INFLUENCE OF VIRTUAL HEIGHTS ON DYNAMIC POSTURAL CONTROL

by

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Abstract

Fear of falling and fall risk are strongly related. As such, falls are a leading cause of injury among older adults. Therefore, there is a need to study how balance is influenced by fear and fear related factors. Previous research has shown that when healthy young adults are exposed to a postural threat, by standing at the edge of an elevated support surface, their postural control is altered in both static and dynamic situations. While responses to support surface rotations have been studied in a high postural threat scenario [Carpenter et al., 2004b], the effects of threat on dynamic responses to support surface translations, a more ecologically valid type of perturbation, have not yet been investigated. This is due to several safety and feasibility issues that preclude translating individuals at height. Virtual reality (VR) has been established as an effective means for simulating height-related postural threat and can therefore be used to avoid the limitations associated with translating subjects at physical height. The purpose of this study was to examine the influence of fear and anxiety on postural reactions to surface translations during exposure to virtual heights. Twenty-one healthy young adults experienced support surface translations in the forward and backward directions, while immersed in a low and then high height virtual environment. Postural responses were significantly affected by height. Specifically, muscle activity in the lower leg and arm, and COP peak displacements, were earlier and larger in response to backward perturbations. No changes were observed in the responses to forward perturbations. In conclusion, virtual heights significantly altered neuromuscular responses to translational perturbations. Virtual height was capable of eliciting responses similar to real height, and thus may be used as an alternative method in investigating fear and anxiety related balance deficits in populations with a known fear of falling.

Preface

This thesis was reviewed and approved by the University of British Columbia Clinical Research Ethics Board (Central and Peripheral mechanisms contributing to human balance control, UBC CREB# H06-70316; see Appendix B).

The study presented in this thesis was not submitted for publication at the time of thesis submission.

I was the lead investigator in this thesis. I was responsible for concept development, data collection, data analysis, virtual environment development, and manuscript composition. Members of the Neural Control of Posture and Movement Lab were assistive in data collection. Carpenter, MG was the principle supervisor and was involved in manuscript edits and concept development.

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Chapter 1: Introduction

1.1 Fear of Falling

Fear of falling is prevalent in a number of populations, including older adults and individuals with Parkinson's disease, and is strongly related to fall risk [Friedman, Muniz, West, Rubin, & Fried, 2002]. Falls, in turn, are a leading cause of injury among older adults and can lead to social, psychological and physiological consequences including reduced balance confidence, increased fear of falling and risk of injury, and other negative consequences [Hindmarsh & Estes, 1989]. Given that falls have been reported in approximately 35% of community dwelling older adults [Blake et al., 1988; Tinetti, Speechly, & Ginter, 1988], there is a need to understand how balance is influenced by fear and related co-factors such as anxiety, balance confidence, and perceived stability.

1.2 Traditional Relationship between Fear and Falls

Typically, fear of falling has been associated with a 'vicious cycle' of falls whereby a fall leads to fear and avoidance of activities that are perceived to pose a fall risk. This more sedentary lifestyle, in turn, increases the risk of future falls [Friedman et al., 2002]. Fear of falling has been shown to be a direct consequence of a fall [Tinetti et al., 1988], and is a major contributor to the 'post-fall syndrome' [Murphy & Isaacs, 1982]. This fear of falling has been shown to be present in 26% [Howland et al., 1993] to 92% [Aoyagi et al., 1998] of community dwelling older adults who have experienced a fall. With an increase in fear, individuals can suffer from a wide range of adverse consequences including social withdrawal [Howland et al., 1993], anxiety and depression [Burker et al., 1995], increases in inactivity [Tinetti, de Leon, Doucette, & Baker, 1994; Howland et al., 1993; Nevitt, Cummings, Kidd, & Black, 1989], loss of independence [Vellas, Cayla, Bocquet, de Pamille, & Albarede, 1987], and decreased quality of life [Howland et al., 1993; Nevitt et al., 1989]. In particular, a decline in physical activity can cause a reduction in muscle strength and overall health. In elderly individuals specifically, this consequence is debilitating due to a pre-existing age related decline in muscle strength and coordination [Grimby & Saltin, 1983]. The ability to perform activities of daily living is compromised and can cause the individual to fall again, thus restarting the vicious cycle.

1.3 A Link between Fear and Balance: An Alternative Explanation

Fear is not always a consequence of falling, as fear can directly influence balance even before a fall has occurred. Fear of falling and falls have been shown to be predictors of each other and each can lead to a spiraling effect of increased falls, fear, and functional decline [Friedman et al., 2002]. For example, a fear of falling has been shown to be present in 12% to 65% of older adults who live independently and *do not* have a history of falling [Lachman et al., 1998; Lawrence et al., 1998; Howland et al., 1998; Tinetti et al., 1994; Howland et al., 1993]. Furthermore, there is new evidence to suggest a more direct link between fear and balance. First, Balaban and Thayer [2001] suggest neuroanatomical structures related to postural control are also related to anxiety. For instance, anxiety disorders and balance disorders are thought to be interconnected by a link between the parabrachial nucleus, vestibular nuclei, amygdala, hypothalamus, and orbitofrontal cortex [Balaban & Thayer, 2001; Balaban, 2002]. Reciprocal connections between some of these structures are responsible for vestibular processing, autonomic function, emotional responses, and anxiety [Balaban & Thayer, 2001; Balaban, 2002]. Second, behavioural studies have shown direct effects of fear on postural control. Specifically, studies using a support surface height manipulation, known to induce an increase in physiological arousal, anxiety, and fear, have demonstrated changes in both quiet stance and dynamic postural control [Carpenter, Frank, Adkin, Paton, & Allum, 2004b; Davis, Campbell, Adkin, & Carpenter, 2009]. A change in support surface height directly manipulates postural threat, allowing for a unique comparison between low and high levels of threat within otherwise healthy humans.

1.4 Quiet Stance Affected by Fear

To date, when examining the influences of fear on balance, a quiet stance paradigm has typically been used. When young healthy individuals stand quietly at heights of 1.6 meters or 3.2 meters, a “theoretically” more stable posture is implemented by adopting a stiffening strategy (decreasing the amplitude variability and increasing the frequency of centre of pressure (COP, mean position of ground reaction forces) and centre of mass (COM) displacements) [Carpenter, Frank, & Silcher, 1999; Adkin, Frank, Carpenter, & Peysar, 2000; Carpenter, Frank, Silcher, & Peysar, 2001; Davis et al., 2009]. It has been hypothesized that subjects adopt this stiffening strategy to enhance the control of the COM over the base of support (BOS) when the level of threat is perceived to increase [Carpenter et al., 1999]. Subjects also lean away from the perceived threat shifting the COP [Carpenter et al., 1999; Davis et al., 2009; Huffman, Horslen, Carpenter, & Adkin, 2009] and COM [Carpenter et al., 2001] mean position backwards. The described stiffening strategy with threat is not consistently observed across studies as other factors can change the way threat influences stance. For example, conscious control influences the amount of backward lean as more consciously aware subjects leaned further away [Huffman et al., 2009], and when people become very fearful, COP amplitudes increase [Davis et al., 2009]. It is apparent that balance is influenced by emotion, specifically fear and anxiety, however further investigation of this effect on balance is necessary in order to fully understand its role in postural control.

1.5 Shift to Dynamic Postural Control

It has been hypothesized that threat related changes, while beneficial during moments of quiet stance, may place the individual at a greater risk of falling in situations where reactive postural control is required. For example, if individuals are leaning and adopting a stiffer posture, similar forces will lead to greater postural instability [Bloem, Allum, Carpenter, Verschuuren, & Honegger, 2002; Allum, Carpenter, Honegger, Adkin, & Bloem, 2002]. Furthermore, understanding postural threat effects on dynamic postural control is necessary because 30% to 60% of falls occur due to the inability to produce adequate responses to external perturbations [Horak, Henry, & Shumway-cook, 1997; Maki & McIlroy, 1996; Blake et al., 1988]. For example, 50% of falls in elderly individuals living at home are thought to arise from tripping [Blake et al., 1988], and 54% of falls in elderly individuals are thought to come from a perturbation to the BOS [Maki & McIlroy, 1996]. Horak et al. [1997] state that of the falls suffered from insufficient postural responses, 50% come from a sudden movement of the BOS, such as a slip. Therefore, dynamic postural control paradigms need to be considered when investigating the effects of emotion on balance.

There is some evidence to suggest that fear influences the neural mechanisms that underlie the human responses to postural perturbations. Following a perturbation to balance, the central nervous system detects and predicts instability [Horak et al., 1997] to allow the coordination of muscle activity and limb movements to regain balance. Sensory information, primarily from proprioceptive and vestibular systems, used in balance correcting responses is thought to pass through the brainstem and higher cortical areas to trigger and modulate the postural responses [Macpherson & Fung, 1999]. Recent evidence suggests that both these sensory systems may be influenced by fear and anxiety and therefore may contribute to changes observed in dynamic postural control [Yardley, Watson, Britton, Lear, & Bird, 1995; Wada, Sunaga, & Nagal, 2001; Carpenter et al., 2004b; Sibley, Carpenter, Perry, & Frank, 2007; Horslen, 2010]. For example, spindle sensitivity in postural muscles (triceps surae) has been shown to be facilitated when standing in an environment of increased postural threat. Soleus tendon reflexes are facilitated [Davis et al., 2011] while soleus Hoffman reflexes are attenuated [Sibley et al., 2007] or remain constant [Horslen, 2010] when standing at elevated heights. Tendon reflexes are also facilitated in highly aroused seated subjects [Bonnet, Bradley, Lang, & Requin, 1995; Both, Boxtel, Stekelenburg, Everaerd, & Laan, 2005]. No height related changes to tendon tap-evoked potentials and somatosensory-evoked potentials have been observed, suggesting that increases in spindle sensitivity with height do not lead to increased afferent feedback gain at the somatosensory cortex [Davis et al., 2011]. Vestibular function has also been shown to be altered by fear and/or arousal [Yardley et al., 1995; Wada et al., 2001; Bolmont, Gangloff, Vouriot, & Perrin, 2002]. For example, increased anxiety results in increases in vestibular-ocular reflex gain [Yardley et al., 1995], and has been proposed to increase vestibular gain [Carpenter et al., 2004b]. However, the extent to which height related changes in proprioceptive and vestibular function contribute to dynamic balance control is unknown.

1.6 What is Dynamic Postural Control?

Dynamic postural control can be explored by exposing subjects to perturbations artificially created within a lab. Mechanical perturbations, displacing the body away from equilibrium by shifting body segments and changing the COM position [Horak et al., 1997], can come in different forms. Upper body, or trunk, perturbations can be caused by a push to the back [Brown & Frank, 1997], or the sudden release of a tonic load attached to the trunk [Sibley, Mochizuki, Frank, & McIlroy, 2010]. Another example is support surface perturbations, where the support surface beneath the feet is rotated around the ankle joint in either a pitch, roll or combined pitch/roll direction [Carpenter et al., 2004b], or translated in a horizontal manner in either the antero-posterior [Okada, Hirakawa, Takada, & Kinoshita, 2001] or medio-lateral direction [Campbell, 2012]. A balance correcting response to an unexpected perturbation is typically reactive, where whole body responses are used to equilibrate the COM over the BOS. However, there is evidence to suggest proactive strategies are present during any perturbation [Horak, 1996] as well. Two approaches to maintaining upright stance can be used when regaining equilibrium during COM disturbances: a) the body generates muscle activation patterns and joint torques in order to keep the COM within the BOS; or b) the body creates a new BOS (by taking a step or grasping a standing aid) to keep the COM within a BOS [Maki & McIlroy, 1996]. Irrespective of the strategy used to remain upright, the balance correcting responses, which are stereotypical, multi-jointed, and direction specific to the unexpected perturbation, are more complex than simple stretch reflexes [Rothwell, 1994]. Postural responses typically have latencies of approximately 120 ms, placing them between early stretch reflexes and later voluntary responses [Diener, Bootz, Dichgans, & Bruzek, 1983; Carpenter et al., 2004b]. These responses can be formed in different patterns, or synergies, depending on the postural systems strategy to remain upright. One well-documented pattern is an ankle strategy [Horak & Nashner, 1986], which uses a distal to proximal activation pattern to restore the body to a stable upright position. For example, a forward translation causes a backward shift in COM and the body sways backward. In the ankle strategy, the shank, thigh, and then trunk and arm muscles contract to bring the body forward. Another known strategy is the hip strategy. An example of a hip strategy is seen in adults who do not elicit the typical distal to proximal muscle activation pattern when standing on support surfaces shorter than foot length. Here, the same directional perturbation elicits a proximal to distal muscle activation in comparison to the ankle strategy [Horak & Nashner, 1986]. To note, a stepping action can take precedence in situations where feet are not held in place or when physical barriers do not constrain stepping. Irrespective of the postural response used to remain upright, the postural system can use different strategies to acquire the same goal of maintaining equilibrium. Of interest in this study is how an individual might adapt their strategy when exposed to a postural threat.

1.7 Current Research on Fear and Dynamic Stance

To date there have only been a few studies examining the effects of fear and anxiety on biomechanical, kinematic, and kinetic characteristics of dynamic posture. Brown and Frank [1997] manipulated support surface height to examine the effect of threat on kinematic responses to a destabilizing trunk perturbation. Participants experienced a series of external trunk perturbations in two threat conditions; a non-threatening condition performed on the ground, and a threatening condition performed at the edge of a hydraulic lift raised to a height that exceeded the maximum height they could step down from (81.44% of each participant's leg length). The use of a height paradigm allows for the experimenter to manipulate the level of threat, which allows for independent manipulation of fear and anxiety within a subject, as opposed to testing different populations with predisposed levels of fear. When at the high height, subjects reacted with anticipatory and reactive strategies different to what was seen at the low height. A lean away from the edge was seen prior to any perturbations, which falls in line with what has been reported in quiet stance experiments [Carpenter et al., 1999; Carpenter et al., 2001; Adkin et al., 2000; Davis et al., 2009; Huffman et al., 2009; Cleworth, Horslen, & Carpenter, 2012] and may act to counter the effect of the forward push. In response to the perturbation at height, subjects adopted a reduction, or even reversal in the direction, of COM displacement. Also, time to first peak COM velocity was decreased in the high threat condition. The reason for why these changes occur is thought to be from the adoption of a 'safer' postural strategy. The reduction in COM displacement and posterior shift in COM position decreases the risk of falling by creating a greater distance between the COM and the limits of the BOS. Therefore, as the severity of consequences to instability is increased, the COM may be more tightly controlled.

A second study on emotion and dynamic postural control used surface translations to examine the influences of fear of falling on balance ability in elderly women [Okada et al., 2001]. In this study, participants experienced horizontal support surface translations in the anterior direction. The key analysis was performed on the deceleration component of the forward translation (which was in the posterior direction). The goal of this study was to examine the effect of fear of falling by comparing elderly women with and without a fear of falling. Therefore, two groups (fearful and non-fearful) were matched in motor ability. The results showed that fear of falling was associated with changes in the postural reaction to a support surface deceleration. First, fearful individuals showed a larger displacement in COP (termed centre of foot position) and a longer time to this displacement in the first active phase of the response to the perturbation. The fearful group also co-contracted the tibialis anterior (TA) and gastrocnemius (GAS) to a larger degree, and had a smaller relative onset (difference between onset of TA and GAS) between the two muscles, than the non-fearful group. Therefore, fear of falling affects the recovery to a small perturbation only during the initial period of the response. It was suggested that a stiffening strategy is used when fearful to minimize the risk of falling. This stiffening strategy (proposed in other work during quiet standing [Carpenter et al., 1999]) is supported by the increase in co-contraction and increased displacement of COP in the fearful group. The results, as proposed by Okada et al. [2001] demonstrated an unfavourable effect of fear on postural control. A stiffer control, seen by increased co-contraction in the lower leg in people with a fear of falling, may have resulted in the larger COP displacement validating that a similar force can have a greater toppling effect in fearful adults.

A third study examining fear influences on dynamic postural control observed changes in postural reactions to surface rotations under conditions of postural threat [Carpenter et al., 2004b]. The goal of the study was to examine how postural anxiety influences the muscular and biomechanical responses to perturbations. Carpenter et al. [2004b] placed subjects on a rotating platform located on a hydraulic lift, allowing for two conditions of threat (a low threat height of 60 cm, and a high threat height of 160 cm), where a series of multi-directional perturbations were delivered at each height. Under conditions of increased postural threat, and subsequently increased anxiety, subjects experienced: a) a reduction in magnitude of COM displacements; b) decreased amplitude of segment movements of the leg and trunk; c) reduced angular displacements of the leg and trunk; d) increased amplitude of muscle activity in balance correcting responses of the lower leg, trunk and arm muscles; and e) early onset of the deltoid muscle. Therefore, Carpenter et al. [2004b] also demonstrated responses to unexpected perturbations are affected by increased postural threat through observations in muscular and biomechanical characteristics.

Few studies have simultaneously observed the influences of fear and anxiety on dynamic postural control via cortical activity related to postural perturbations, and postural responses. Sibley et al. [2010] and Adkin, Campbell, Chua, and Carpenter [2008] both report an increase in the amplitude of the N1 potential when experiencing unpredictable perturbations (to the upper trunk) at height; however, different EMG responses were reported. When subjects experienced a force applied to the upper back, no differences in EMG were reported between a low and high height [Adkin et al., 2008]. The limited results may be explained by the nature of the perturbation (small in amplitude, and unidirectional). When subjects experienced the release of a load, causing forward displacement of the body, muscle activity of the lower leg activated earlier and larger at a high height compared to a low height [Sibley et al., 2010]. COP displacements were also influenced in the load release paradigm, as subject's peak COP displacement was earlier and larger at height [Sibley et al., 2010]. The results from these studies suggest that anxiety may influence postural control through cortical processes during responses to external perturbations.

1.8 Limitations and Gaps in Current Research

It is clear that there is an influence of postural threat induced fear and/or anxiety on dynamic postural control. However, there are obstacles to this line of work that require more research in order to truly understand how reactions to unexpected perturbations are affected by fear and anxiety. The problem in trying to extract common findings from current literature is that the studies use different methods to perturb individuals, and use two different forms of fear manipulation (height-induced and pre-existing). Different perturbations cause different postural reactions, and therefore comparing across these studies proves to be difficult. In terms of fear manipulation, one study used a fear trait to compare between fearful and non-fearful groups, whereas the other two used height manipulation to change the level of fear (state anxiety/fear) within an individual. Both methods of observed fear have their benefits and drawbacks. The height manipulation allows for a within subjects comparison, where the only theorized manipulations are the levels of fear and/or anxiety. Although promising, fear and/or anxiety levels may not be consistent across all subjects. This is not the same for the predisposed fear studies which does not allow for a within subject analysis, and therefore increases sample variability. When testing between subjects, one with a fear of falling and one without, subjects' functional ability can be, and usually is, different. When subjects are different in variables other than fear, the changes observed between these subjects cannot solely be attributed to fear. The between subjects variability limitation applies to the Okada et al. [2001] study, where differences were observed in two motor ability tests between the two groups (one legged stance time with eyes closed and longer COP displacements with eyes closed). However, there are many benefits of testing individuals with a predisposed fear, including: a) the subjects predetermined split into groups eliminates the need for fear manipulation (one group will always have a higher level of fear); and b) ecological validity is improved by testing the population most affected by increased falls risk related to fear of falling.

Aside from the difference in attributed fear, there are many other limitations associated with these studies. The measures used in the three biomechanical and kinetic studies are not the same, again making comparisons very difficult. For electromyography (EMG) analysis, one observed amplitudes and absolute onsets [Carpenter et al., 2004b], one observed relative onsets and relative co-contraction [Okada et al., 2001], and one did not report any EMG [Brown & Frank, 1997]. EMG observation is critical as Carpenter et al. [2004b] found significant changes specifically within their EMG analyses. Furthermore, two studies did not analyze COP measures [Brown & Frank, 1997; Carpenter et al., 2004b] while another excluded COM analysis [Okada et al., 2001]. Without consistent variables between studies, it is difficult to get a true representation of the effects fear and anxiety has on postural reactions to different perturbations.

Another limitation to this line of work is the use of direction specific threats and perturbations, which can lead to anticipatory postural adjustments. Both Brown and Frank [1997] and Carpenter et al. [2004b] used a uni-directional threat to induce their postural threat. This creates a direction specific threat, not typically encountered by individuals who have a fear of falling, and also creates a 'safety zone' in which subjects may perceive as a desired location to step towards. Anticipatory actions such as leaning, which can change the stretch and balance correcting responses during postural responses to a perturbation [Diener et al., 1983; Horak & Moore, 1993], can be used as a safety adjustment when the perceived threat is unidirectional. Unidirectional perturbations, such as those used in previous studies [Brown & Frank, 1997; Okada et al., 2001; Adkin et al., 2008; Sibley et al., 2010] can also lead to anticipatory postural changes. One final limitation is the length, in terms of displacement and time, of the perturbation used. A short amplitude or duration perturbation, seen in the support surface perturbation studies, makes it very difficult to dissociate the difference between the acceleration and deceleration phases of the perturbation. Both acceleration and deceleration of the support surface perturbs the body and requires a postural response; however it is the initial acceleration that first destabilizes the body and typically is of interest to researchers [Visser, Carpenter, van der Kooij, & Bloem, 2008]. Postural responses to both the acceleration and deceleration phases can mix when they are presented too close together [McIlroy & Maki, 1994] leading to contaminated responses. Also, the deceleration phase can be used as a stabilizing component as it counteracts the effects driven by the acceleration phase [van Asseldonk, Carpenter, van der Helm, & van der Kooij, 2007]. Therefore, there is still a need to examine the influences fear and its co-factors (including anxiety, balance confidence, and perceived stability) have on balance correcting responses. The goal of this study is to use height manipulation as a postural threat in a bi-directional manner, examining how postural reactions to a translation are influenced by fear and anxiety. The proposed study will improve upon the prior limitations by: a) focusing on young healthy adults to ensure any change will be attributed to the manipulation of threat, b) using a translating sled capable of delivering long (0.75 m) support surface translations in two directions to eliminate the confounding effects of deceleration and anticipatory postural changes, c) including measurements of psycho-social state, kinematic, kinetic and electromyography variables, and d) a bi-directional threat created by a similar distance to the edge in the anterior and posterior direction.

1.9 Feasibility and Safety Issues

The placement of a long translating sled (1.5 m) on a hydraulic platform raises many feasibility and safety issues. For example, the sled length makes it difficult, especially when an edge is present, for a spotter to reliably be at the participant's side while being perturbed. Also, the torque that would be generated by the sled accelerating on an elevated hydraulic lift would cause the lift to shake considerably. This sway in the lift would make COP measurements unreliable, and further decrease the stability of the participant, thereby increasing the risk of falling. Therefore, there is a need to study the effect of fear on balance while overcoming the safety and feasibility issues related to the confines of a laboratory. Virtual reality (VR) has been proposed as a suitable replacement for placing individuals at a perceived height while studying balance in a safe environment [Simeonov, Hsiao, Dotson, & Ammons, 2005; Cleworth et al., 2012].

1.10 Virtual Reality and Perceived Height

VR allows humans to be immersed within computer-generated environments that simulate real, and even imaginary, world experiences and environments. It has been shown to be a useful tool in posturography by replicating the real world experiences seen in daily living while still remaining within the confines of the laboratory [Whitney et al., 2006]. These environments allow for the further study of the sensory system and the central nervous system in otherwise inaccessible or dangerous real life situations.

VR has been frequently used in behavioural research to place subjects in environments capable of eliciting emotional responses. Emotional responses can include high levels of anxiety associated with having to make an oral presentation in front of an audience [Anderson, Rothbaum, & Hodges, 2003], or increased levels of fear when standing at a simulated height [Cleworth et al., 2012]. Virtual heights have been used to test and treat people with acrophobia (a fear of heights) to help diminish their phobia (termed virtual reality exposure therapy). It was reported that when experiencing virtual heights as a tool for rehabilitation, an improvement in attitudes toward and at height were observed [Hodges et al., 1995; Rothbaum et al., 1995, Emmelkamp, Bruynzeel, Dorst, & van der Mast, 2001]. Although promising, this line of work is limited in its ability to generalize, as the researchers could only conclude that VR is beneficial when observing height related changes in those with a predetermined fear of heights.

Placing individuals at perceived heights relies on the effectiveness of the virtual environment. Since the goal is to change the emotional response of an individual by placing them near an undesired source (e.g., an elevated height for people with acrophobia), it is important to make the individuals feel as though they are in a stressful environment. Hodges, Rothbaum, Kooper, Opdyke, Meyer, de Graaff, Williford, and North [1994] suggest presence is the most important factor when implementing VR in treatment of acrophobia. ‘Sense of presence’ (or illusion of presence) is the subject’s perception of *being within* the virtual environment, as opposed to *watching* the environment on a computer screen while being in a laboratory [Meehan, Razzaque, Insko, Whitton, & Brooks, 2005]. Presence can be inferred using physiological measures such as heart rate, skin conductance and skin temperature [Meehan, Insko, Whitton, & Brooks, 2002] or self-reported psychological questionnaires that probe many aspects of reported and behavioural presence, including: sense of “being within” the environment, feelings of the virtual world becoming reality, frequency of “visiting” instead of “viewing” the virtual environment, and whether behavior is the same when in a similar real environment [Slater, Usoh, & Steed, 1995; Usoh et al., 1999]. An increase in heart rate and skin conductance demonstrated that there are methodological and technical features that can help increase the effectiveness of virtual environments (physiological reaction and reported presence) including immersion (practice in a non-stressful situation to situate the individual within the virtual world), haptic cues (proprioceptive sensations from physical elements in the real environment are matched visually by objects seen in the virtual environment), and frame rate (reduction in time between actual head movements and scene movements) [Meehan et al., 2005].

1.11 Virtual Heights versus Real Heights

It is important to understand the similarities and differences seen in humans when placed in real versus virtual environments, specifically ones that have a postural threat. There have only been a few studies that directly compare the influence of height manipulation on balance in both real and virtual environments. Virtual heights have been shown to elicit similar changes with height during quiet stance as real height exposure. Simeonov et al. [2005] observed changes to postural stability during quiet stance when faced with a real and virtual height and observed similar changes in psychological and physiological aspects of fear and anxiety. However, due to the multifactorial design used (support firmness, visual target distance in addition to surface height manipulation) it is difficult to allocate where the differences or similarities lie between the heights in each environment. Cleworth et al. [2012], in a more focused study design, demonstrated similar changes to postural control, psycho-social state and autonomic function from low to high heights when compared across real and virtual environments. COP amplitude decreased and frequency increased, in accordance to what has been reported previously [Carpenter et al., 1999; Adkin et al., 2000; Davis et al., 2009; Huffman et al., 2009], in both real and virtual environments. Also, levels of fear, anxiety, and physiological arousal increased, while perceived stability and balance confidence decreased with height in both visual environments.

Fear related changes in gait have also been observed in both real and virtual environments, although not in the same study. Hsiao et al. [2005] observed smaller stride lengths when people walked at a virtual height. Similarly, stride length is reduced when people walk while exposed to a postural threat in real scenarios [Brown, Gage, Polych, Sleik, & Winder, 2002]. These results across studies illustrate that real and VR scenarios can cause similar changes in balance tasks, other than quiet stance, when height is manipulated.

1.12 Virtual Reality Effects on Balance

Although it has been demonstrated that VR is a useful tool in posturography, there are a number of limitations in its potential use. Balance can be affected by: a) mechanical factors, such as the weight of the head mounted display (HMD), changes in visual flow, and acuity [Akizuki et al., 2005; Kelly, Rieche, Loomis, & Beall, 2008; Horlings et al., 2009]; b) psychological factors, such as an increased fear [Giphart, Chou, Kim, Bortnyk, & Wagenaar, 2007]; and c) physiological factors, such as sensory re-weighting [Akizuki et al., 2005]. Immersion protocols [Usoh et al., 1999; Meehan, Razzaque, Whitton, & Brooks, 2003; Giphart et al., 2007] and practice trials [Meehan et al., 2002] can be implemented to serve as suitable solutions to overcome direct effects of VR on balance.

1.13 Conclusion of Literature Review

There are many studies that examine the influence of fear and anxiety during static conditions and very few that demonstrate this effect on dynamic postural control. No studies have manipulated fear and anxiety to examine how it may change dynamic postural responses to long surface translations in young adults. Unfortunately, a study involving surface translations and threat manipulation does not appear to be feasible in real world settings, and so there is a need to use VR to mimic the height induced changes in fear and anxiety while still remaining in a safe controlled laboratory environment. In order to normalize posture, the individual needs to have a high ‘sense of presence’ which allows them to feel as though they are within the novel environment and not just viewing it on a computer screen, as well as experience a practice condition in order to escape possible first trial effects. Therefore, there is a need to examine fear and anxiety effects on dynamic postural control through virtual height simulation.

1.14 Purposes and Hypotheses

The main goal of this study is to examine the influence of fear and anxiety on postural reactions to support surface translations during exposure to virtual heights in young adults. It was hypothesized that virtual heights will induce significant psycho-social changes, including increases in fear and anxiety, and decreases in balance confidence and perceived stability. The virtual height was also hypothesized to induce significant changes to the timing and amplitude of electromyography, kinetic and kinematic responses to a support surface translation. Specifically, it was hypothesized that when at a high height:

- muscle activity during balance correcting responses (occurring 100 ms post muscle onset) will be larger in amplitude for those muscles used to regain equilibrium (for example, posterior muscles for backward translations) [Carpenter et al., 2004b]
- greater and earlier EMG activity in the upper arm is thought to be present during balance correcting stages [Carpenter et al., 2004b]
- the range of AP-COM displacement and time to peak COM velocity will be reduced, and will be dependent on the direction of translations [Brown & Frank, 1997; Carpenter et al., 2004b]
- there will be no difference in peak velocity for COM displacements [Brown & Frank, 1997]

- a decrease in amplitude for leg, trunk and arm segment displacements
- COP displacements during first active phase of response will be larger and delayed [Okada et al., 2001]

Chapter 2: Methods

2.1 Subjects

Twenty-one young healthy adults participated in this study aged between 19 and 35 years old. They were recruited from the University of British Columbia and surrounding area. Subjects could not participate in this study if they reported any known neurological or balance deficits, which may affect their postural control. Also, subjects with a known fear of closed spaces (claustrophobia) or with an extreme fear of heights were not allowed to participate. The University of British Columbia Clinical Research Ethics Board (UBC CREB# H06-70316; see Appendix B) approved this study, and all subjects provided written informed consent prior to participation.

2.2 Apparatus

Postural perturbations were supplied by support surface translations in the sagittal plane while in a real environment or immersed in a virtual environment. Participants stood on a force platform (40x60 cm, model BP 400600-1000, Advanced Mechanical Technology Inc., Watertown, MA, USA) mounted to a translating sled (DRS 120-09-176, H2W Technologies Inc., Valencia, CA, USA). Foot placement was marked on to the force platform to ensure a constant foot position across conditions. Subjects were asked to stand 'normal', with their hands at their side, gazing upon a target located at eye level on a real or a virtual wall 4.2 m in front of the subject. A surround platform was placed at the side of the platform (see Figure 1) to allow a spotter to be at the side of the participant during all experimental conditions (including 'height') in case a fall was to occur. The top surface of the platform was 26 cm above the ground; the length of the platform was 60 cm; and the width, including the additional surround, was 123 cm. Subjects were placed in the middle of the platform and asked to refrain from stepping in response to the perturbations to limit the amount of stepping responses [McIlroy & Maki, 1993; Carpenter, Thorstensson, & Cresswell, 2005]. If the subject was to step, the trial was dismissed from further analysis due to the subject's inability to perform the task at hand (react without stepping). The visual dimensions of the real platform were re-created in the virtual environment.

2.3 Virtual Reality Setup

The virtual environment was displayed using a piSight HMD placed on the head of the participant (see Figure 1), with a resolution of 20 pixels/degree in an approximately 150 degree horizontal and 60 degree vertical field of view. This relatively large field of view is smaller than normal vision (200 degrees in the horizontal plane [Gibson, 1979; Werner, 1991; & Barfield, Hendrix, Bjorneseth, & Kaczmarek, 1995] as cited by Arthur, [2000]) however, is larger than binocular vision in the horizontal plane (130 degrees [Herman, 2007]). The virtual scene was custom made using Vizard software (Worldviz, California, USA) by the experimenter, and was used to simulate the scene of a laboratory. During the VR conditions, a black cloth and foam pads were situated around the edges of the HMD to ensure the subject could only see the virtual scene, and therefore could not gain any visual inputs from the real world [Cleworth et al., 2012]. Optotrak cameras were used to determine head position and update the scene at a frequency of 125 Hz, minimizing the time difference between actual head movement and movement of the scene, known as end-to-end latency. With small end-to-end latencies becoming undetectable by participants, presence was potentially increased, and the risk of motion sickness decreased [Akizuki et al., 2005]. The subject also was visually represented by an avatar in the virtual world created and displayed using MotionBuilder (Autodesk, California, USA). The avatar was used to generate a greater ‘sense of presence’, allowing the subject to perceive standing within the virtual lab. When there is no avatar, participants are not able to self-localize within the environment, and may experience a fear of falling [Giphart et al., 2007]. Therefore, a visual reference (an avatar) is important in virtual reality research.

2.4 Procedures

Subject’s experienced16 physical perturbations (8 forward and 8 backward, with a maximum acceleration of 0.6 m/s^2, constant velocity of 0.25 m/s, and total displacement of 0.75 m) while immersed in a virtual environment. Before the first experimental trial, subjects experienced 10 perturbations (5 forward and 5 backward) in the VR low condition as practice trials to familiarize them with the experimental procedures. These practice trials also served to minimize first trial effects. The VR low and VR high conditions where then administered. An ascending order of heights was used in order to maximize the potential threat [Adkin et al., 2000]. The perturbations within each condition were randomized in terms of direction in order to reduce any anticipatory postural changes. For example, if a forward lean is adopted in preparation for a forward perturbation, it would not be beneficial when exposed to a backward perturbation.

Subjects were asked to stand in the middle of the force platform (guided by the experimenter) to ensure equal distance from the edge of the platform in both the anterior and posterior direction. Their feet were spaced relative to their foot length, and then marked. Lower limb kinematics was monitored before each perturbation to ensure anticipatory postural adjustments were not adopted [Carpenter et al., 2004b]. See data analysis for details on monitoring position. If the subject was outside the determined acceptable range the experimenter asked the subject to stand 'normally' (without anticipating or guessing the direction of the perturbation) before testing could commence.

Immersion protocols help increase 'sense of presence'. Therefore, a brief immersion period was used to familiarize the subject within the new environment and help them accommodate to potentially confounding mechanical factors, such as the weight of the HMD, before testing could occur. First, the HMD was positioned in the optimal position for displaying a single, seamless, image and the subject was guided to their standing position up on the force plate. Then, a series of object search and identification exercises were performed to immerse the subject. Shapes and words were placed in the room while subjects were asked to locate and identify the objects. Also, subjects were asked to explore the surface and edges of the platform with their toes to observe the constraints of the platform. The presence of an avatar, as explained earlier, was advantageous in the platform exploration and allowing subjects to visually observe themselves on a 60 x 123 cm platform. The immersion period lasted no less than five minutes to ensure a 'sense of presence' was accomplished.

2.5 Virtual Environment

The virtual environment conditions consisted of 16 perturbations at each height, virtual low (0.26 m) and virtual high (3.2 m) height. Although physical elevation did not change, subjects perceived themselves being raised to an unknown height. In order to enhance the perception of being physically raised to height, auditory and haptic cues were used to simulate platform movement. A recording of a hydraulic lift being raised was played through headphones and the platform on which subjects stood was oscillated in a subtle manor by the experimenter to simulate movement of a lift. These features have been shown to elicit a strong ‘sense of presence’ in prior work under similar conditions [Cleworth et al., 2012].

2.6 Real Environment

The virtual environment was designed to replicate the real environment in which the subjects stood. The subject was not allowed to see the physical space used in this study until testing had been completed. This ensured that subjects were unaware that the translating platform was unable to physically rise to a height.

2.7 Measurements

2.7.1 Data Collection

2.7.1.1 Psycho-Social Questionnaires

Prior to each block of perturbations, subjects were asked to rate how *confident* they were that they could maintain balance and avoid a fall during the balance task. After each block of perturbations subjects completed a series of questions used to rate their perceived anxiety [Carpenter et al., 2004b; Adkin, Frank, Carpenter, & Peysar, 2002; Hauck, Carpenter, & Frank, 2008; Davis et al., 2009], fear [Davis et al., 2009] and stability [Carpenter et al., 2004b; Adkin et al., 2002; Hauck et al., 2008; Davis et al., 2009]. State anxiety was measured using a 16-item questionnaire, modified from Smith, Smoll, and Schutz [1990] [Adkin et al., 2002]. It was used to quantify three elements of state anxiety: somatic anxiety, worry and concentration. Each item uses a 9-point scale ranging from (1) “I don’t feel this at all”, to (5) “I feel this moderately”, and to (9) “I feel this extremely”. Balance confidence, perceived stability and fear were answered using a percentage scale, ranging from 0-100%. 0% confidence/stability refers to “not feeling confident/stable at all” and 100% confidence/stability will refer to “feeling completely confident/stable”, whereas 0% fear referred to “I did not feel fearful at all” and 100% fear referred to “I felt fearful” (see Appendix A-1).

The reliability across these psychological questionnaires (aside from fear) has been verified in young healthy adults [Hauck et al., 2008]. Intra-class correlation coefficients were calculated to determine the test-retest reliability of all measures across three independent balance tasks. Moderate to strong reliability was seen for psychological measures, where the more challenging balance tasks (one legged stance) showed the strongest reliability; state anxiety (r= 0.85), perceived anxiety (r=0.88), and overall confidence (r=0.86) [Hauck et al., 2008].

2.7.1.2 Ratio of Overestimation

Following the high height condition, subjects were asked to estimate the height of the support surface on which they stood. The height at which a person stands is typically overestimated when standing on elevated surfaces, and is overestimate more so when subjects are fearful [Clerkin, Cody, Stefanucci, Proffitt, & Teachman, 2009; Teachman, Stefanucci, Clerkin, Cody, & Proffitt, 2008; Cleworth et al., 2012]. The ratio between the estimated and actual heights, called the ratio of overestimation, was calculated and used to determine if individuals actually perceived themselves to be standing at a height of 3.2 meters.

2.7.1.3 Electromyography (EMG)

EMG data were collected at 3000 Hz, amplified 500x, and bandpass filtered between 10 and 500 Hz (Telemyo 2400R, Noraxon, USA), and sampled at 1000 Hz (Power 1401, Cambridge

Electronic Design, UK). The bias was removed by calculating a mean 300 ms before platform onset and subtracting this mean from the entire signal of that trial. The signal was then offline high pass filtered at 30 Hz using a dual pass Butterworth filter. Finally, the filtered EMG signal was full-wave rectified. Pairs of surface electrodes, spaced 2cm apart, were placed along the muscle bellies of 8 muscles. The muscles were soleus (SOL), medial gastrocnemius (MGAS), tibialis anterior (TA), biceps femoris (BF), rectus femoris (RF), L3 paraspinals (PAR), external obliques (OBL), and middle deltoid (DEL); all recordings were from the right side.

2.7.1.4 Kinematics

Kinematic data were collected using OPTOTRAK (Northern Digital Canada, Waterloo, Canada) motion capture system and sampled at 125 Hz (the maximum amount for the system with the number of markers used). If a marker was out of view for 40 ms or less, a cubic spline interpolation method was used to fill in the missing data. If there was missing data for longer than 40 ms, the trial was omitted from kinematic analysis. Kinematic data were low pass filtered at 5 Hz using a dual pass Butterworth filter. Infrared (IRED) markers were placed on the right side of the body to obtain a 2 dimensional sagittal view of body segments. Individual markers were placed in line with the 5th metatarsal, lateral malleolus (ankle), acromion process of scapula, lateral epicondyle of humerus, styloid process of ulna, mastoid process, and zygomatic arch. Rigid body markers were placed on the lateral aspect of the thigh to allow imaginary markers to represent the greater trochanter (hip) and lateral femoral epicondyle (knee) location. An additional IRED was place on the translating sled to determine antero-posterior movements of the support surface. IRED markers were also placed in line with the styloid process of the left ulna, left lateral malleolus (ankle), and three were situated in a rigid body placed on the anterior aspect of the pelvis in order to allow bilateral control of an avatar in the virtual world.

2.7.1.5 Kinetics

Ground reaction forces and moments were measured using a force platform, collected at 100 Hz and used to calculate COP displacements. The COP data were low pass filtered with a 5 Hz dual pass Butterworth filter.

2.7.2 Data Analysis

To determine the onset of each perturbation, first movement of the marker on the translating sled was used. The displacement of the marker was double differentiated to calculate acceleration. A mean and standard deviation were calculated for 200 ms prior to perturbation onset. Perturbation onset was determined when the first inflection of the acceleration profile was outside the threshold range of mean ± 4 SD. Once the onset was determined, an experimenter manually checked to ensure a correct onset was calculated. Technical difficulties obscured the platform marker and prevented it from being used to determine the perturbation onset for one subject. In this case the ankle marker was used as an alternative to the platform marker. The same criteria were used to determine first movement of this marker. Note, in order to confirm the ankle marker was suitable for perturbation onset calculation in the select participant, ankle and platform onsets were calculated and compared between 2 other participants individually, where similar onsets were observed between the ankle and platform marker.

In order to ensure there were a sufficient number of trials to confidently reflect a given condition, subjects had to have a minimum of three successful trials (excluding the first trial) in both the virtual low and high conditions to be included in subsequent analysis [Tokuno, Cresswell, Thorstensson, & Carpenter, 2010]. Trials were excluded if the subject stepped in response to the perturbation, or if there was more than 40 ms of consecutive missing marker data (only kinematic analysis was not conducted in this case, all other measures were still calculated for these trials).

2.7.2.1 Psycho-Social Questionnaires

Fear, balance confidence and perceived stability raw scores were used in the analysis. Total anxiety was calculated by summing the two scores of somatic anxiety (6 items) and worry (4 items) [Geh, Beauchamp, Crocker & Carpenter, 2011].

2.7.2.2 Ratio of Overestimation

The ratio of overestimation was calculated offline. The estimated height was divided by the actual virtual height (3.2 meters) to determine the ratio. A ratio above 1 reflected an overestimation, less than 1 reflected an underestimation, and 1 reflected a correct estimation.

2.7.2.3 Electromyography (EMG)

After determining platform onset, the point of first inflection above a threshold of 2 SD's about the mean (calculated from 500 ms prior to perturbation onset), and staying above that threshold for longer than 30 ms determined EMG onset. This onset was then visually confirmed, and if needed, manually corrected by an experimenter. The same experimenter examined all EMG latencies in order to keep consistent analysis across conditions. No trials were accepted for analysis where the EMG onset preceded 30 ms or occurred 400 ms or more after perturbation onset.

EMG response amplitudes, calculated from the integrated area of the rectified signal (iEMG), were calculated form EMG onset to 100 ms after onset [Campbell, 2012; Carpenter, Tokuno, Thorstensson, & Cresswell, 2008]. The EMG amplitude100 ms prior to EMG onset was subtracted from the iEMG response to get the true amplitude of the signal. In trials where EMG onsets were not observed, amplitude level was set to zero (meaning EMG activity before and after onset were equal for that trial).

2.7.2.4 Kinematics

Total body COM calculations used 2 dimensional filtered coordinates defining the body segments foot, shank, thigh and HAT (head, arms, and trunk), which were then used in conjunction with anthropometric data [Winter, 2005] to determine the horizontal coordinates of whole body COM. A four-segment model was used to determine COM:

$$\text{COM} = (m1_{(x1)}+m2_{(x2)}+m3_{(x3)}+m4_{(x4)}) / (m1+m2+m3+m4) \text{ [Brown \& Frank, 1997]}$$

where m is the mass of the segment and x is the horizontal location of the segments COM.

The measures of interest included stepping frequency (to allow for exclusion of trials when stepping occurs), COM peak displacement, time to peak COM displacement, COM peak velocity, time to peak COM velocity, and angular displacement of trunk (defined as greater trochanter to acromion process), upper leg/thigh (defined as lateral epicondyle of femur to greater trochanter), lower leg/shank (defined as lateral malleolus to lateral epicondyle of femur), and arm (acromion process to lateral epicondyle of humerus). The trunk, thigh and shank segment angular displacements were calculated in the sagittal plane with respect to absolute horizontal, where a rotation in the posterior direction (away from the toes) resulted in a negative value (see Figure 2). Arm segments were calculated relative to the trunk, and were characterized by two actions, pitch (flexion/extension) and roll (abduction/adduction) directed movements.

Prior to perturbation, ankle angle was monitored in real time to ensure a consistent starting posture. Foot (toe to ankle marker) and shank coordinates were used to calculate a real time ankle angle in the sagittal plane. A mean angle was calculated after a quiet stance period prior to any experimental conditions. Perturbations were not administered unless the ankle angle was within 2 SD's of the mean. To note, subjects may have swayed outside the described window prior to initiation of platform translation as there was 1 to 1.5 sec between experimenter's last observation of subject's stance and initiation of perturbation.

2.7.2.5 Kinetics

Peak COP displacement was used to determine the maximum excursion of the individual's COP. Both time to peak and peak amplitude were analyzed. Note, one subject was excluded from the COP analysis as there were technical difficulties with the force plate.

2.7.3 Statistical Analyses

Paired-samples t-tests were used to compare virtual low to virtual high conditions. These contrasts were performed for all measures including balance confidence, perceived stability, fear, and anxiety, peak COP displacement, time to peak COP displacement, COM displacement, time to COM displacement, COM peak velocity, time to peak COM velocity, segment displacements, and EMG latencies and amplitudes. Statistical significance was set at $p \leq 0.05$. An independent sample t-test was used to determine if the ratio of overestimation at height was different from an estimation equal to 1 (actual height and perceived height are the same).

Chapter 3: Results

3.1 Psycho-Social Aspects of Fear and Anxiety

The results showed that when participants stood at height in the virtual environment there was a significant change in fear, anxiety, and balance confidence. At height, participants reported a significant increase in state anxiety ($t(20)=2.729$, $p=0.013$), and fear ($t(20)=3.342$, $p=0.003$), and a significant decrease in balance confidence ($t(20)=3.347$, $p=0.003$). There were no significant differences in participants' perceived stability; however, overall stability levels were similar to those previously reported in more threatening quiet stance paradigms (Cleworth et al., 2012). Finally, subjects overestimated the height at which they virtually stood, as participant's ratio of overestimation (mean = 1.6 ± 0.1) significantly exceeding a correctly estimated height of 1 ($t(20)=3.792$, $p=0.001$).

3.2 Backward Perturbation Results

Stepping responses occurred in a total of 1 trial in the low condition, and 1 trial in the high condition across all 21 subjects. Also, the cubic spline method was applied to 0% of trials in the low condition, and 3.8% of trials in the high condition across all 21 subjects.

3.2.1 Total Body Horizontal Centre of Mass

In general, a posterior translation caused the horizontal COM to be displaced forward, opposite to the direction of the platform. In the low condition, the horizontal COM reached a peak displacement of 40.7 ± 1.9 mm at 656.7 ± 19.0 ms after perturbation onset, and had a peak velocity of 116.8 ± 4.2 mm/sec occurring at 337.5 ± 13.9 ms after perturbation onset. At height, peak displacement and peak velocity were similar to those observed at the low height, however the COM displacement had a trend to peak significantly earlier, 24.2 ± 13.9 ms earlier in comparison to the low height ($t(20)=1.742$, $p=0.097$).

3.2.2 Segment Displacements

In general, in both conditions, backward translations initially caused the shank and thigh to rotate forward, causing dorsiflexion of the ankle, while the trunk remained relatively erect (approximately 90 degrees from horizontal) (see Figure 4). Following this, the trunk began to rotate forward causing flexion of the hip.

As described earlier, the arm movements are characterized in two directions, pitch and roll. In the low condition the shoulder is flexed, on average, 15.1 ± 3.3 degrees following backward perturbations. The shoulder pitch displacements peak in a range from approximately 1 to 56 degrees (between participants) then reverses to a slightly flexed shoulder position (see Figure 5 for mean displacements). In the roll direction, the arm is abducted 9.7 ± 1.8 degrees following a backward perturbation. This peak displacement was followed by movement towards a less abducted position (similar to the profile of arm flexion, see Figure 5). Manipulation of virtual height elicited changes in arm movements. The arm flexed and abducted with a greater magnitude at height compared to the low condition ($t(20)=2.669$, $p=0.015$, and $t(20)=2.390$, $p=0.027$, respectively). When participants are placed in the virtual height condition, the shoulder flexes 3.8 ± 1.4 degrees more, and abducts 3.2 ± 1.4 degrees more compared to the low condition.

Following the onset of the perturbation, the shank initially rotated forward with amplitudes of 2.7 ± 0.3 degrees in the low condition and 2.9 ± 0.4 degrees in the high condition (not statistically different). This component holds both the passive component caused by the platform movement, as well as the active component, which is reflected by the reversal/stopping of the lower limb rotating forward. An average trace shows the shank then rotates back to initial position. However, upon further observation, the position of this later component is widespread across participants. Some participants remain in a dorsiflexed position (at approximately 5 degrees); some over shoot their initial position ending in an absolute plantarflexed position (approximately -7 degrees); and some plantarflex back to initial position (approximately 0 degrees).

The backward movement of the platform caused an initial forward rotation of the thigh. This rotation was likely due to the knee angle remaining unchanged while the subject rotated around the ankle, causing the thigh relative to horizontal to have a change in position in the forward direction and therefore an angular movement resembling extension. The thigh then rotates backwards, and continues to rotate, on average, until the end of the analyzed period (1.2 sec), reaching an amplitude of 8.8 ± 2.0 degrees in the low condition and 9.4 ± 2.2 degrees in the high condition (not statistically different).

At approximately 1 sec, the forward rotation of the trunk reaches a single peak amplitude of 21.7 ± 2.9 degrees in the low condition and 21.0 ± 2.5 degrees in the high condition, on average (see Figure 4). No significant differences were observed for trunk angular displacements between heights.

3.2.3 Kinetics (Ground Reaction Forces and COP)

The COP displaced in the anterior direction as a backward translation induces a body displacement opposite to the perturbation. In the low condition, the COP was displaced 112.5 ±

1.8 mm, peaking at 400.5 ± 12.0 ms after the onset of the perturbation. Significant changes in the trajectory of COP displacements were observed with an increase in postural threat as evidenced by a significant decrease in time to peak COP and a significant increase in peak COP amplitude at height. Specifically, when raised to height, participants reached peak COP 15.8 ± 7.2 ms earlier ($t(19)=2.304$, $p=0.033$) and displaced their COP 3.4 ± 1.5 mm more ($t(19)=2.178$, $p=0.042$) in comparison to the low condition.

3.2.4 Leg and Trunk Muscle Latencies and Amplitudes

For muscles on the posterior side of the body, typically used in balance recovery for a backward perturbation [Rothwell, 1994], muscle activation occurred in SOL and MGAS at 164.7 ± 10.9 ms and 152.1 ± 8.4 ms, respectively, followed by biceps femoris at 237.6 ± 18.9 ms and then PAR at 270.5 ± 11.0 ms. This distal to proximal activation pattern was elicited in combination with early OBL (when onset was found, n = 6) at 177.1 ± 8.1 ms. Aside from the OBL, other analyzed ventral muscles showed similar onset latencies at 211.3 ± 14.1 and 210.2 ± 10.7 ms (for TA and RF, respectively). In general, all muscles showed earlier onset latencies (however not all are significant, see Table 1) in the high compared to low condition. When participants were raised to height, the SOL was activated 18.8 ± 8.1 ms earlier ($t(9)=2.322$, $p=0.045$), and the medial gastrocnemius showed a trend toward earlier activation as it was activated 15.3 ± 8.3 ms ($t(17)=1.839$, $p=0.083$) earlier compared to low height condition. The earlier activation patterns were accompanied by larger EMG activity; however no significant changes were reported for EMG amplitude in trunk and leg muscles. The amplitude of the MGAS had a trend towards significance ($t(20)=1.997$, $p=0.060$), where it responded with larger amplitude at height compared to low. All muscles, aside from the OBL, have mean amplitudes larger at height. The SOL, BF and PAR increase activity more so than the TA. The external oblique does not appear to change activity between conditions.

3.2.5 Arm Muscle Latency and Amplitude

In the low threat condition, the DEL was activated 176.0 ± 11.0 ms after perturbation. This onset is similar to what is observed in the distal leg muscles and the oblique muscle. At height, however, the deltoid muscle activated 12.2 ± 7.5 ms earlier (not significantly) at height ($t(15)=1.632$, $p=0.123$). DEL amplitude was significantly larger at height ($t(20)=2.181$, $p=0.041$) as it increased activity by 8.1 ± 3.7 µV.s (Figure 6). It is interesting to note that the larger activity of the upper arm corresponds with the biomechanical changes observed (larger shoulder abduction and flexion).

3.3 Forward Perturbation Results

Due to technical difficulties, and participants' inability to refrain from stepping during a forward perturbation (and therefore failing to meet the task requirements), there is a significantly smaller sample size (9 out of 21) for forward perturbations. Stepping responses occurred in a total of 56 trials in the low condition, and 60 trials in the high condition across 9 subjects. Also, the cubic spline was applied to 26.5% of trials in the low condition, and 15.6% of trials in the high condition across 9 subjects. A lack of significant findings may be due to sample size issues.

3.3.1 Total Body Horizontal Centre of Mass

Forward perturbations elicit responses occurring in the posterior direction. In general, the horizontal displacement of the COM peaked at 710.7 ± 39.1 ms after perturbation, reaching a maximum displacement of 59.9 ± 6.4 mm (to note, this is greater than what is seen in backward perturbations) in the low condition. Also, the first peak velocity occurred at 252.8 ± 26.3 ms after perturbation onset, with a peak of 166.3 ± 12.0 mm/sec. At height, the COM reaches peak displacement 29.6 ± 19.8 ms later, and has a 6.1 ± 4.4 mm/sec smaller peak velocity. These were not statistically different ($t(8)=1.490$, $p=0.174$, and $t(8)=1.392$, $p=0.201$, respectively). Similar to the backward perturbations, the results showed that the horizontal displacements of the COM in response to the perturbation were similar between height conditions.

3.3.2 Segment Displacements

In general, leg and trunk displacements are characterized by a distal to proximal movement sequence. Similar to the backwards perturbations, the shank passive rotation is the first to appear, followed by backward rotation of the thigh and passive rotation of the trunk (see Figure 4). The three segments appear to reach their peaks at roughly the same time, 0.8 sec after perturbation.

For the shank, due to the forward translation of the platform, there is a passive plantar flexion movement at the ankle joint, which is less than approximately 2 degrees. The shank then rotates in the anterior direction giving a dorsiflexion movement of the ankle. This movement peaks at 6.8 ± 2.1 degrees in the low condition and 7.6 ± 2.0 degrees in the high condition. Subjects range from approximately 1 to 18 degrees dorsiflexion across heights. The thigh has a single peak, unidirectional movement in response to a forward perturbation. The thigh rotates in a posterior direction, indicating knee flexion. This flexion has an average peak of 10.2 ± 1.1 degrees in the low condition, and 10.4 ± 1.2 degrees in the high condition. The trunk has an initial passive component similar to the shank. The forward acceleration of the platform causes flexion of the trunk. Due to the limited segments observed, it is postulated early trunk flexion may be indicative of pelvic rotation, while the upper trunk remains still [Carpenter et al., 2005]. The trunk then moves into an extension position, which has a peak displacement of 10.0 ± 2.7 and 11.0 ± 3.0 degrees in the low and high conditions, respectively.

Arm angular displacements had a peak flexion of 16.6 ± 10.3 degrees and a peak abduction of 22.0 ± 6.7 degrees, on average, in the low condition. When participants are placed in the virtual height condition, the shoulder (however not significant) flexes 3.8 ± 3.2 degrees less, and abducts 2.5 ± 1.9 degrees less ($t(8)=1.199$, $p=0.265$, and $t(8)=1.277$, $p=0.237$, respectively). Arm movements are generally larger in response to forward perturbations compared to backward; however, changes with respect to height are in the opposite direction.

3.3.3 Kinetics (Ground Reaction Forces and COP)

COP excursions in response to forward perturbations, on average, had peak displacements of 96.1 ± 2.0 mm in the low condition and 96.2 ± 2.1 mm in the high condition (not statistically different). The amplitude of displacement is less than what is seen in backward perturbations. Following perturbation onset, COP peak displacement occurs at 382.1 ± 16.2 ms in the low condition and 383.9 ± 35.9 ms in the high condition (not statistically different).

3.3.4 Leg and Trunk Muscle Latencies and Amplitudes

Overall, muscle activity appears to occur earlier in the forward perturbations than they do in the backward perturbation, while still holding a distal to proximal activation pattern. In the low condition, the TA activates 111.4 ± 6.1 ms after perturbation onset. Following the lower leg muscle, the RF and BF activate at 147.9 ± 9.0 ms and 168.5 ± 19.3 ms, respectively, and the

PAR activates 173.3 ± 11.4 ms after perturbation onset. The OBL activates at 207.2 ± 15.8 ms after perturbation, which is used to erect the rotated trunk; this trunk muscle is much earlier than the PAR used to erect the trunk in the backward perturbations. The SOL and MGAS are not active in response to forward perturbations and so no analysis was completed. When participants were placed at height, the RF was activated 11.7 ± 8.2 ms later ($t(13)=2.004$, $p=0.066$) compared to the onsets observed at the low height. No notable differences were observed in EMG amplitude across height conditions for leg and trunk muscles in response to forward perturbations.

3.3.5 Arm Muscle Latency and Amplitude

In the low height condition, the deltoid has an onset of 161.5 ± 11.8 ms after perturbation onset, which is earlier compared to backward conditions. The DEL activates in line with muscles of the upper leg in the forward conditions, contrary to what is seen in backwards perturbations. This activation, again, breaks the distal to proximal activation pattern typically seen in response to external perturbations. In the high height condition, the deltoid muscle activates 11.7 ± 8.2 ms earlier (similar to what is seen in response to backward perturbations), however this is not significant ($t(12)=1.420$, $p=0.181$). The amplitude of the deltoid muscle at height is, however, smaller than at the low height (which falls in line with the smaller movements of the arm when at height). Smaller amplitude at height is opposite to the observed height effect in response to backwards perturbations.

3.4 Pre-stimulus Posture

When participants were standing still prior to perturbations, there were significant changes observed between heights. Background activity of the leg changed at height prior to backward perturbations. Specifically, the TA and DEL significantly increased ($t(20)=2.785$, $p=0.011$ and $t(20)=2.636$, $p=0.016$, respectively); the RF increased (however, not significantly, $t(20)=1.721$, $p=0.101$); and the SOL and BF decreased (however, not significantly, $t(20)=$, $p=0.194$ and $t(20)=1.425$, $p=0.170$, respectively) at height. Individual segment angles did not change between conditions.

Chapter 4: Discussion

The aim of this thesis was to examine the influence of fear and anxiety on postural reactions to surface translations during exposure to virtual heights in young adults. This was addressed by using a novel virtual environment, where individuals were exposed to physical perturbations while at a low and high virtual height; that is, subjects were virtually placed at ground level and at an elevated height while physical elevation did not occur. The results of the current study confirmed that virtual heights can induce significant psycho-social changes, and neuromuscular and kinetic changes in response to dynamic balance perturbations, particularly those in the backward direction.

4.1 General Findings

Confirming my hypothesis, virtual heights were able to induce significant changes in psycho-social state. When participants were standing at virtual height, increases in fear and anxiety, and decreases in balance confidence were observed (see Figure 7). This is consistent with previous studies that have shown similar findings when individuals stand at real [Davis et al., 2009; Adkin et al., 2002] or virtual [Cleworth et al., 2012] heights of similar magnitude. Participants also overestimated the height at which they stood (see Figure 8), which is also consistent with previous work [Cleworth et al., 2012; Teachman et al., 2008; Clerkin et al., 2009]. An average ratio of 1.55 reported in the current study is similar to Clerkin et al. [2009], who reported overestimation ratios of 1.3 to 1.5 from participants standing at real height, and similar to Cleworth et al. [2012] who reported ratios over 1.5 when participants stood at virtual heights. Therefore, it is evident the virtual elevation in the current study was successful at inducing emotional change, thus placing subjects in a perceptually threatening environment, replicating real height perceptions without physical elevation.

It was originally hypothesized that virtual heights would induce significant changes in EMG amplitude, specifically in the muscles used to regain equilibrium, when responding to a perturbation independent of direction. In contrast, no changes in the onset latencies of balance correcting or stretch reflexes were expected based on real height studies [Carpenter et al., 2004b]. Contrary to the first hypothesis, changes in EMG amplitude with increased height were found to be dependent on the direction of the perturbation. At the low height, EMG activity was observed in a distal to proximal activation pattern (lower leg, thigh, and then trunk muscles), similar to what has been reported in studies with large amplitude translations [Tokuno et al. 2010]. When raised to height, backward perturbations elicited larger EMG responses (however only the DEL was significantly different). Specifically, SOL, MGAS, BF, DEL and PAR increased activity more so than TA and OBL, as expected, as these are responsible for bringing the body back to equilibrium [Rothwell, 1994]. However, no significant increases were observed for forward directed perturbations. In fact, DEL decreased activity (not significantly) at height in response to forward perturbations. The results did not confirm my hypothesis that onset latencies would not change with height given that MGAS and SOL were activated 15 and 19 ms earlier at height compared to low, respectively, and all other muscles activated at least 10 ms earlier at height (except TA). This effect of height on latency was also direction specific as forward perturbations did not elicit any changes with height. However, RF activity was slightly later at height (not significant). This is contradictory to previous reports, as increased amplitude of muscle activity was associated with increased postural threat independent of direction [Carpenter et al., 2004b] and changes in EMG onsets did not occur. In the current study where EMG onsets were affected, stretch and balance responses occur in the same muscle. Previous work where onsets were not affected was in response to rotations where these two responses occur in separate and antagonistic muscles.

Previous work has demonstrated earlier and larger activity in arm muscles in response to perturbations during more threatening scenarios [Carpenter et al., 2004b]. Results from this thesis showed similar changes in arm muscle activation when threat was manipulated virtually. Participants increased the displacement of the arm, in the pitch and roll direction, as well as increased activity of the DEL in response to backward perturbations. Arm movements can be utilized to keep humans from falling, as they can help stabilize the COM over the BOS, and can serve as protective strategies, as they can shield the head from an impact [Maki & McIlroy, 1997]. Earlier and larger responses seen at height have been thought to be startle-like reflexes. However, it is unlikely responses in the current study are due to startle as startle-like reflexes occur much earlier (164 ms compared to 90 ms [Valls-Solé, Rothwell, Goulart, Cossu, & Muñoz, 1999]). More likely is the notion that arm responses may be protective and be executed earlier when a threat is detected.

In addition to changes in EMG, the effect of fear was expected to also impact the kinematic and kinetic outcomes of a postural response. Specifically, I hypothesized an increase in fear would a) induce significantly larger yet later COP peak displacements [Okada et al., 2001]; b) reduce the magnitude of AP-COM displacements, and leg and trunk segment displacements; and c) reduce the time to peak COM velocity [Brown & Frank, 1997; Carpenter et al., 2004b]. When participants were perturbed, ground reaction forces under the feet were used to displace the COP ahead of the COM. Irrespective of direction, the COP is used to accommodate the movement of the COM [Henry, Fung, & Horak, 1998]. In the current study, the COP translated further in response to backwards (112.5 ± 1.76 mm) perturbations compared to forward (96.1 ± 1.99 mm), in accordance with previous work [Henry et al., 1998]. Contrary to hypothesis (a), I observed larger and earlier COP peak displacements in response to backward perturbations when participants stood at height (see Figure 9). In response to forward perturbations, there were no changes between heights. Novel findings in the current study may be due to a change in; i) the duration of the interval between the acceleration and deceleration phase of the perturbation, and ii) analysis of different aspects of the perturbation, in comparison to previous work. Both reactive and anticipatory mechanisms are influenced by short duration intervals, and can manipulate the reactions to translations [Carpenter et al., 2005]. Furthermore, previous work [Okada et al., 2001] observed changes in elderly subjects, whereas I observed changes in young adults. Kinematic outcomes in response to backward or forward perturbations were not affected by increased fear and anxiety. The time to peak COM displacement during the backwards perturbations was slightly earlier at height, similar to previously reported earlier peak COM velocity during trunk perturbations at real height [Brown & Frank, 1997], however this was not significant. Changes in kinematics have been reported when individuals stand at real heights and are perturbed. The results of the current study may allude to the implications an added weight (HMD and knapsack) has on kinematic responses, or to the potentially variable responses seen with unrestricted long perturbations, as opposed to more controlled rotations and upper body perturbations previously used [Carpenter et al., 2004b; Brown & Frank, 1997]. The null result of fear on kinematic outcomes is not surprising as the muscle activity that controls the trunk and upper leg did not significantly change. The profound changes observed occurred in lower leg muscle activity and COP excursions, and arm responses (muscle activity and segment displacements).

4.2 Influence of Virtual Heights on Pre-stimulus Posture

Two issues related to fear-induced changes to pre-stimulus posture that might confound the observed changes in dynamic responses arise when placing young healthy individuals at height. First, people lean backward away from the edge, which is reflected in a posterior shift of mean COM and COP positions [Carpenter et al., 2001]. Maximally leaning in the opposite direction from platform directed movement (such as leaning forward before a backward directed translation), will change EMG responses of the lower leg (dependent on direction), increase hip movement, and increase proximal muscle activity [Horak & Moore, 1993]. A relative shift in COP position opposite to the direction of the perturbation has also been shown to reduce latencies of the antagonist muscles, and increase activity in the majority of muscles [Tokuno, Carpenter, Thorstensson, & Cresswell, 2006]. In the current study, no changes were seen in joint angular position prior to perturbations. The lack of change in pre-stimulus posture was expected given the steps taken in the methods designed to minimize leaning (by implementing a bi-directional threat, monitoring pre-stimulus ankle position, and using unpredictable perturbation onset and direction). It was disadvantageous for participants in the current study to adopt a leaning strategy, as leaning in one direction would not be beneficial if the next perturbation was in the opposite direction. As such, it is unlikely that pre-stimulus leaning can explain the changes to postural responses seen in this study.

The second issue is that changes in background muscle activity are known to occur when standing in a threatening environment. There was a change in muscle activity prior to perturbations when participants stood at height in this study (see Figure 10). Similar to previous studies that used real heights [Carpenter et al., 2001; Sibley et al., 2007], participants increased activity in TA and RF, while decreasing activity in SOL and BF. These changes are not necessarily dependent on maintaining a backward lean, as increased TA activity has also been reported in anxious subjects, who lean forward when performing secondary tasks [Maki & McIlroy, 1996]. The observed relative change in muscle activity is hypothesized to coincide with a stiffening strategy [Carpenter et al., 2001]. However, whether a stiffening strategy would be beneficial or harmful to dynamic stability has been debated and is still an unresolved issue [Bloem, Allum, Carpenter, Verschuuren, & Honegger, 2002; Allum, Carpenter, Honegger, Adkin, & Bloem, 2002]. It has been argued that increased stiffness may lead to larger kinematic responses to an external perturbation, which, in turn, requires larger EMG responses [Allum et al., 2002]. However, there was no evidence to suggest that stiffness led to different kinematic responses to the same perturbation in this study, as there were no significant changes in kinematic responses observed following perturbations in the high compared to low condition.

In addition, increased tonic activity is known to increase the amplitude of short and medium latency stretch reflexes, and has been implicated in gain increases of longer latency balance correcting responses used in regaining equilibrium after experiencing external perturbations [Sinha & Maki, 1995; Horak and Moore, 1993]. Background activity did not influence balance correcting responses in this study because response activity was increased regardless of whether background activity was increased or decreased in a particular muscle.

Therefore, it is unlikely that the pre-stimulus changes in background activity with height were responsible for the observed changes in postural responses.

4.3 Neural Mechanisms

Recent evidence has suggested changes in neuro-mechanical processes may be responsible for the modulation of postural responses. The following section will examine some of the proposed neural mechanisms that might contribute to threat-related changes in terms of earlier onsets and increased amplitude of postural responses.

4.3.1 Afferent Mechanisms Influenced by Virtual Height

When proprioception is reduced from the lower leg, as seen in people with diabetic neuropathies, translational perturbations elicit distal to proximal activation patterns with EMG responses occurring 20-30 ms later compared to healthy controls [Inglis, Horak, Shupert, & Jones-Rycewicz, 1994]. Furthermore, purely rotational perturbations elicit delayed and diminished balance correcting responses in a functionally stabilizing muscle (TA for toes up rotations) [Bloem, Allum, Carpenter, & Honegger, 2000]. Also, patients with diabetic polyneuropathies have reduced EMG responses in functionally stabilizing muscles in response to rotational perturbations (either pure rotation or with a translational component to enhance ankle movement) [Bloem et al., 2000]. Therefore, results from the current study (earlier onset and increased activity of lower leg muscles) would suggest an opposite effect; postural threat increases proprioceptive sensitivity and gain. Spindle sensitivity has been shown to be facilitated by increased levels of fear and anxiety [Horslen, 2010; Davis et al., 2011] and has the potential to increase afferent feedback to supra-spinal mechanisms used in postural responses, although not via direct cortical pathways [Davis et al., 2011].

Individuals with reduced vestibular information, as observed in patients with vestibular impairments (for example, acute unilateral vestibular loss), have been shown to have similar onsets and latencies of postural responses to balance perturbations compared to normal subjects [Allum & Pfaltz, 1985; Allum, Honegger, & Schicks, 1994; Keshner, Allum, & Pfaltz, 1987; Runge, Shupert, Horak, & Zajac, 1998]. This preserved response, however, was accompanied with decreases in amplitude of muscle activity in balance correcting responses [Keshner et al., 1987; Allum et al., 1994], thus implying that vestibular gain influences balance correcting responses. In the current study, increased amplitude of activity in the muscles used to equilibrate the body after a destabilizing perturbation when at height compared to the low height might suggest tuning of the vestibular system to increase the gain. This effect has been postulated in previous work that has observed changes to vestibular function or postural responses [Yardley et al., 1995, Carpenter et al., 2004b].

Neural links between areas of the brain associated with posture and areas associated with mood and emotion have been documented in anatomical studies [Balaban, 1996; Scheuerger & Balaban, 1999], and have also been postulated in behavioural studies [Yardley et al., 1995, Carpenter et al., 2004b]. Reciprocal connections exist between the vestibular network and sub-cortical structures, including the parabrachial nucleus, locus coeruleus, and raphe nuclei [Balaban, 1996; Scheuerger & Balaban, 1999; Halberstadt & Balaban, 2003]. The sub-cortical structures have been shown to be active in times of increased arousal, anxiety and/or danger, and have the potential to influence the gain of vestibular information during moments of postural threat [Balaban & Thayer, 2001; Balaban, 2002; Carpenter et al., 2004b].

4.3.2 Efferent Mechanisms Influenced by Virtual Height

Postural responses, occurring later than stretch reflexes, are evidently facilitated by supra-spinal mechanisms as observed by activation in cortex and cortico-spinal tracts during or preceding moments of postural responses [Jacobs & Horak, 2007; Taube et al., 2006], and by modified muscle activity in decerebrate cats when responding to destabilizing stimuli [Honeycutt, Gottschall, & Nichols, 2009]. In addition, anticipatory postural changes and attenuated postural responses have been associated with processes that require supra-spinal pathways, such as cognition, attention and experience [Horak, Diener, & Nashner, 1989; Redfern, Jennings, Martin, & Furman, 2001; Maki & McIlroy, 2007]. Furthermore, when individuals are exposed to startling stimuli in environments capable of inducing destabilizing perturbations, components of postural responses are evoked, even in the absence of the perturbation [Campbell, 2012], thus illustrating triggered postural responses can be initiated in the absence of ascending sensory information gained after a perturbation. Therefore, modulation of postural responses due to increase fear and anxiety may not solely be due to changes in the processing requirements of sensory information, but also may be due to the modification of these supra-spinal pathways used in postural responses. Cortico-spinal excitability, examined through transcranial magnetic stimulation (TMS), increases with negative emotion [Hajcak et al., 2007; Coelho, Lipp, Mainovic, Wallis, & Riek, 2010; Oathes, Bruce, & Nitschke, 2007; Schutter, Hofmen, & van Honk, 2008]. Specifically, motor evoked potentials produced by TMS are facilitated when observing unpleasant pictures [Hajcak et al., 2007; Coelho et al., 2010], fearful faces [Schutter, et al., 2008], and in moments of increased worry [Oathes et al., 2007]. As a result, when participants are placed at height, virtual or real, the higher level neural pathways are modulated by areas associated with fear and anxiety, and thus may adjust the motor output and cortico-spinal tracts associated with postural responses to the impending threat.

4.4 Confounding Influences

The first limitation is the direct effect VR may have on balance itself. Posture has been shown to be less stable when placed in a virtual environment compared to a real environment [Akizuki et al., 2005; Kelly et al., 2008; Horlings et al., 2009]. Differences were found to be independent of mechanical factors, such as the weight of the HMD and field of view, when controlled for [Akizuki et al., 2005; Kelly et al., 2008]. Alternatively, differences observed in postural control between real and virtual environments may be explained by sensory re-weighting. It has been proposed that visual information becomes unreliable, or even inappropriate, when immersed in a virtual environment, and sensory re-weighting occurs such that the relative contributions of vestibular and somatosensory increase [Akizuki et al., 2005]. Other possible contributions to VR instability may involve psychological components, for example a fear of falling [Giphart et al., 2007]. A lack of visual representation of oneself within the virtual environment does not allow for self-localization and, therefore, can be daunting.

A number of possible solutions were implemented to control for VR effects on balance. The first was implementation of an immersion period, allowing subjects to become as comfortable as possible with the novel paradigm/environment. When immersed, participants become accustomed to the mechanical factors of the VR system (for example, the weight on the HMD), which should result in a reduction in potential influences of VR explained above. The time frame of this period is unknown but appears to be a large component in creating a ‘sense of presence’ as VR experiments usually contain training conditions [Usoh et al., 1999; Meehan et al., 2003; Giphart et al., 2007]. A second component was implementing practice conditions in order to escape any first trial effects associated with VR exposure. When placed in a novel environment and asked to perform complex tasks, individuals may react in a different manner than what would typically be seen. Practice conditions are used to orient the subject with the paradigm in order to escape these first trial effects, and to further habituate the participant to the VR system. A third component was implementing an avatar into the virtual world, allowing visual representation and self-localization of the participant inside the environment, thus reducing previously explained unwarranted psychological effects. The combination of immersion protocols, practice trials and avatar implementation potentially reduce the risk of participants experiencing changes in postural control compared to normal conditions.

A second limitation of this thesis was in the limited response strategies permitted to be used by the participant, as they were instructed not to step in response to the perturbation. If they stepped, it was considered a failure to achieve the task requirements. This instructional set has been shown to alter participants reactions compared to when only told to remain upright and not given instructions on how to do so [McIlroy & Maki, 1993]. Stepping has been described as an important protective strategy [Maki & McIlroy, 1996] and may provide further evidence of height related changes to balance correcting responses. However, participants were advised against stepping in the current study for safety reasons, ensuring participants did not step off the small support surface, and to keep conditions consistent with previous work.

A third limitation in this study is that there was no direct measurement of physiological arousal. Many studies have demonstrated an increase in EDA when standing at height, real or virtual [Cleworth et al., 2012; Simeonov et al., 2005; Meehan et al., 2002; Horslen, 2010]. Also, when standing on elevated support surfaces, participants' self-reported fear and anxiety increase, and balance confidence decreases [Horslen, 2010; Davis et al., 2009; Cleworth et al., 2012]. Therefore, since the observed changes in psychological components have been demonstrated to change along with increased levels of EDA when standing on real and virtual elevated platforms [Cleworth et al., 2012; Davis et al., 2009], it is likely, in the current study, physiological arousal was increased with height.

4.5 Clinical Implications

In a clinical setting, balance disorders are often described in terms of physiological deficits and age related decline. Due to the described changes caused by fear (both in the current and previous studies), balance impairments, seen in populations with increased fall risk, may be incorrectly attributed to these physiological disorders. Populations who are at risk for fear related balance deficits include older adults, and patients with Parkinson's disease or vestibular loss. In contrast to the observed changes in the current study, older adults produce later and larger EMG responses when exposed to a long translational perturbation [Tokuno et al., 2010] and produce later and diminished EMG activity in the lower leg and arm muscles in response to rotational perturbations [Allum et al., 2002]. Individuals with vestibular loss produce reduced amplitude balance correcting responses in lower leg muscles in response to a rotational perturbation [Allum et al., 1994]. Patients with Parkinson's disease however, appear to elicit responses which correspond to the observed changes in young healthy adults when placed in a threatening environment. Increased background activity prior to perturbations, increased amplitude of balance correcting responses and significantly earlier arm muscle activity in response to a perturbation were observed in patients with Parkinson's disease [Carpenter et al., 2004a]. Similar observations between individuals with Parkinson's disease and young adults placed in a fear evoking environment, as in the current study, demonstrate the need to consider fear of falling as a possible contributor to balance deficits in clinical situations. The use of dynamic posturography in clinical testing can assist practitioners in determining the underlying problems associated with the balance impairment. For example, testing those with a fear of falling and those without, when analyzing within populations with balance deficits.

The use of novel, situation specific or population specific environments in rehabilitation can also be used to assist health practitioners in reducing known balance deficits. Virtual environments have been used in exposure therapy techniques to reduce fears [Oskam, 2005]. By exposing patients to fear related events, they can habituate emotional responses and overcome fear related deficits. In addition, the implementation of VR into rehabilitation settings will allow health practitioners the opportunity to expand practices outside the normal clinical setting, creating daily occurring scenarios where patients can be observed during realistic events (as realistic as a virtual environment can be).

4.6 Future Research

4.6.1 VR Application in Posturography

Further work still needs to be completed on how to normalize and/or test balance in a virtual world. To my knowledge, there is no work demonstrating the effects of VR, via HMD, on support surface translations. The direct effect of VR has been quantified in quiet stance [Akizuki et al., 2005; Kelly et al., 2008; Horlings et al., 2009] but it is difficult to generalize these findings to dynamic stance. One way to reduce the risk of VR effects is to validate criteria for immersion protocols. Since 'sense of presence' has been described as the most important factor in VR in behavioural research [Hodges et al., 1994], immersion protocols need to be created which drive the highest 'sense of presence'. This will allow individuals to become better acquainted with the virtual world, and will decrease any potential mechanical/psychological/physiological factors that may change the way we stand. In addition, measurement of the 'sense of presence' should be implemented into future research. There have been many different methods used to measure ones 'sense of presence' including physiological arousal (heart rate, EDA, or body temperature) and subjective questionnaires ("How often did you feel the virtual scene become reality?") [Meehan et al., 2005]. Finally, further research should observe the 'least acceptable' parameters that can be used in VR. Mechanical factors such as FOV and end-to-end latencies are important in VR research as they have been influential in increasing 'sense of presence' [Meehan et al., 2005], and decreasing postural changes associated with VR. Various studies have observed significant changes with regards to these types of factors; however, this work may be limited to the type of VR system used; for example, experimenters may use a CAVE system (projection screens placed around the subject display the virtual environment) instead of an HMD. Therefore, it is important to describe limits as to how wide a FOV should be or how long end-to-end latencies can be before differences emerge.

In addition to advancing VR protocols, my research can lead to implementing VR into normal visual flow experiments. Typically, when VR is used in posturography, it is used to test and measure the visual system. Most research looks at, for example, visual perturbations [Bugnariu & Fung, 2007] or visual motion [Keshner & Kenyon, 2009], which is typically used to observe the relative contribution of the visual system in various balance paradigms. Many of these studies also combine optic flow with support surface perturbations. My research demonstrates the utility of VR in normal visual conditions, and explores the efficacy of applying stable visual environments in postural research.

4.6.2 Fear of Falling Research

There is still a need to move from quiet standing to dynamic balance and possibly locomotion when investigating the influence of fear and anxiety. One main question that has progressed through recent literature is the idea that the changes during quiet standing may aid or hinder balance performance in dynamic stance. It has been postulated that stiffening can modulate movements in dynamic situations [Allum et al., 2002], and so further studies may address this idea.

Most investigations of fear and balance occur within young healthy adults, which in itself is important as it demonstrates a direct link between emotion and balance; however, there is an essential component missing: testing populations who are at risk of falling. While still remaining within the safety confines of the lab, researchers can assess older adults (for example) in threatening scenarios not otherwise accessible in real laboratories by using virtual environments. Thus it becomes possible to explore the role psychological, and not just physiological, factors have in moderating postural control in populations with a risk of falling. Finally, virtual reality can be used to experiment with other forms of anxiety and postural threat, such as social anxiety

[Geh et al., 2011] or specific phobias like arachnophobia (spider fear), and allow researchers the ability to further the generalization of anxiety related changes in postural control.

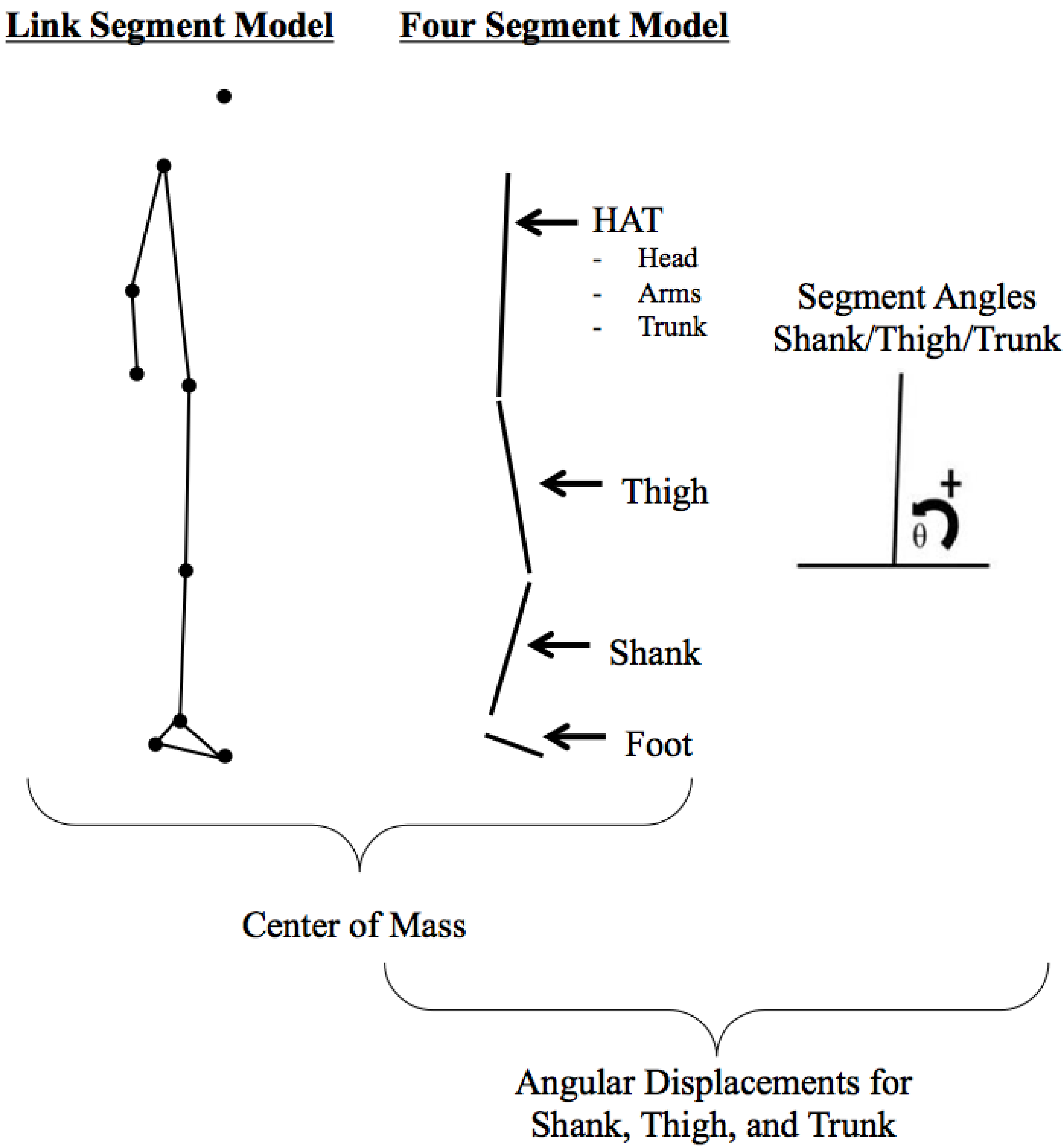
4.7 Conclusions

Virtual heights are capable of eliciting increased levels of fear and anxiety in young healthy adults. These emotional changes affect the postural responses observed when exposed to support surface translations. The results of the current study demonstrate changes in neuromuscular and kinetic characteristics of human postural control when participants are exposed to a bi-directional postural threat. The implications of this thesis may provide useful evidence to guide clinicians in testing not only the physiological impairments associated with balance deficits, but also the psychological disorders; as well as open up new avenues for observing fear related effects on postural control. The results of this thesis, demonstrating the utility of virtual heights, expand on previous work validating the efficacy of VR in dynamic postural control. Due to many feasibility and safety issues surrounding lab settings, it is important to discover new methodological techniques in order to study all forms of balance. VR can be used in future research studying balance deficits in various populations including the elderly or patients with Parkinson's disease, by applying situation specific fears and observing ecologically valid events.

Figure 1. Experimental Setup

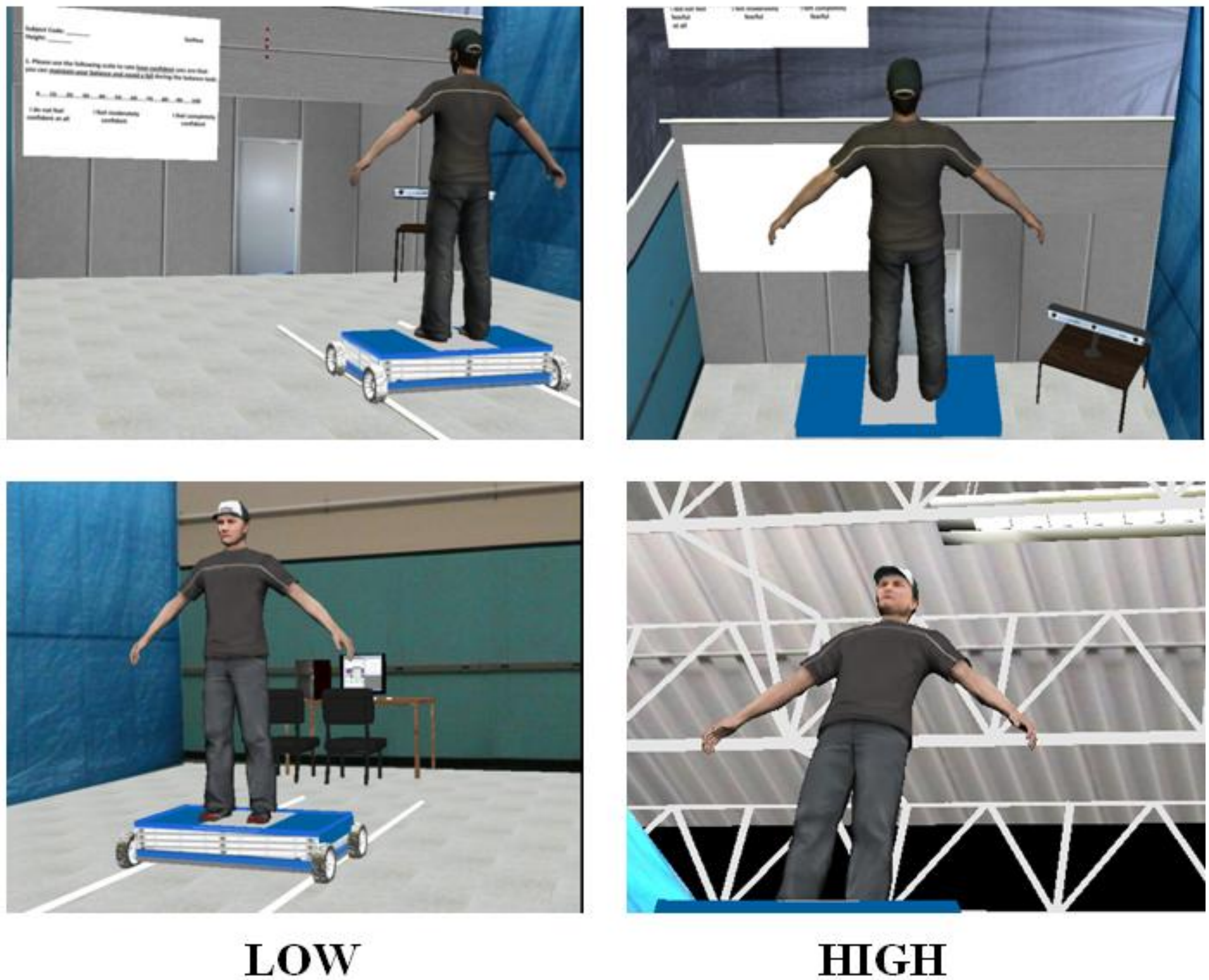
Image of the experimental setup used. The HMD was placed around the head of the participant. Participants stood on the force plate (grey square) located in the middle of the platform.

Figure 2. Kinematic Models



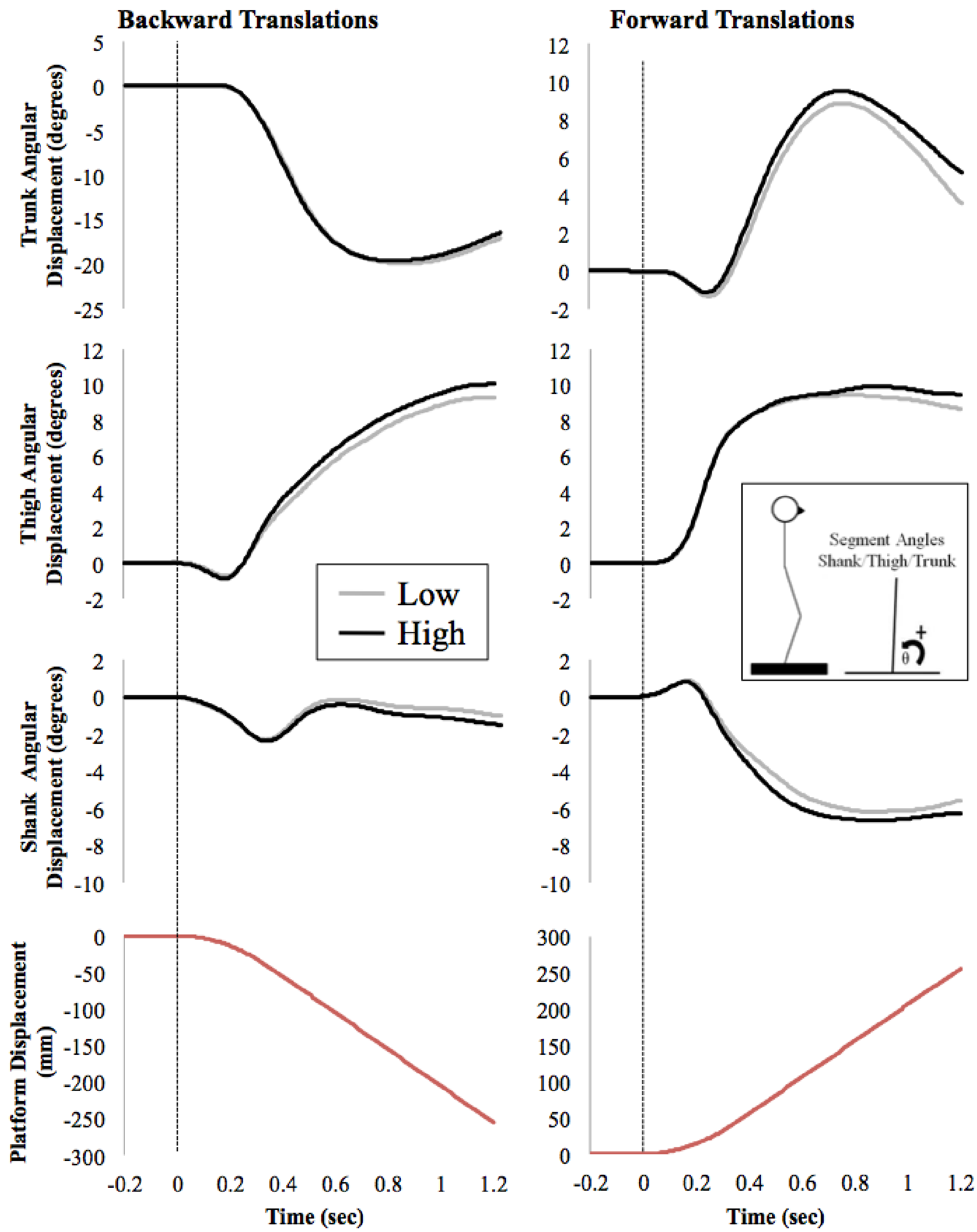
A depiction of the COM model and segment angle calculations for shank, thigh, and trunk.

Figure 3. Virtual Environment



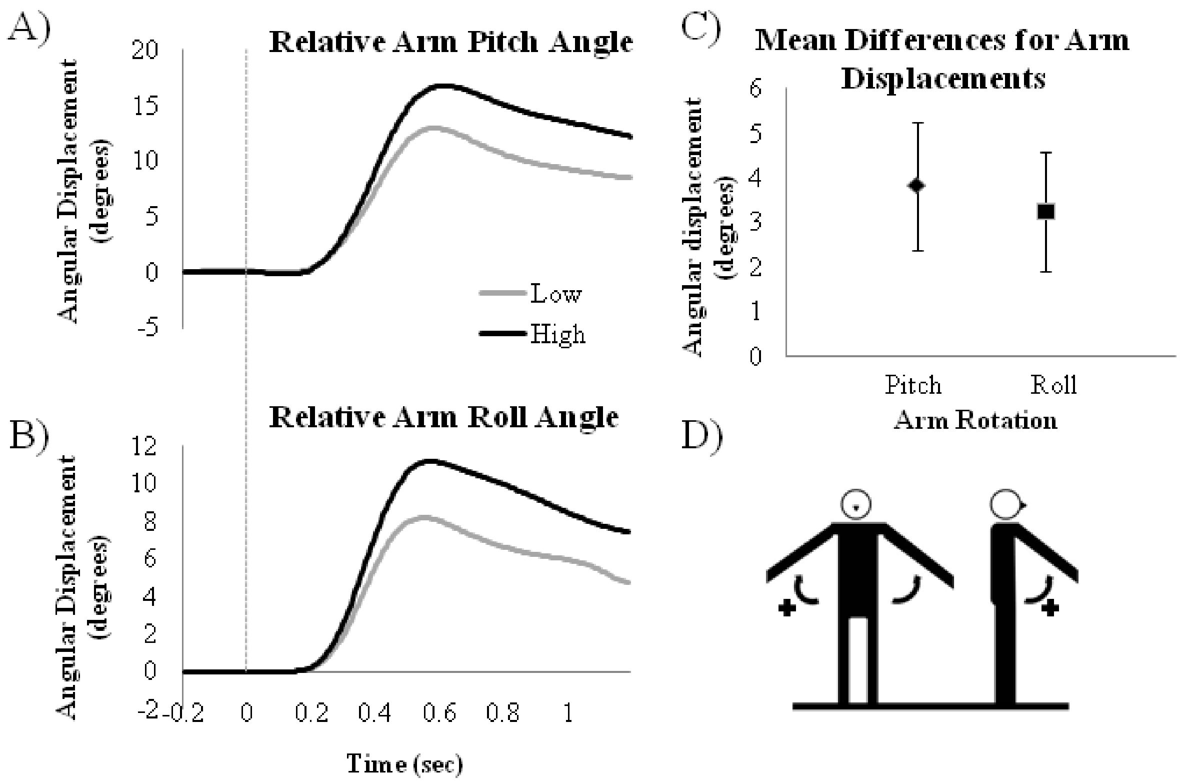
Images of the virtual environment used for the low (left panel) and high (right panel) conditions. Avatar in view is a representation of where the individual stood. Participant's movements were mirrored by this avatar.

Figure 4. Kinematic Responses



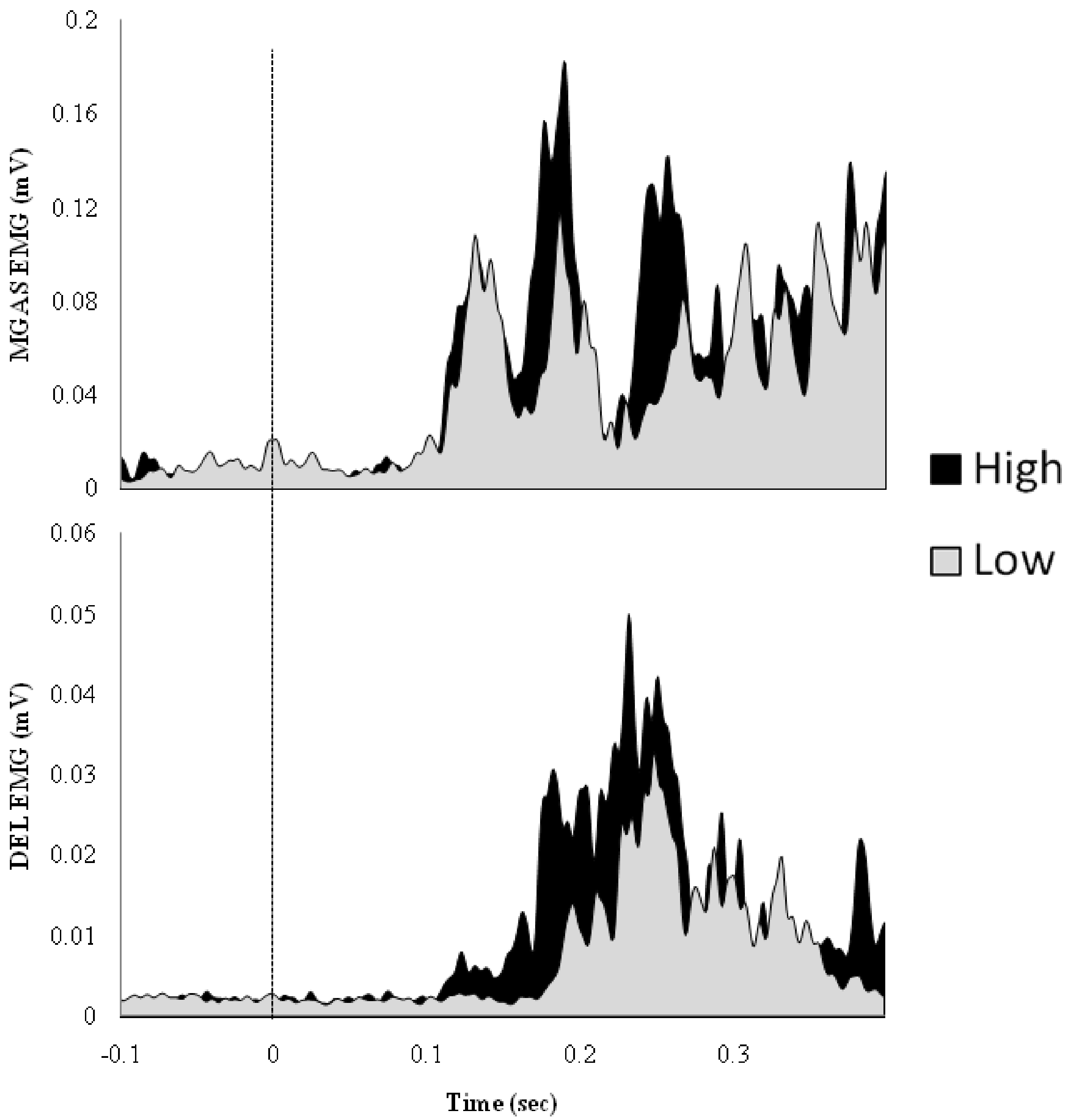
Group mean traces for angular displacements of the shank, thigh and trunk, and platform displacements (red lines) for backward (left panel) and forward (right panel) perturbations in the low (grey lines) and high (black lines) conditions. Vertical dashed lines indicate onset of platform movement.

Figure 5. Arm Displacements



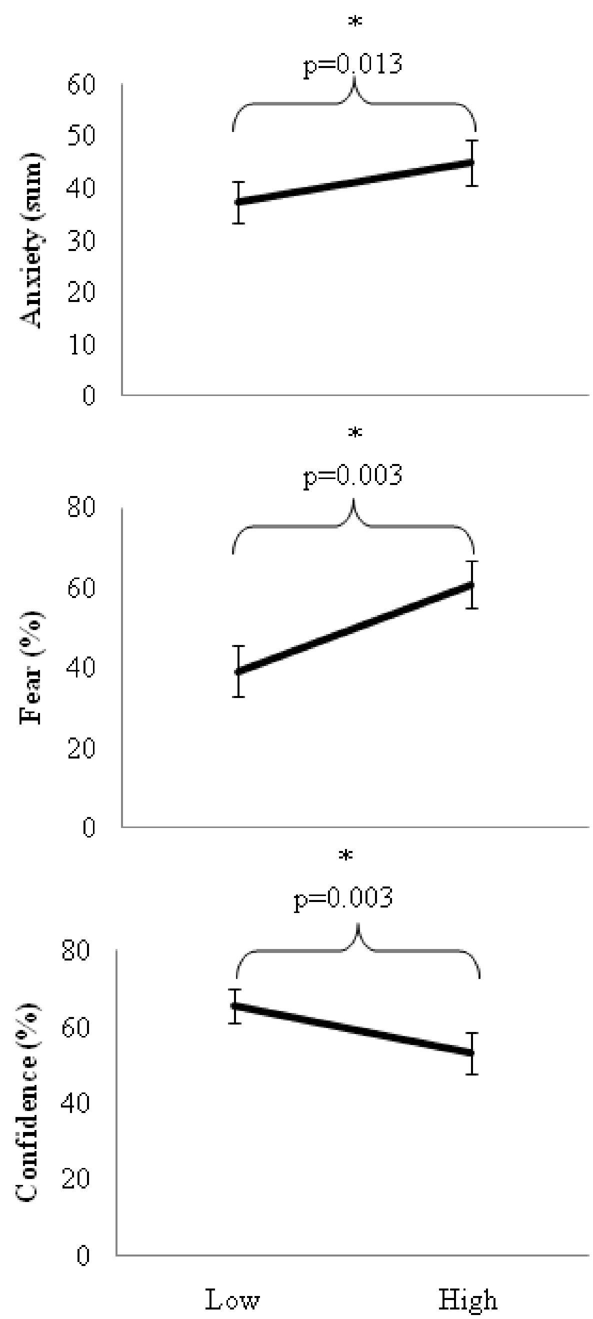
(A and B, left panel) Group mean arm displacements for low (grey lines) and high (black lines) conditions in response to backward perturbations in the pitch (A) and roll (B) direction; Positive values represent a larger flexion or abduction; Vertical dashed line indicates onset of perturbation movement. (C, top right panel) Mean differences (± 1 SE) for arm displacements; (low – high) results in positive values representing larger displacement at height. (D, bottom right panel) Relative angle calculations between arm and trunk for roll (left) and pitch (right) displacements.

Figure 6. Representative Subject EMG Responses



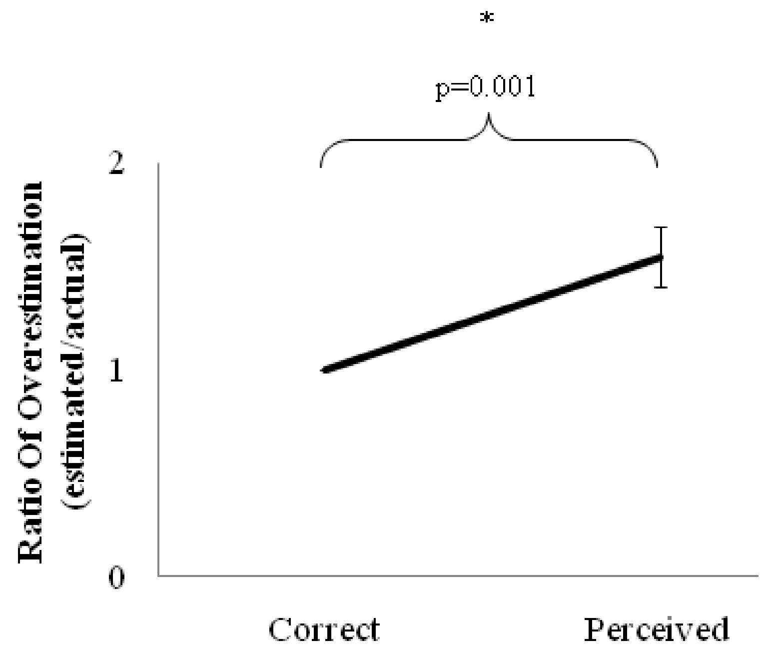
Representative subject average plot of muscle responses (top panel: MGAS, bottom panel: DEL) to backward perturbations in the low (grey area) and high (black area) conditions. Data was 100 Hz low pass filtered for this figure.

Figure 7. Psycho-social Measures



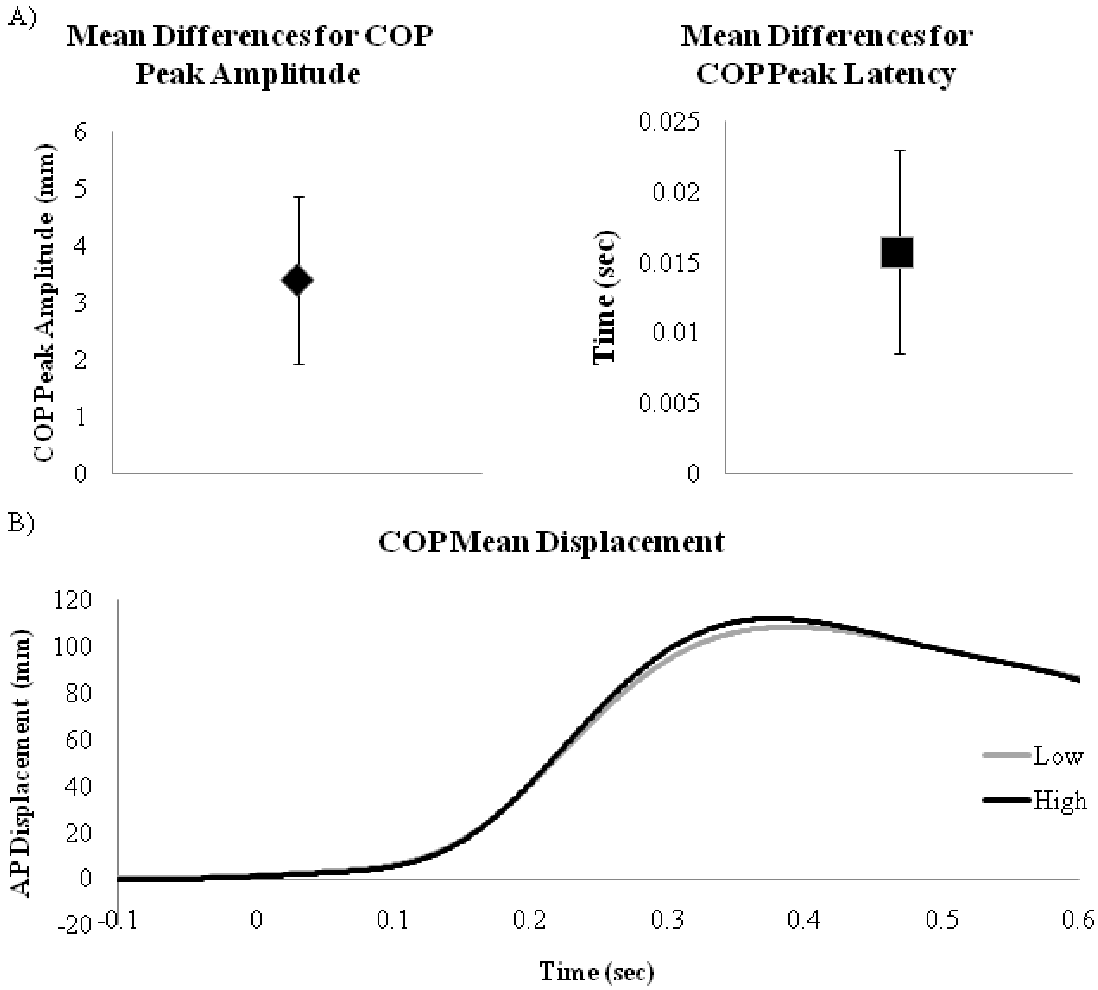
Group means (± 1 SE) for anxiety (top panel), fear (middle panel), and balance confidence (bottom panel).

Figure 8. Height Estimation



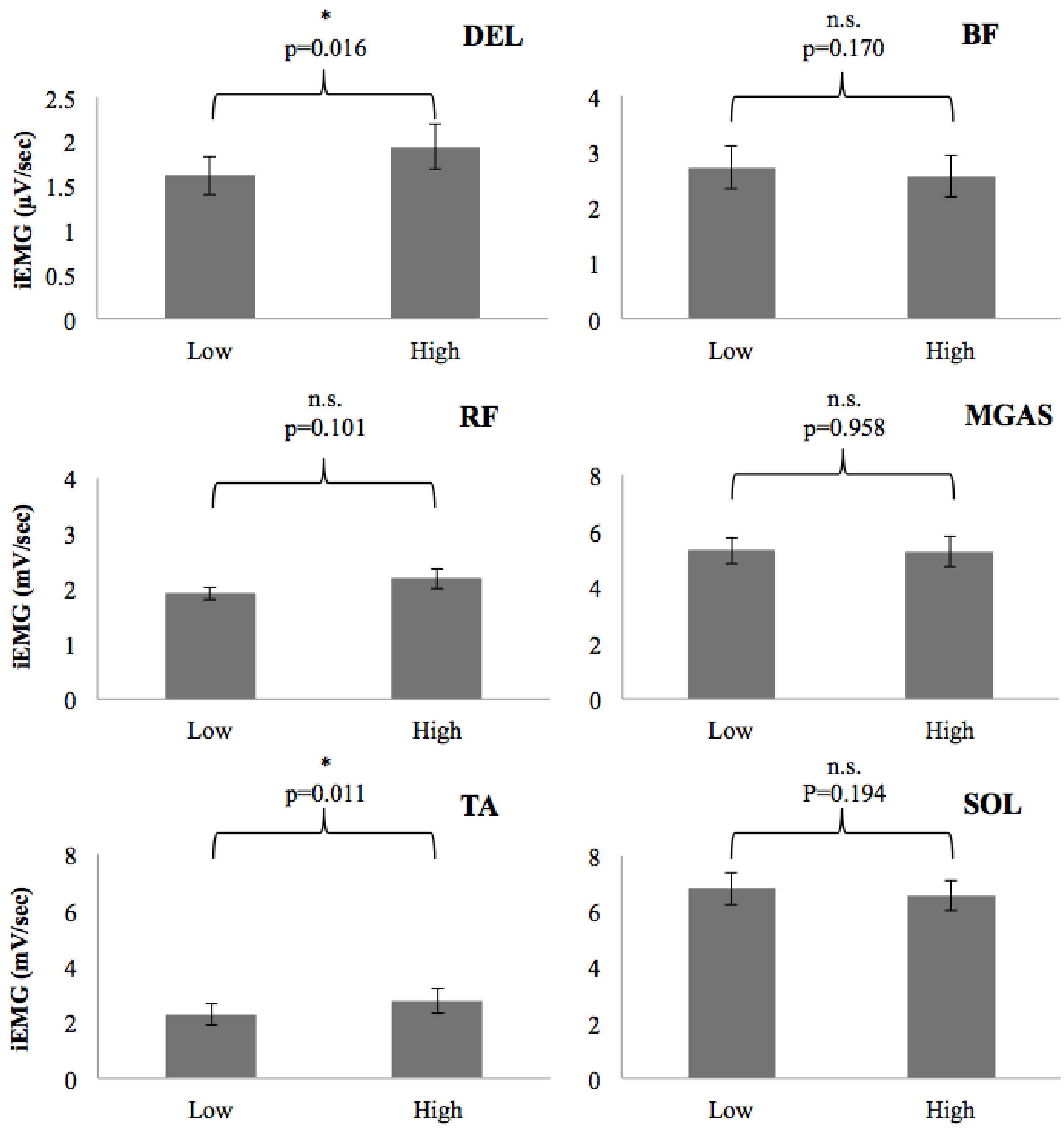
Ratio of overestimation calculated from the estimated height given from participants when standing at the virtual height (mean ± 1SE). Estimates are compared to a ratio of 1, where actual height and perceived height are equal.

Figure 9. COP Displacement



(A, top panel) Mean differences for COP amplitude (left panel) and latency (right panel); larger values represent a larger and earlier response at height. (B, bottom panel) Group average trace for COP displacement in response to backward perturbations in the low (grey line) and high (black line) conditions.

Figure 10. Background EMG



Group mean (± SE) background EMG (amplitude) in low and high conditions for DEL (top left panel), BF (top right panel), RF (middle left panel), MGAS (middle right panel), TA (bottom left panel) and SOL (bottom right panel).

Table 1. Mean (SE) latencies for all muscles in response to backward perturbations

	Backward Perturbations **EMG Latency (sec)**							
	SOL	MGAS	TA	RF	BF	PAR	DEL	OBL
Low	0.165	0.152	0.211	0.210	0.238	0.270	0.176	0.177
	(0.011)	(0.008)	(0.014)	(0.011)	(0.019)	(0.011)	(0.011)	(0.008)
High	0.146 *	0.137	0.205	0.190	0.218	0.260	0.164	0.165
	(0.006)	(0.006)	(0.015)	(0.008)	(0.016)	(0.012)	(0.010)	(0.008)

* represents a significant difference between low and high conditions ($p \leq 0.05$)

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Appendices

Appendix A Balance Confidence Questionnaire

Subject Code: _______   Surface Height: _______

1. Please use the following scale to rate <u>how confident</u> you are that you can <u>maintain your balance and avoid a fall</u> during the balance task:

0……10……20……30……40……50……60……70……80…..90……100

I do not feel confident at all	**I feel moderately confident**	**I feel completely confident**

Appendix B Anxiety Questionnaire (light grey questions not used in anxiety score)

Subject Code:_________________ Surface Height:___________

Please answer the following questions about how you honestly feel just after standing at this height using the following scale:

1	2	3	4	5	6	7	8	9
I don't feel at all				I feel this moderately				I feel this extremely

1. **I felt nervous when standing at this height**
2. **I had lapses of concentration when standing at this height**
3. **I had self doubts when standing at this height**
4. **I felt myself tense and shaking when standing at this height**
5. **I was concerned about being unable to concentrate when standing at this height**
6. **I was concerned about doing the balance task correctly when standing at this height**
7. **My body was tense when standing at this height**
8. **I had difficulty focusing on what I had to do when standing at this height**
9. **I was worried about my personal safety when standing at this height**
10. **I felt my stomach sinking when standing at this height**
11. **While trying to balance at this height, I didn't pay attention to the point on the wall all of the time**
12. **My heart was racing when standing at this height**
13. **Thoughts of falling interfered with my concentration when standing at this height**
14. **I was concerned that others would be disappointed with my balance performance at this height**
15. **I found myself hyperventilating when standing at this height**
16. **I found myself thinking about things not related to doing the balance task when standing at this height.**

Appendix C Stability and Fear Questionnaires

Subject Code:_________________ Surface Height:___________

Please answer the following questions about how you honestly feel just after standing at this height using the following scale:

1. Using the following scale, please rate <u>how stable you felt</u> when performing the balance task:

0……10……20……30……40……50……60……70……80…..90……100

I did not feel completely stable at all	**I felt moderately stable**	**I felt stable**

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2. Using the following scale, please rate how <u>fearful of falling you felt</u> when performing the balance task:

0……10……20……30……40……50……60……70……80…..90……100

I did not feel fearful at all	**I felt moderately fearful**	**I felt fearful**