the OA-Maj subgroup was more pronounced than in the YA-Maj subgroup (Figure 5.4). With an increase in displacement, OA-Maj participants showed a larger decline in force required to induce a step than did YA-Maj participants, as the slope of trend line shown by YA-Maj participants was shallower (-0.07) than that shown by the OA-Maj subgroup (-0.13). Above the respective displacement thresholds, both the YA-Maj and OA-Maj subgroups showed similar sensitivity to displacement, as the slopes were equal (-0.01). The displacement thresholds suggest that above the thresholds, participants reached a minimum force requirement, ~5%BW for YA-Maj and ~4%BW for OA-Maj participants, that was required to induce a stepping response no matter how much the displacement increased. The relationship between the YA-Maj and OA-Maj subgroup stepping boundary plots is intuitive, however, as previously discussed, not all OA participants demonstrated balance recovery performance that was objectively inferior to that demonstrated by YA participants.

Contrary to previous findings (Mille et al., 2003; Pai et al., 1998), four OA participants, who were placed in OA-High subgroup, stepped following perturbations of larger intensity than participants in the YA-Maj subgroup (Figure 5.3). The stepping boundary profile demonstrated by participants in the OA-High subgroup is interesting in that the displacement-force relationship differs markedly from that shown by the rest of the YA and OA subgroups (Figure 5.4). While the majority of subgroup stepping boundary plots showed steeper negative slopes below the group-specific displacement thresholds and shallower slopes above the thresholds, the OA-High subgroup demonstrated an opposite relationship, i.e. shallower slope below and steeper slope above the displacement threshold (100%BOS) (Figure 5.4). Therefore, it can be suggested that below displacement threshold, OA-High participants were less sensitive to displacement characteristic of perturbation, unlike the participants in the other four subgroups. However, it is also important to note that a much larger perturbation force of approximately 11.5%BW was required to induce a stepping response in OA-High participants below the displacement threshold. Moreover, for every displacement quantity (data point), OA-High participants showed higher force than the rest of the participants in the other subgroups. Perhaps, participants in the OA-High subgroup, despite the instruction to behave naturally, actively resisted perturbation stimuli in an attempt to prevent themselves from stepping. However, above the displacement threshold of 100%BOS, OA-High participants were no longer able to resist the force of 11.5%BW and, therefore, utilized stepping responses. Thus, the force declined at a larger rate above the 100%BOS threshold than below the threshold. It is not feasible to suggest a reason for the markedly dissimilar performance of participants in the OA-High subgroup from their peers, and arguably, better performance than any of the younger participants. Other tests, such as balance confidence questionnaires, might be valuable and provide insight into why there were such differences between the subgroups. Of course, it is also possible that the OA-High subgroup were simply members of the OA-as-YA subgroup who wanted to demonstrate their youthfulness and vigor by trying not to step. Further research is needed to investigate the cognitive aspects of balance recovery in the OA participants who demonstrate balance-correcting performance similar to that shown by participants in the OA-High subgroup.

The remaining subgroups, where YA and OA participants performed similarly to those placed in the OA-as-YA and YA-as-OA subgroups, should be investigated further with a larger sample size. YA-as-OA participants may have been unmotivated and, thus, demonstrated lower stepping boundary, similar to that demonstrated by participants in the OA-Maj group (Figure 5.4). Participants in the OA-as-YA subgroup, who demonstrated stepping boundary profiles similar to that of YA-Maj participants, may have been motivated to perform as if they were younger due to being under observation by the researchers. It is also possible that OA-as-YA participants may have been fitter than their peers in the OA-Maj subgroup and were truly no different than YA-Maj participants. Future research should include measures of functional fitness and balance confidence questionnaires.

5.6. Conclusion

The current study was a preliminary examination of the relationship between perturbation characteristics in younger and older adults. Contrary to previous research, the current study showed no difference between younger and older adults, which could be due to the use of different perturbation paradigms. Further, the current study revealed an unexpected finding manifested by the lack of homogeneity in responses observed within participant groups, which prompted a follow-up investigation and creation of participant subgroups. The current study underscores that responses within age groups may not be stereotyped. Future investigation of the relationship between perturbation characteristics should consider psychosocial factors. Perhaps, the inclusion of questionnaires, such as fear of falling and balance confidence, may expose the reasons for the stepping boundary profile differences which were observed within age groups in the current study.

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CHAPTER 6

General Discussion

6.1. Dissertation Objectives Revisited

The objectives of the current dissertation were: 1) to directly compare two different types of perturbation on postural control and balance recovery, 2) to describe kinematic and neuromuscular responses to different perturbations, to explore similarities and differences in the nature of the responses, and 3) to examine the effect of age on the relationships between perturbation characteristics in determining balance-correcting response. It was found that the balance-correcting responses are unique to the modes of perturbation and that a dynamic examination of balance correcting responses is more revealing of characteristics that differentiate groups. Furthermore, the probing of the balance-correcting responses across a spectrum of perturbation characteristics might reveal subtleties about balance correction that are lost or not observed when a single measure is used.

6.2. Discussion

Our knowledge of human balance recovery, a critical aspect of successful locomotion, emanates from studies that have intentionally disrupted balance control. There has been a lot of disagreement in literature regarding the fundamental mechanisms of balance recovery. These studies, however, have utilized a number of different postural perturbation methods (e.g. Gage et al., 2008; Jensen et al., 2001; Mille et al., 2003; Verniba and Gage, 2014). The use of different perturbation methods may have resulted in disagreements within the research community (Allum et al., 2003). Recently, it has been suggested that balance-correcting responses may be method-specific (Mansfield and Maki, 2009). To date, however, the research addressing this topic is sparse. Thus, the central goal of the current dissertation was to compare balance-correcting responses induced with different perturbation methods: platform-translation and cable-pull (specifically shoulder-pull), which are among the most commonly used, ecologically valid, as well as temporally and spatially unpredictable perturbation methods.

The individual response to perturbation, feet-in-place or stepping, is predicated on the intensity of perturbation. Previous research has investigated the nature of the response using different perturbation methods and manipulating a combination of perturbation characteristics such as: acceleration or force, and the duration of acceleration or displacement resulting from translation (e.g. Jensen et al., 2001; Maki et al., 1996; Mille et al., 2003). In this dissertation, I have suggested, that in order for a reasonable comparison between balance-correcting responses that are induced with two different perturbation methods to be made, the experiment needs to be designed in such a way that perturbation types are the only manipulated conditions, while the stimulus characteristics, or the degree of disruption, is held constant. Therefore, prior to conducting any study that compared balance-correcting responses induced with two different perturbation methods, it was necessary to establish a method to equate the stimulus intensity between the two perturbation methods, which was addressed in Study #Thresholds (Chapter 2).

For the purposes of Study #Thresholds and, of course, the rest of the studies in the current series, the equipment was designed and built with the intent to allow for both types of perturbation set-up: platform-translation and shoulder-pull. All participants experienced both types of perturbation. Since the perturbation stimulus consists of multiple components, specifically the magnitude of applied force and distance over which the force is applied, it was necessary to establish the relationship between these characteristics in terms of the balance-correcting response. Thus, in Study #Thresholds (Chapter 2), the stepping boundary threshold, a relationship between perturbation characteristics that was required to induce a step, was determined for both perturbation methods by manipulating the applied force and perturbation displacement. As a result, an inverse linear force-displacement relationship was established for both platform-translation and shoulder-pull methods. The interpretation of the inverse relationship was such that any combination of force and displacement on or above the regression lines that represented that relationship would produce a stepping response with the associated perturbation method. Finally, a point was chosen at the intersection between the regression lines generated for platform-translation and shoulder-pull perturbations, which represented common perturbation characteristics for both methods. The perturbation characteristics at the regression function intersection were found to be equivalent to the force of 8.75% of the participant's body weight and displacement and of 105% of the participant's base of support. The intersection of the platform-translation and shoulder-pull force-distance regression lines was, therefore, the lowest force-displacement combination required to elicit a stepping response with both perturbation methods.

Subsequently, the perturbation characteristics common to both perturbation methods were used in Studies #MOS and #EMG (Chapters 3 and 4), which quantified and compared balance recovery strategies elicited with both methods.

Historically, position limit based measures, such as peak centre of mass or centre of pressure displacement during perturbation, have been used to quantify balance-correcting responses (e.g. Hlavacka and Horak, 2006; Verniba and Gage, 2014). However, it has been suggested that position limit based measures may not be adequate for dynamic situations such as those investigated in the current series of studies (Pai and Patton, 1997). Thus, for the purposes of Study #MOS (Chapter 3), which compared stepping balance-correcting responses induced with platform-translation and shoulder-pull perturbation methods, a relatively new concept in human balance control studies, the margin of stability, was used as the primary measure to quantify and compare balance-correcting responses between perturbation methods. The margin of stability has been suggested to be a measure that is better suited to describe the dynamic nature of postural control and balance recovery because it incorporates the velocity of center of mass movement in addition to the instantaneous position, which may be a more reflective model of how the human system estimates the degree of balance disruption at any given time (Hof et al., 2005). It is accepted that a larger margin of stability during balance recovery, compared to a smaller or negative margin of stability, is indicative of superior postural stability (Arampatzis et al., 2008; Carty et al., 2011; Karamanidis and Arampatzis, 2007). The margin of stability was calculated and reported at two time points during balance recovery: at step initiation and at foot contact. It was found that during platform translations, as opposed to shoulder pulls, participants demonstrated five times smaller margin of stability at step initiation and two times smaller margin of stability at foot contact. Furthermore, the larger destabilizing effect of a platform-translation perturbation, as opposed to a shoulder-pull, was also evident from the larger number of trials in which participants required a second step to fully recover balance: 14% of platform-translation trials, but only 3% of shoulder-pull trials. In older adults, an association between multistep balance-correcting response and larger postural instability and falls has been well documented (Maki and McIlroy, 2006). Thus, as previously suggested, the platform-translation perturbation appeared to be more challenging and destabilizing than shoulder-pull perturbation (Mansfield and Maki, 2009), which has been confirmed here.

That platform translations were more destabilizing than shoulder pulls, may be explained by the distribution of weight in the human body and the inertial properties of its segments. Assuming a double-pendulum model of the human body during quiet standing, hinged at the hips, and that two-thirds of the body mass is located in the upper body (Winter, 2005), the distribution of mass in the upper body is such that, unlike the lower body where greater mass is located proximally to the hips, the upper body has greater mass located distally to the hips. This creates a larger moment of inertia in the upper body when compared to the lower body. A larger moment of inertia means greater resistance to change in the object's state of motion. Likely because the upper body had larger moment of inertia than the lower body, it offered greater resistance to a perturbation stimulus than did the lower body. The margin of stability difference between platform-translation and shoulder-pull trials was, therefore, likely due to the mechanical effect of the specific type of perturbation used, rather than the neuromuscular control associated with the postural perturbation method. However, Study #MOS (Chapter 3) did not address the differences in the neuromuscular control between perturbation methods during balance recovery. Thus, Study #EMG (Chapter 4) has built on Study #MOS and advanced the understanding of the similarities and differences between platform-translation and shoulder-pull perturbations by examining the organization of balance-correcting responses and of neuromuscular control during balance recovery.

The purpose of Study #EMG (Chapter 4) was to investigate the organization of balance-correcting responses induced with platform-translation and shoulder-pull perturbation methods. The similarities and differences in balance-correcting responses were examined using spatiotemporal measures (step initiation latency and step distance) and electromyographic measures (muscle activation latency and muscle activation amplitude) in both stance and stepping side anterior (rectus abdominis, rectus femoris, and tibialis anterior) and posterior muscles (erector spinae, biceps femoris, and gastrocnemius medialis). The spatiotemporal and electromyographic measures in anterior muscles showed no difference across perturbation methods. Step initiation latency of approximately 240ms and time from perturbation initiation to foot contact of approximately 550ms, both similar to that previously published (King et al., 2005; McIlroy and Maki, 1996), and step length of 0.35m were not different between perturbation methods. The anterior muscles became active between 160ms and 275ms; however, the activation latencies and muscle activity amplitudes were not different across perturbation methods. Though Study #EMG (Chapter 4) identified many similarities between methods in organization of balance-correcting responses, some critical differences in posterior muscles were observed.

Consistent with the hypotheses, posterior muscles showed a distal-proximal muscle activation sequence on both stance and stepping sides during platform-translation. During the shoulder-pull trials, muscle activation was proximal-distal; however, this patterning was observed only in the stance side muscles. On the stepping side, all three posterior muscles became active at the same time. A similar effect has been previously described in a study where the participants' head was accelerated, using a specialized head rig, resulting in simultaneous proximal and distal posterior muscle activation, which suggested a vestibular mechanism of balance-correcting response triggering (Horak et al., 1994). Moreover, it has been also suggested that the trunk muscle stretch reflex can trigger whole-body balance-correcting responses (Allum and Honegger, 1998). Therefore, in the current study, the stretch response of the erector spinae muscle may have contributed to activation of the stepping side posterior muscles in response to shoulder-pull perturbations. Furthermore, dramatic differences across perturbation methods were noted in posterior muscle activity. Posterior muscles, specifically erector spinae and biceps femoris, consistently demonstrated larger activity during shoulder-pull perturbations as opposed to platform-translations, suggesting a more robust response to shoulder pulls. Perhaps, the platform translations were not perceived by the participants as adequately as the pulls were perceived, and maybe that is why platform translations have been suggested as more destabilizing by Mansfield and Maki (2009) and the current research (Study #MOS, Chapter 3). Though it is clear that different postural perturbations may elicit fundamentally different responses, there seems to be promise in utilizing balance recovery paradigms to probe the effects of injury, disease, or age on mobility and fall risk, and the dynamic threshold approach investigated in the current dissertation may prove to be more sensitive to differences or changes in balance control than the traditional means.

There is a wealth of research that has investigated the effect of age on balance-correcting responses; for instance, it has been shown that older adults when perturbed are more likely to step than younger adults and at lower thresholds of instability (Jensen et al., 2001; Mille et al., 2003). Further, it has been shown that the nature of response, feet-in-place or stepping, is determined by the magnitude perturbation stimulus which is, as previously discussed, composed of multiple components such as acceleration and the distance over which the perturbation is applied (e.g. Rogers et al., 2003). To date, the investigation of the relationship between perturbation characteristics which determine the type of correcting response is generally limited to platform-translation (e.g. Jensen et al., 2001) and waist-pull studies (e.g. Mille et al., 2003). Moreover, there is limited research that has examined the effect of age on this relationship (e.g. Jensen et al., 2001; Rogers et al., 2003), and no research has investigated such relationship using upper body perturbation paradigm such as shoulder-pull. This gap in research was addressed by Study #YA/OA (Chapter 5), which investigated the effect of age on the relationship between perturbation characteristics, displacement and force, and the perturbation stimulus intensity required to elicit forward stepping responses with an anterior shoulder-pull paradigm. The stimulus intensity was defined as the minimum amount of mechanical work done on a participant in order to induce a stepping response. Though previous literature, which utilized platform-translation (Jensen et al., 2001) and waist-pull (Mille et al., 2003) paradigms, suggested deficits in postural control and larger sensitivity to postural perturbations associated with age, the Study #YA/OA (Chapter 5) failed to reach the same conclusions. The relationship between perturbation characteristics (displacement and force) was qualitatively similar between younger and older adults. Moreover, the perturbation stimulus intensity (mechanical work) required to induce stepping balance-correcting responses was statistically not different between younger and older groups. That the results of Study #YA/OA, suggesting no difference between younger and older adults, are in disagreement with the previous findings, which were obtained using platform-translation (Jensen et al., 2001) and waist-pull (Mille et al., 2003), may be the function of different perturbation paradigms since it has been previously suggested that balance-correcting responses are method-specific (Mansfield and Maki, 2009) and was further supported by the current series of studies (Study #MOS and #EMG; Chapter 3 and 4). While there were no differences found between younger and older adults overall, an interesting observation was made. Upon further inspection of individual stepping boundary profiles it became clear that within the younger adult group, there were participants who demonstrated stepping boundary profiles as well as perturbation stimulus intensity values (mechanical work) remarkably similar to that demonstrated by the majority of older adults, and vice versa. Since there were no psychometrics or measures of functional fitness collected in Study #YA/OA, it was difficult to suggest the reasons for this unexpected finding; thus, further work is needed.

Limitations and future directions

This series of studies is limited by the recruitment of male participants. Since the primary purpose of these studies was to develop a methodology that would equate different perturbation types and to compare balance-correcting responses across perturbation methods, I opted to recruit a homogenous sample of participants. For the Study #YA/OA, the recruitment of participants was extended to older males as well. Future research should extend the investigation to include women, older women, as well as both male and female pathologic populations. Furthermore, the future research should investigate the repeatability of stepping boundary profile measures across both platform-translation and shoulder-pull methods and develop a robust statistical model that would allow for comparison across the stepping boundary profiles to be made. Studies #Threshold and #YA/OA may suffer from experimenter bias in terms of determining partial-step and complete-step thresholds, since no objective measures were collected. Future research should utilize objective data collection procedures such as the use of video or retroreflective markers. Lastly, Studies #MOS and #EMG may be limited by the mechanical properties of the perturbation system, which may have affected participant anticipation. There was a ~110ms and ~120ms delay between the drop of weights and the stimulus onset during PLAT and PULL trials, respectively. The sound of dropping weights was likely not an issue for anticipation as none of the muscles became active before the onset of perturbation, as defined by the movement of heel marker in PLAT and C7 marker in PULL trials. Contrary, the muscle activation latencies, locked to the perturbation stimulus onset, were highly consistent with previous literature.

6.3. Conclusion

There is potential in utilizing balance recovery paradigms to probe the effects of injury or disease on mobility and fall risk since dynamic scenarios resulting from balance disturbance can reveal more about the mechanisms of postural control than static scenarios (e.g. quiet standing). However, it appears that there is a lack of clarity in the literature, or perhaps only anecdotal recognition, that different postural perturbations may induce fundamentally different responses. Further work is needed in order to understand these differences before a selection of methods could possibly be used to best understand the deficiencies of different patient groups or special populations (e.g. older adults). The current series of studies took steps toward identifying the spatiotemporal, kinematic, and neuromuscular differences in balance-correcting responses induced with different perturbation methods, as well as examining the effect of age on the relationships between perturbation characteristics which determine the type of balance-correcting response (feet-in-place vs stepping). The current collection of studies suggests that while there are similarities in the balance-correcting responses between perturbation methods, there are also critical differences that are unique to the modes of perturbation utilized. The current dissertation underscores that caution is required when interpreting results of studies utilizing different perturbation methods and that individual differences between participants, which can mask age-related differences, need to be recognized.

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