POSTURAL THREAT INFLUENCES POSTURAL STRATEGY AMONG
HEALTHY YOUNGER AND OLDER ADULTS

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DEDICATION

I would like to dedicate this thesis to my parents, my mother who taught me the value of persistence and patience and my father who taught me the value of hard work and effort. Without them, this thesis would never have been possible.

ABSTRACT

The effects of postural threat on postural control among younger and older adults were examined. Fifteen younger (YA; 8 females and 7 males; mean age 22.00 ± 2.17 years) and fifteen older (OA; 10 females and 5 males; mean age 69.98 ± 5.35 years) adults performed quiet standing and forward reaching under four conditions of postural threat. Postural threat was achieved by the manipulation of height (low (0.43m) and high (1.4m)) and stepping constraint (unconstrained (0.91m from the anterior edge of an elevating platform) or constrained (0m from the anterior edge of an elevating platform). Younger and older adults demonstrated conservative modifications to postural control that may reduce the likelihood of a fall in tenuous conditions. Interestingly, age-related differences emerged in the mechanism of achieving these accommodations to postural threat. Our findings indicate that older adults may adopt more proximal postural strategies under conditions of postural threat. The shift toward a more proximal control of balance may reflect the age-related declines in the ability to control the movement of the trunk. Although these adaptations appear beneficial to older adults, the possibility exists for detrimental consequences to postural recovery following a balance disturbance.

ACKNOWLEDGEMENTS

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I. GENERAL INTRODUCTION

The purpose of this thesis is to investigate the effects of postural threat on postural strategy among healthy younger and older adults. This thesis will include a general introduction, two research papers, and a general discussion. The general introduction is aimed at providing background into the area of postural control and aging and to provide insight into the problem of fear of falling in the elderly. The first research paper examines the effects of postural threat on the maintenance of an upright stance among healthy younger and older adults during quiet stance. The second research paper examines the effects of postural threat on postural control during preparatory and focal movements of a functional forward reach among healthy younger and older adults. The general discussion summarises the major research findings and describes their contributions to the current body of literature.

A. Overview

The human body may be modelled as an inverted pendulum in which two-thirds of the body mass is located two-thirds of the body height from the ground (Winter, 1995). Due to the inherently unstable nature of this system, maintaining balance is a very challenging task. To accomplish this function, the Central Nervous System (CNS) constructs an internal picture of the body and the surrounding environment from sensory information obtained from the somatosensory, visual, and vestibular systems. Constant changes in the system cause adjustments to this schema. To ensure equilibrium, the CNS continuously directs appropriate responses to the musculoskeletal system that maintain balance. If pathology in the function of any of these systems arises, dysequilibrium may occur.

Falls are a well-known phenomenon in the elderly with approximately 30 to 50 percent of seniors, aged 65 and over, incurring a fall each year (Tinetti & Williams, 1997; Suzuki, Shimamoto, Kawamura, & Takahasi, 1997). Falling is a serious and debilitating problem that often results in death and serious injury (Kellogg International Work Group on the Prevention of Falls by the Elderly, 1987). Fall-related injuries lead to reduced confidence, inactivity, and loss of independence in many elderly fallers (Coni, Davison, & Webster, 1992).

To prevent falling, the heterogeneous process of aging must be investigated. Although researchers are unclear on the model for aging (Woollacott, 2000; Woollacott, Shumway-Cook, & Nashner, 1986), marked decreases in the function of all systems that contribute to balance have been established as a precursor to instability. Older adults suffer from peripheral neuropathy, poor vision, and reduced vestibular function (Woollacott,

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2000). In general, sensory system deficits reduce the ability to detect changes in the environment. Further instability may result from age-related changes in the musculoskeletal system. Age-related declines in muscle strength and flexibility may result in a decreased ability to respond to imbalance (Whipple, Wolfson, & Amerman, 1987). However, perhaps the most serious threat to balance may be deficits in CNS function (Hay, Bard, Fleury, & Teasdale, 1996; Teasdale, Stelmach, Breunig, & Meeuwsen, 1991). Age-dependent impairments in CNS processing can be observed in the timing and effectiveness of the postural response necessary to maintain upright stance.

Although physical changes subserve the loss of balance function, psychological factors have also demonstrated influence on balance ability. Because falls have such a major influence on the health and quality of life of the elderly, many older adults experience an anxiety or fear about losing their balance, regardless of previous fall history (Downton & Andrews, 1990; Silverton & Tideiksaar, 1989). Fear of falling is a psychosocial phenomenon that leads to further deterioration in balance through activity restriction and immobility (Howland, Lachman, Peterson, Cote, & Jette, 1998; Grimley Evans, 1992).

Past clinical research has demonstrated that fearful older adults have poorer balance than nonfearful older adults (McAuley, Mihalko, & Rosengren, 1997; Franzoni, Rozzini, Boffelli, Frisoni, & Trabucchi, 1994; Tinetti, Richman, & Powell, 1990). In these experiments, fear is determined by self-report and balance function is measured by performance on classic balance tests such as the Berg Balance Scale (Berg, Maki, Williams, Holliday, & Wood-Dauphinee, 1992) or the functional reach (Duncan, Weiner, Chandler, & Studenski, 1990). Current research has adopted a more robust, laboratory approach that induces balance anxiety through the environmental contexts that alter the potential consequences of a fall (Brown, Sleik, Gage, & Polych, 2002; Brown, Gage, Polych, Sleik, &

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Winder, 2002; Adkin, Frank, Carpenter, & Peysar, 2002; Carpenter, Frank, Silcher, & Peysar, 2001; Adkin, Frank, Carpenter, & Peysar, 2000; Carpenter, Frank, & Silcher, 1999; Brown & Frank, 1997). Results indicate that increased postural threat imposes modifications to postural control; however, it remains unclear whether or not these accommodations are beneficial to postural control.

B. Postural Control

Balance is an integral part of many of the activities of daily living (ADL). Even seemingly simple tasks such as standing and walking can be quite challenging when observed from a mechanical perspective. For example, standing requires that the large mass of the body be balanced over two spindly structures like the legs. Although the body is essentially stationary during static balance tasks such as standing, continuous processing of the Central Nervous System (CNS) is crucial for the maintenance of balance. The CNS monitors the status of the body and the external environment through the peripheral sensory systems. From this information, the CNS exacts appropriate musculoskeletal corrections to ensure ongoing government of balance.

CNS regulation is also required for dynamic balance tasks such as walking. Dynamic activities pose a greater threat to balance; therefore, greater CNS involvement is required than during static tasks (Teasdale, Bard, LaRue, & Fleury, 1993; Lajoie, Teasdale, Bard, & Fleury, 1993). Dynamic balance refers to conditions when the body is in a state of motion. Gait is initiated by a voluntary forward step that moves the mass of the body forward in a manner similar to the beginning of a fall. In order to preserve balance during locomotion, another step must occur. It is this self-initiated movement that enables the body to propel forward in locomotion (Winter, MacKinnon, Ruder, & Wieman, 1993).

As a further threat to balance, the CNS is continuously presented with internal and external disturbances or perturbations. Internal disturbances are caused by regulating involuntary body processes such as circulation and respiration and voluntary movements such as leaning and reaching. Similar to gait, these perturbations can cause the body to move in the same direction as the movement, increasing the risk of a fall event. Therefore,

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to prepare for an intended movement, the CNS executes anticipatory postural adjustments that minimise the probability of a loss of balance.

Although most internal perturbations are highly regulated and accommodated for by the CNS, there are many disturbances caused by the external environment that are less predictable. External perturbations such as slips and trips present the greatest challenge to balance because we are often unable to prepare for the impending imbalance. This compensatory mechanism requires early detection of the disturbance and the implementation of an appropriate and effective response.

1. Definition of Postural Control

Postural control is the process of regulating the position of the body in space for the purpose of achieving an upright and stable stance (Shumway-Cook & Woollacott, 2001). To achieve this goal, the centre of mass (COM) of the body must be maintained within the base of support (BOS). The COM refers to a theoretical point in space that represents the net location of the body mass. The BOS refers to the area of the body in contact with the support surface, commonly prescribed by the dimensions of the feet. Effective postural control requires that an appropriate relationship be maintained between the body and the environment to preserve equilibrium. If this relationship is violated, a loss of balance may occur.

2. The Biomechanics and Neural Mechanisms of Postural Control

The human body is an inherently unstable system because two-thirds of the body mass is positioned two-thirds of the body height from the ground (Winter, 1995; Winter, Patla, & Frank, 1990). Therefore, researchers have simplified the multisegmented nature of the human body using an inverted pendulum model to investigate the control of balance

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(Fig. 1.1). The inverted pendulum model accurately depicts postural control in the frontal and sagittal planes (Rietdyk, Patla, Winter, Ishac, & Little, 1999; Winter, Prince, & Patla, 1997; Winter, 1995; Yang, Winter, & Wells, 1990) even though the antero-posterior (AP) and medio-lateral (ML) control of balance operate completely independent of each other (Winter, Prince, Stergiou, & Powell, 1993). AP balance is controlled at the ankle joints through the contraction of plantarflexor and dorsiflexor calf muscles while the ML balance is controlled at the hip joints through the contraction of abductors and adductors using a load/unload
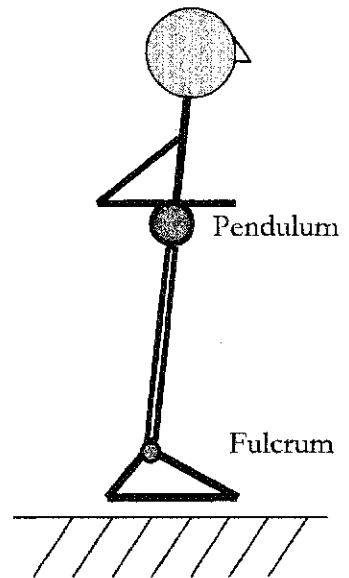
Figure 1.1: Schematic diagram of the inverted pendulum model for postural control. The COM represents the pendulum and the ankle represents the fulcrum about which the COM moves.

mechanism (Deniskina & Levik, 2001; Gatev, Thomas, Kepple, & Hallett, 1999; Winter, Prince, Frank, Powell, & Zabjek, 1996; Winter et al., 1993). Research has primarily focused on postural control in the AP dimension because movement is naturally constrained to the sagittal plane during quiet stance and gait (Diener & Dichgans, 1988); however, recent research has put forward the need to understand ML control of balance because the ML dimension may be more indicative of fall-risk (Maki, Edmonstone, & McIlroy, 2000; Maki & McIlroy, 1996).

The position of the COM can be described in three-dimensional (3D) space (Fig. 1.1). In the AP plane, the COM is located slightly anterior to the ankle joint. Medio-laterally, the COM is found at the midpoint between the feet (i.e. an even weight distribution), and vertically (V), it is located at approximately two-thirds of the body height from the ground. The centre of pressure (COP) serves to control the COM movement.

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COP is the net location of the ground reaction force. The ground reaction force is the product of the muscular forces exerted by the body to control the movement of the COM and the gravitational force of the ground acting on the body.

During all activities, there are continuous and natural fluctuations in the position of the COM due to gravitational forces acting on the body and concomitant reactions to this force. These small movements are referred to as postural sway. Consistent with the inverted pendulum model, these displacements occur at the fulcrum of the pendulum, the ankle joint. Spontaneous postural sway is defined as the minute movements of the COM that occur when trying to remain still during quiet stance. Spontaneous sway occurs because the COM is not perfectly aligned with the ankle joint during upright standing. This offset creates a torque

Destabilizing Moment

Stabilising Moment — COP

Figure 1.2: Schematic diagram of the inverted pendulum model for postural control. During upright stance, a destabilizing moment is produced by the downward projection of the COM anterior to the ankle joint; to oppose this torque and maintain an upright stance, the CNS generates a stabilizing moment at the ankle joint through the contraction of posterior leg muscles.

or moment about the ankle joint that can be calculated as the product of the body's centre of gravity (COG; the vertical projection of the COM) and the horizontal distance between the COM and the ankle (i.e. moment arm distance) (Fig. 1.2). The position of the COM produces a clockwise, gravitational moment that causes the body to rotate forward. Without a counteractive force, this moment would be destabilizing and would cause the body to fall forward. However, to oppose the gravitational moment, the CNS produces a stabilizing moment at the ankle joint through the contraction of posterior calf muscles (Fig. 1.2). This

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moment can be calculated as the product of the reactive force and the horizontal distance between the COP and the ankle. The magnitude and direction of the stabilizing moment exceeds the destabilizing moment and functions to overcome the forward sway of the body and direct the COM posteriorly. As the body begins to sway backward, muscles of the anterior lower leg contract and produce a clockwise ankle moment that directs the COM anteriorly. Thus, the CNS is constantly correcting the imbalance of the COG by generating muscle forces that minimise the movement of the COM within the BOS.

The inverted pendulum model predicts a relationship between the COM and the COP. Winter and colleagues (1998; 1995; 1993; 1990) have tested this hypothesis and revealed that the movement of the COP is highly correlated with the activity of the COM. In fact, in the absence of horizontal acceleration, the COP and COM are perfectly aligned. Even in a dynamic system such as the human body, where accelerative forces always exist, the COP position must equal the COM position when averaged over a prolonged period of time (Eng & Winter, 1993). The COP displacement must be larger in amplitude and frequency than the COM displacement because it must oscillate on either side of the COM to maintain the COM within a relatively fixed position. When the COM is limited to a smaller range of movement, the probability of exceeding the stability limits and incurring a loss of balance is minimised. Figure 1.3 depicts the dynamic interaction between the COP and the COM during the quiet stance of a healthy younger adult. Once the COM position exceeds the COP position, the CNS alters the position of the COP until it is anterior to the COM. The anterior position of the COP directs the COM posteriorly to avoid a forward loss of balance.

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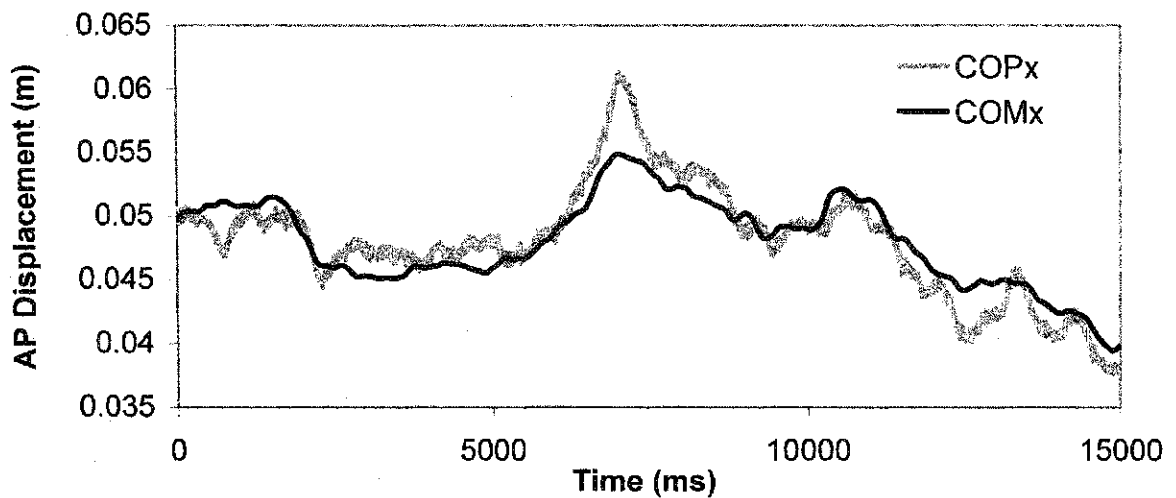
Figure 1.3: The relationship between the COP and COM in the Anterior/Posterior (AP) directions of a 21 year old female participant during 15s of feet together, eyes open quiet standing. Note that the movement of the COP controls the position of the COM. For example, if the COM position becomes to anterior, the COP will move anterior to the COM to drive the COM posteriorly.

Another prediction of the inverted pendulum model is that the horizontal acceleration of the COM equals the difference in the COM and COP position (Winter et al., 1990). This difference is referred to as the 'error signal' and is overcompensation by the CNS to correct the imbalance of the COM. Winter and collaborators (1998; 1996; 1993; 1990) provided evidence in support of this hypothesis. They discovered that the COP always maintains a slightly larger displacement than the COM. In fact, the error is approximately 0.8 mm and 0.5 mm in the AP and ML direction respectively (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). The linear acceleration of the horizontal COM is proportional to this error; therefore very small changes in the COP cause large movements of the COM (Winter et al., 1998; Winter, 1995; Eng & Winter, 1993; Winter et al., 1990) (Fig. 1.4). These variables maintain a strong negative correlation with each other with experimental values yielding correlations of −0.93 to −0.99 (Winter et al., 1998). As

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demonstrated in Figure 1.4, when the COP is anterior to the COM, a posterior horizontal

acceleration is generated to drive the COM posteriorly.
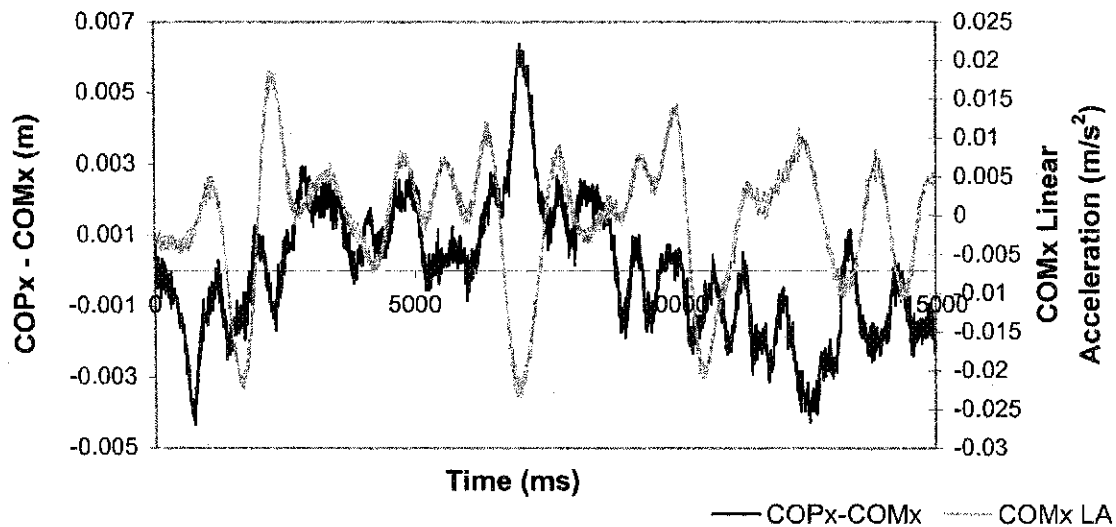
Figure 1.4: The relationship between the error signal (COP-COM) and the COM linear
acceleration of a 21 year old female participant during 15s of feet together, eyes open
quiet standing. In an inverted pendulum model, the error signal and the horizontal
COM acceleration are strongly negatively correlated (Winter et al., 1998).

The error signal represents the stiffness control of the inverted pendulum. The

stiffness control model states that the CNS controls balance by setting the stiffness of the

muscle tone in the postural muscles. Stiffness is achieved through the simultaneous

activation of agonist/antagonist muscle pairs. The stiffness must be large enough to

overcome the gravitational load by directing the COP to surpass the COM (Eng & Winter,

1993). The CNS regulates the degree of stiffness in the muscles according to task demands.

For example, during upright stance, the control of balance is not very difficult and,

therefore, the muscle activity is relatively low. In contrast, when postural conditions become

more challenging, tighter regulation of balance must occur; therefore, the CNS increases the

amount of muscle stiffness to restrict the range of movement at the joint (Winter, Patla,

Rietdyk, & Ishac, 2001; Winter et al., 1998; Winter et al., 1996). In this way, the movement of the COM position is fixed to an even smaller area to prevent a loss of balance from occurring.

a) Methodological Techniques in Postural Control

Postural control can be measured using laboratory or clinical techniques. Clinical balance tests are typically used to assess functional balance ability in patients. These tests evaluate balance during common activities of daily living. Laboratory balance tasks can provide more in-depth knowledge about postural control during a diverse range of movements, some of which may contain functional components.

(1) Laboratory Assessment

Laboratory tests of postural control are designed to provide a rigorous assessment of postural sway. These tests examine gross body movement, the forces and moments that produce the observed movement, as well as, the neural mechanisms underlying the observed movement.

(a) Kinematic Measurement

An optical imaging approach is a common method for describing the movement of the COM. The general principle behind optical techniques is to record the movement of the body using a camera system. In the past, optical systems mainly included cinematography, television, and multiple exposure imaging systems; however, today's research tends to shift toward light reflective and optoelectric systems. These motion analysis systems use specialized high-speed cameras to record the exact location of passive or active markers located on specific landmarks of the body. After marker digitisation, the position versus

time history of each marker in 2D or 3D space can be plotted for the entire trial. From this information, displacement, velocity, and acceleration of the landmarks and joints can be calculated. The COM is the weighted sum of the centre of mass of all the segments of the body. Although the COM can be calculated with any number of segments, accuracy increases proportionately with the number of segments (n) used in the model for calculating the COM. Because the COM is designed to represent the entire body, error of calculation is minimised through the inclusion of all large body segments (Eng & Winter, 1993). The algorithm for 'n' segments is as follows:

$$\text{COM} = \frac{M^*L_{(1)} + M^*L_{(2)} + M^*L_{(3)} + M^*L_{(4)} + M^*L_{(n)}}{M_{(1)} + M_{(2)} + M_{(3)} + M_{(4)} + M_{(n)}}$$

where M refers to segment mass and L refers to the segment position

Although time series data adequately describe the fluctuations of the body, summary measures, such as mean position, range or sway area, variability (e.g. standard deviation (SD) or root mean square (RMS), frequency (e.g. mean power frequency (MPF)) are used to conceptualise COM kinematics.

(b) Kinetic Measurement

To quantify the neural control of the COM, a force platform is used. A typical force plate contains four triaxial force transducers mounted at right angles from one another and located in each of the platform's quadrants. These transducers measure changes in the amount of strain within the transducer in response to an applied force. The strain is translated and output to the user as forces (F) and moments of force (M). With force transducers in each quadrant, forces and moments of force can be measured in three dimensions (x = AP, y = ML, and z = V). The COP is calculated as the quotient of the moment by the ground reaction force in both the AP and the ML directions.

$COPx = -My/Fz$
$COPy = Mx/Fz$

As with kinematic descriptors, time series data and summary variables (e.g. mean position, range or sway area, SD or RMS, MPF) are the most common measures used for analysis and interpolation.

(c) Neuromuscular Measurement

Another technique for the investigation of postural control is electromyography (EMG). EMG provides a window into the activity of the CNS. During EMG collections, electrodes are either inserted into the muscle or placed superficially over the muscle bellies to measure electrical activity. From the EMG output, the spatial and temporal patterns of muscle activity are analysed. The most frequently examined spatial characteristics determine which muscles are active, the amplitude of activity, and the sequence of activation. The key temporal variable is the onset of muscle activations. The onset of EMG represents the amount of time required by the CNS to mediate a musculoskeletal response to correct imbalance.

Another important variable used in EMG analysis is the amount of co-activation between agonist and antagonist muscle pairs (e.g. simultaneous activity of the Soleus m. and the Tibialis Anterior m.). Winter (1998) has suggested that cocontraction of the lower leg muscles is the mechanism for stiffness control about the ankle joint. According to this hypothesis, passive changes in muscle tone, accomplished through increased agonist muscle coactivity, drives a tighter regulation of COM through increased frequency and decreased variability in postural sway (Winter et al., 2001; Winter et al., 1998; Winter et al., 1996).

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(2) Clinical Assessment

Postural control can also be investigated using clinical tests. Clinical tests, such as functional reach, are designed to obtain a general estimate of functional balance mobility. Typical clinical tests investigate timed characteristics of balance performance under challenging postural tasks such as the sharpened Rhomberg (i.e. tandem stance where the feet are aligned heel to toe) and one-legged stances (Berg et al., 1992). These stances are more difficult to maintain because the BOS is reduced. Therefore, the displacement of the COM must be restricted to a smaller area to ensure the limits of stability are not breached. In addition to the manipulation of the BOS dimensions, these stance tests are often performed under different sensory conditions (usually eyes open and eyes closed) for the purpose of identifying balance pathologies (Tang & Woollacott, 1996). Balance ability is measured by the duration of time that the participant is capable of maintaining each stance.

There are also other applied clinical tests that focus on balance performance during ADLs. These functional tests include a battery of static and dynamic tasks such as standing up or sitting down, turning, reaching, and walking (Tang & Woollacott, 1996; Bogle Thorbahn, Newton, & Chandler, 1996; Duncan et al., 1990; Tinetti, 1986; Mahoney & Barthel, 1965). These tests are often more useful indicators of balance pathology as they reflect balance ability under conditions representative of real world situations.

3. Reactive Postural Control

Daily living is filled with many unpredictable disturbances to balance such as suddenly appearing obstacles. These situations present the greatest threat to balance because the CNS must recover balance after the COM has been displaced. The CNS is forced to react, or the individual will fall.

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In the laboratory, unexpected perturbations displace the COM. There are two techniques that simulate these unexpected perturbations: 1) a push paradigm, and 2) a translating platform. With a push paradigm, an external force is applied to the body (e.g. shoulders or trunk) in a manner similar to push (Rietdyk et al., 1999; Brown & Frank, 1997). With a translating platform, the support surface is moved linearly beneath the feet (uni- or multi-directionally) in much the same way as a rug might be pulled out from under the feet (Horak & Nashner, 1986; Nashner & Woollacott, 1979; Nashner, 1977). The findings of these studies revealed that balance is recovered through the use of distinct movement patterns (Horak, Henry, & Shumway-Cook, 1997; Horak & Nashner, 1986; Nashner & McCollum, 1985; Nashner & Woollacott, 1979). For the past two decades, researchers have attempted to identify these movement strategies and have revealed two main strategies for the control of balance in the AP plane: the "feet in place", and the "change in support" (Horak et al., 1997; Maki & McIlroy, 1996). In the ML plane, the predominant postural strategy is a "hip load/unload" mechanism (Deniskina & Levik, 2001; Rietdyk et al., 1999; Winter, 1995).
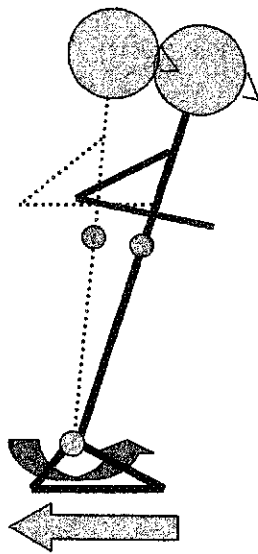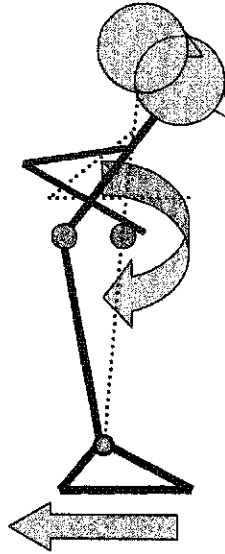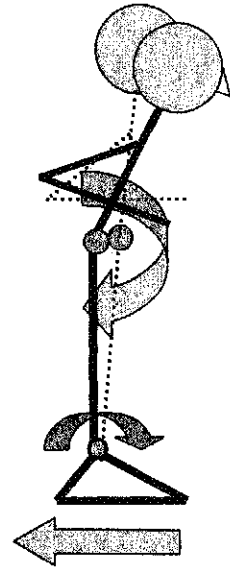
Fig. 1.5.1   Fig. 1.5.2   Fig. 1.5.3

Figure 1.5: Schematic diagram of the different postural strategies employed by the CNS to recover balance following a disturbance. 1.5.1 demonstrates the "ankle" strategy. 1.5.2 illustrates the "hip" strategy. 1.5.3 demonstrates the "mixed" strategy. The blocked arrow represents the direction of perturbation. Curved arrows are used to indicate moments generated about the selected axis of rotation.

The "feet in place" strategy manipulates the position of body segments for the purpose of maintaining the COM within an unchanged BOS. In this strategy, the COM can be controlled at the ankle or the hip joints. The "ankle" strategy conforms to the inverted pendulum model where the body moves as a single segment around the ankle joint (Horak, Nashner, & Diener, 1990). In an "ankle" strategy, the disturbance is compensated for through the activation of lower leg muscles (i.e. dorsiflexors or plantarflexors), and very little movement is permitted around any other joint (Fig. 1.5.1). The stabilizing moments produced about the ankle joint function to recover the COM to its equilibrium position without misaligning other body segments. In contrast, the "hip" strategy causes the body to decouple at the hip creating a double-segment or compound inverted pendulum (Runge, Shupert, Horak, & Zajac, 1999). With the "hip" strategy, the body bends at the waist in the

direction of the sway, but the vertex of the segments, the hip joint, is moved in the opposite direction of the sway (Fig. 1.5.2). The movement of the upper segment would appear to cause a destabilizing displacement of the COM; however, the relocation of the hip joints counters the bending of the upper segment minimising the movement of the COM within the BOS (Runge et al., 1999). A "hip" strategy is accomplished through the rapid and high amplitude activations of the hip flexors/extensors that work with gravity to produce rotation of the trunk segment about the hip joint in the same direction as the sway (Fig. 1.5.2). The use of a "pure hip" strategy is quite rare; instead, the "hip" strategy is often paired with the subsequent execution of the "ankle" strategy. In this "mixed" strategy, both the upper and the lower segments produce movements that conjunctively stabilize the body and prevent the COM from exceeding the BOS (Fig. 1.5.3). The activation of the hip flexors/extensors is succeeded by the contraction of the ankle dorsiflexors/plantarflexors. Therefore, rotation of the upper segment occurs around the hip joint in the direction of the perturbation, and a subsequent rotation of the lower segment occurs around the ankle joint in a direction opposite to the perturbation.

The "change in support" strategy uses compensatory mechanisms such as stepping or grasping that extend the BOS dimensions. The asymmetrical loading and unloading of the legs serves to alter the dimensions of the BOS under the displaced COM (Horak et al., 1997). Because the BOS is larger, the COM is able to move further before a fall will occur. This strategy reduces the likelihood that the COM displacement will exceed the stability limits (Maki & McIlroy, 1996).

The postural strategy executed by the CNS is based on a number of factors. Research indicates that perturbation magnitude has a significant influence on the CNS and its corresponding postural strategies. If the perturbation is sufficient to displace the COM

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beyond the stability limits, a "change in support" strategy is used (Horak & Nashner, 1986). Otherwise, a "feet in place" strategy is typically employed to control the displacement of the COM within a fixed BOS. The CNS responds to low magnitude perturbations with an "ankle" strategy and high magnitude perturbations with a "hip" strategy (Runge et al., 1999; Horak et al., 1997; Horak & Nashner, 1986). The "ankle" strategy is best for maintaining an upright orientation while the "hip" strategy is best for maintaining stability. As predicted by Kuo (1995) and evidenced by Runge and colleagues (1999), perturbations of intermediate magnitude result in a "mixed" strategy. The majority of these results were found in studies that constrained the postural response to a "feet in place" strategy; however, McIlroy and Maki (1993) have demonstrated the "change in support" strategy may occur regardless of disturbance magnitude.

Although perturbation magnitude has a strong influence on strategy selection, the CNS must initiate a musculoskeletal response at the earliest onset of peripheral detection. Thus, the CNS must make postural adjustments prior to knowing all of the characteristics of the disturbance. This fact implies that the CNS relies on other external and internal factors for cues regarding an appropriate postural response. Environmental factors, such as support surface conditions, play a large role in strategy selection. McCollum and collaborators (1984) conducted an experiment where perturbations were delivered under different support surfaces. The findings revealed that the "ankle" strategy is used when the contact surface is wide and firm and the "hip" strategy is used when the surface is narrow or compliant. Another interesting discovery emerged from this work. Immediately following changes in support surface, the CNS adopted a "mixed" strategy for balance recovery in the new support surface conditions. Thus, the CNS utilises an intermittent "mixed" strategy to respond to a perturbation that prior to previous testing conditions, would have evoked a

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typical dichotomous "ankle" or "hip" strategy. This finding implies that the CNS was influenced by prior experience. The work of Horak and Nashner (1986) reinforced that previous experience influences CNS response to perturbation. Finally, the availability of sensory information is crucial to CNS response. Various researchers have established that individual sensory systems may be responsible for producing each postural strategy (Horak et al., 1997; Forssberg & Hirschfeld, 1994; Horak et al., 1990; Horak, Shupert, & Mirka, 1989; Woollacott et al., 1986; Wolfson, Whipple, Amerman, & Kleinberg, 1986). For example, the "hip" strategy is more prevalent when somatosensory input is absent, regardless of perturbation size or support surface characteristics (Horak, Diener, & Nashner, 1989). In contrast, a "hip" strategy does not emerge in the absence of vestibular information (Forssberg & Hirschfeld, 1994). This finding suggests that the ability to produce "ankle" and "hip" strategies are dependent upon the presence of sensory input and that the execution of these responses rely upon different CNS mechanisms.

Balance recovery from lateral perturbations always involves a "hip load/unload" strategy in which the vertical application of force is unequally divided between the two legs. Research has demonstrated that the "hip load/unload" strategy in the ML control of balance may be analogous to the "ankle" strategy found in the AP control of balance (Deniskina & Levik, 2001; Rietdyk et al., 1999; Winter, 1995). In both strategies, the body behaves as a single-segmented system where the movement is controlled about one joint, and the COM is maintained within an unchanged BOS. The "hip load/unload" strategy alters the existing equal weight distribution such that more body weight is shifted toward the leg opposite to the direction of the perturbation to oppose the destabilizing moments at the hip and ensure equilibrium (Horak et al., 1997; Winter, 1995).

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Research over the past two decades has revealed that postural strategies are accomplished through a very specific activation sequence of independent muscles (Rietdyk et al., 1999; Winter et al., 1993; Horak & Nashner, 1986; Nashner & Woollacott, 1979; Nashner, 1977; Nashner, 1976). The stereotypical nature of the sequences of muscle activation has led to the term of description of a muscle synergy (Nashner, 1977; Nashner, 1976). A muscle synergy refers to a centrally organized pattern of activity from a group of muscles that function as one unit. Researchers have postulated that synergies function to reduce the requirements for CNS activity by reducing amount of central processing necessary to evoke a postural strategy to recover balance (Shumway-Cook & Woollacott, 2001).

Nashner and colleagues (1986; 1979; 1977) performed a series of experiments to identify and characterize these muscle synergies. The experimental paradigm involved the use of translating platforms to deliver external perturbations to participants. The results revealed that muscle synergies are highly prescribed by a distinct spatial and temporal patterns of activation. The "ankle" strategy is accomplished by a stereotypical pattern of muscle activation that proceeds in a distal-to-proximal direction (i.e. Gastrocnemius-Hamstrings-Paraspinals or Tibialis Anterior-Quadriceps-Abdominals) in muscles that are opposite to the direction of sway. This pattern was termed an "ankle" synergy. On the contrary, a "hip" strategy is achieved by a proximal-to-distal activation of muscles in the same direction to the direction of sway (i.e. Abdominals-Quadriceps-Tibialis Anterior or Paraspinals-Hamstrings-Gastrocnemius). This pattern was termed a "hip" synergy. A "mixed" strategy is accomplished by a combination of the proximal-to-distal muscle activation of the hip muscles followed by the distal-to-proximal activation of the lower leg muscles produces the "mixed" strategy. Therefore, the "mixed" synergy reflects a

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combination of the "ankle" and "hip" synergies. Muscle synergies associated with lateral perturbations are more variable than AP disturbances. In general, however, the "hip load/unload" strategy is accomplished through a proximal-to-distal sequence of activation of the muscles of the head, hip, and ankle (Rietdyk et al., 1999; Winter et al., 1993).

4. Anticipatory Postural Control

Daily activities require voluntary movements. Many movements, such as reaching or leaning, threaten balance because the position of the COM is shifted toward the direction of the intended movement and toward the limits of stability. However, while these deliberate movements disturb balance, they rarely result in a loss of balance. Research by Belen'kii and collaborators (1967) investigated muscle activations during voluntary movements and found that prior to any movement, there was an activation of trunk and lower extremity muscles involved in postural control. This highly prescribed pattern of muscle activation, termed a postural synergy, always occurs in a distal-to-proximal order (Frank & Earl, 1990; Cordo & Nashner, 1982; Bouisset & Zattara, 1981; Belen'kii, Gurfinkel, & Paltsev, 1967). Subsequent to this preparatory phase, termed the anticipatory postural adjustment (APA), there is activation of the muscles necessary for the movement (i.e. the focal or prime mover muscles). Further research by Cordo and Nashner (1982) revealed that the observed proactive postural synergies are the same postural synergies used in reactive balance tasks.

Although the sequencing of muscle activation in the APA is fixed (Bouisset & Zattara, 1981), other characteristics of the APA are more variable (Toussaint, Michies, Faber, & Commissaris, 1998; Aruin, Forrest, & Latash, 1998; Horak et al., 1997; Inglin & Woollacott, 1988; Lee & Deming, 1987; Brown & Frank, 1987; Horak, Esselman, Anderson, & Lynch, 1984; Cordo & Nashner, 1982). Research has found that the magnitude and onset

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magnitude and onset of the APA are dependent upon a number of factors (Horak et al., 1997; Inglin & Woollacott, 1988; Lee & Deming, 1987; Brown & Frank, 1987; Horak et al., 1984; Cordo & Nashner, 1982). The most determinant factor in APA characteristics is the nature and goal of the task. Different voluntary movements require the use of different muscles of varying amplitudes and latencies. For example, the type of voluntary movement performed determines the muscles active in the APA. Arm extensions require preparatory activation of the posterior leg muscles while rise-to-toes tasks require preparatory activation of the anterior leg muscles. Another example of a task-dependent factor that alters APAs is the level of difficulty imposed by the task. Thus, the kind of task may require different magnitudes and onsets of preparatory muscle activation. For example, when lifting objects, earlier onsets and larger amplitudes of postural muscle activity are necessary to lift heavier objects (Horak et al., 1997; Lee & Deming, 1987; Horak et al., 1984; Cordo & Nashner, 1982).

From these studies, researchers have suggested that APAs play a stabilizing role on balance during voluntary movements. In particular, the contraction of these postural muscles function to prepare the body for the upcoming disturbance associated with the desired movement (Woollacott, 1989; Bouisset & Zattara, 1981; Marsden, Merton, & Morton, 1977). The CNS executes these synergies to generate stabilizing moments that will oppose the upcoming destabilizing moments generated by the focal movement. These moments cause displacement of the COP in the direction opposite to the intended movement. Researchers speculate that these APAs serve to control the movement of the COM within the BOS, reducing the probability of a loss of balance. For example, prior to forward reaching, the posterior calf muscles will contract to limit forward displacement of the COM. Following from the example, the upper arm and shoulder muscles will contract to produce a forward

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reach. As predicted, this action directs the COM anteriorly. Balance is maintained because the CNS has made prior postural adjustments (i.e. a backward shift) to the COM to accommodate for the upcoming anterior shift in the COM produced by the forward reach. Often, the latency between the onset of the postural and focal muscles is calculated and used as a measure of the amount of processing required to prepare for the movement.

The relationship between the COM and the COP during voluntary movements is similar to quiet stance. Figure 1.6 illustrates the COM-COP relationship in a quiet standing and in a leaning trial in a healthy younger adult. In the leaning trial, the adult performed a forward lean followed immediately by a backward lean and a return to a position of quiet stance. Although the magnitude of the COM and COP displacements was larger during the leaning trial, the overall pattern of neural control remained the same. The CNS always exacts control over the COM through the execution of postural muscles that prevent it from exceeding the BOS.
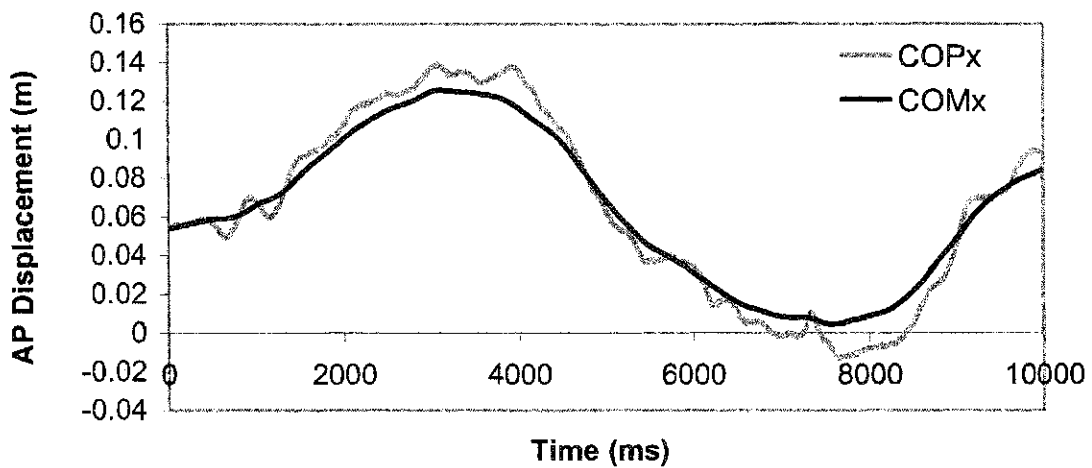
Figure 1.6: The relationship between the COP and COM in the AP direction of a 21 year old female participant during a feet together, forward lean (held for 2s) followed immediately by a backward lean (held for 2s). Note that the COP controls the movement of the COM.

5. Systems of Postural Control

The Systems Model for postural control stipulates that multiple systems operate cooperatively for the purpose of maintaining balance (Shumway-Cook & Woollacott, 2001). Three systems coordinate to achieve postural stability: the sensory system, the CNS, and the musculoskeletal system. The sensory systems provide the CNS with information regarding the state of the internal and external environment. The CNS processes information from the sensory systems and integrates this information with information from other regions of the brain to orchestrate a suitable response to correct for imbalance. The musculoskeletal system is responsible for executing the CNS commands through the generation of muscle forces that must be correct in amplitude, duration, and latency.

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a) <u>Sensory Systems</u>

The sensory systems are responsible for supplying the CNS with information regarding the static and dynamic states of the body relative to the external environment (Shumway-Cook & Woollacott, 2001). This information is derived from the somatosensory, visual, and vestibular sensory systems. Each sense relays a different type of information to provide the CNS with an accurate 3D internal representation of the body in space.

(1) Somatosensory System

The somatosensory system provides information regarding the state of the musculoskeletal system and the external environment. This information is gained from muscle, joint, and cutaneous receptors located throughout the periphery of the body (Horak et al., 1990). Muscle receptors consist of muscle spindles and Golgi tendon organs (GTOs) (Gordon & Ghez, 1991). Muscle spindles detect changes in muscle length as well as the rate of length change. GTOs are encapsulated endings located at the junction between the muscle and the tendon that report changes in muscle tension. Joint receptors include Ruffini-type endings, Paciniform endings, ligament receptors, and free nerve endings that are sensitive to joint movement and stress (Shumway-Cook & Woollacott, 2001). Although their function is not yet determined, research has suggested a role for joint receptors in injury prevention and/or joint position detection (Burgess & Clark, 1969). Cutaneous mechanoreceptors detect tactile changes in the environment and in the position of the body relative to the environment. There are many types of cutaneous receptors each sensitive to a specific type of stimuli. Pacinian corpuscles respond to vibration, Meissner's corpuscles to light touch and vibration, Merkel's discs to pressure, and Ruffini endings to stretch (Shumway-Cook & Woollacott, 2001).

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Researchers have used a number of techniques to investigate the importance of the somatosensory system for balance. These paradigms are based on the premise that somatosensory information from the feet can be removed through an ischemic block at the level of the ankle. Similarly, this information can be made unreliable through the use of a compliant support surface, vibration of lower leg muscle tendons, or rotation of a movable platform in a toes-up or toes-down rotation. Regardless of the technique, the manipulation of somatosensory input results in increased amplitude and variability of postural sway (Diener & Dichgans, 1988; Diener, Dichgans, Guschlbauer, & Mau, 1984; Nashner, 1982). The somatosensory system is crucial for the production of a "ankle" strategy because removal of input results in the use of a "hip" strategy regardless of perturbation size (Forssberg & Hirschfeld, 1994; Horak et al., 1990; Wolfson et al., 1986; Woollacott et al., 1986).

(2) Vision

The visual system relays information regarding the position and the motion of the body with respect to surrounding objects. Thus, the visual system is responsible for information regarding self-motion. Visual acuity, or the ability to clearly see objects, is crucial to fall prevention because identification of objects in the visual field decreases the probability of a slip or trip.

Investigators have developed many methods for manipulating visual inputs. Two frequently used techniques are eye occlusion or the use of a visual surround. Occlusion of the eye can be accomplished through eye closure, blindfolds, or specially designed opaque or translucent goggles that can be triggered to block or distort vision. A visual surround involves the movement of the external environment relative to the body. Lee and Lishman

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(1975) conducted the first visual surround experiment where the "swinging room" was found to induce postural sway in the same direction as the movement of the room. Adults responded like "visual puppets" where they unknowingly move their body in correspondence with the movement of the room (Lee & Lishman, 1975).

A study by Manchester and colleagues (1989) compared the importance of inputs from the peripheral and the foveal visual systems to determine whether both systems contribute equally to the visual control of posture. In this experiment, the visual system was manipulated to restrict vision to one visual subsystem at a time. In the absence of peripheral vision, postural sway was increased significantly. In contrast, the absence of focal vision did not change postural sway. In fact, the presence of peripheral vision was almost as stabilizing as full visual field input. These results suggest that the peripheral visual system is more important than the foveal visual system to postural control (Manchester, Woollacott, Zederbauer-Hylton, & Marin, 1989). Recent research has provided further support for the significance of peripheral vision to the control of balance during upright stance (Nougier, Bard, Fleury, & Teasdale, 1997).

(3) Vestibular System

The vestibular system provides information about the position and movement of the head with respect to gravity. The semicircular canals and the otolith organs are the two vestibular structures involved in postural control. The semicircular canals consist of three orthogonally arranged canals that detect angular accelerations of the head, and the otoliths, composed of the saccule and the utricle, sense linear position and acceleration of the head.

Although there are many ways to manipulate the somatosensory and visual sensory components, modifications to vestibular inputs are much more limited. Studies focusing on

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vestibular deficits either use patients with bilateral peripheral vestibular loss or alter the input through galvanic stimulation or through the performance of a head tilt (Horak & Shupert, 1994; Horak et al., 1990). The disruption of accurate vestibular inputs alone does not affect postural stability; however, when misleading information is received from the visual and somatosensory systems, the vestibular system does provide an absolute frame of reference that enables the maintenance of balance (Horak & Shupert, 1994; Horak et al., 1990; Diener & Dichgans, 1988; Diener, Dichigans, Guschlbauer, & Bacher, 1986; Nashner, 1982). Horak and colleagues (1990) made an interesting finding when they investigated postural responses to external perturbations among healthy participants and patients with bilateral vestibular loss. During small perturbations, both groups executed an "ankle" strategy to recover balance. However, when large perturbations were delivered, healthy adults shifted to a "hip" strategy while patients with bilateral vestibular loss continued to perform the "ankle" strategy. In many cases, the "ankle" strategy was insufficient to generate the necessary moments to recover balance; therefore, patients with bilateral vestibular loss often lost their balance. Further research confirmed that vestibular input to be essential for executing a "hip" strategy (Forssberg & Hirschfeld, 1994).

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(4) Redundancy of Sensory Information

The redundancy of sensory information provides the CNS with an internal representation of the position and movement of the body and the external environment (Horak & Shupert, 1994). This 3D schema is constructed through orientation information from horizontal, vertical, and gravitoinertial frames of reference supplied by the somatosensory, visual, and vestibular systems respectively (Shumway-Cook & Woollacott, 2001). Because each system contributes information regarding postural control, researchers have questioned whether all three sensory system inputs were necessary for normal function. To examine this question, Nashner (1982; 1976) developed a laboratory test that assessed sensory contributions through the systematic manipulation of input from each sense. This protocol has since been adapted in clinical settings and described as the Sensory Organisation Test (SOT) (Shumway-Cook & Horak, 1986). In the SOT, the inputs from each sense may be made inaccurate (e.g. using a visual moving surround or a compliant surface) or removed completely (e.g. eyes closed or ischemia-blocked ankles). Each sense can be examined independently or in conjunction with one or more of the other two sensory deficits. Results from studies provided evidence that the CNS does not require the proper function of all three sensory systems (Diener & Dichgans, 1988). In fact, when one system is compromised, the CNS can fully compensate for the loss through input from the other two senses (Horak et al., 1990; Diener et al., 1984; Nashner, 1982). However, if more than one system fails, the system shows deficits, and postural stability is threatened (Horak et al., 1989; Diener & Dichgans, 1988; Diener et al., 1986). Figure 1.7 demonstrates the increase in postural sway associated with decreased sensory information in a healthy younger adult.
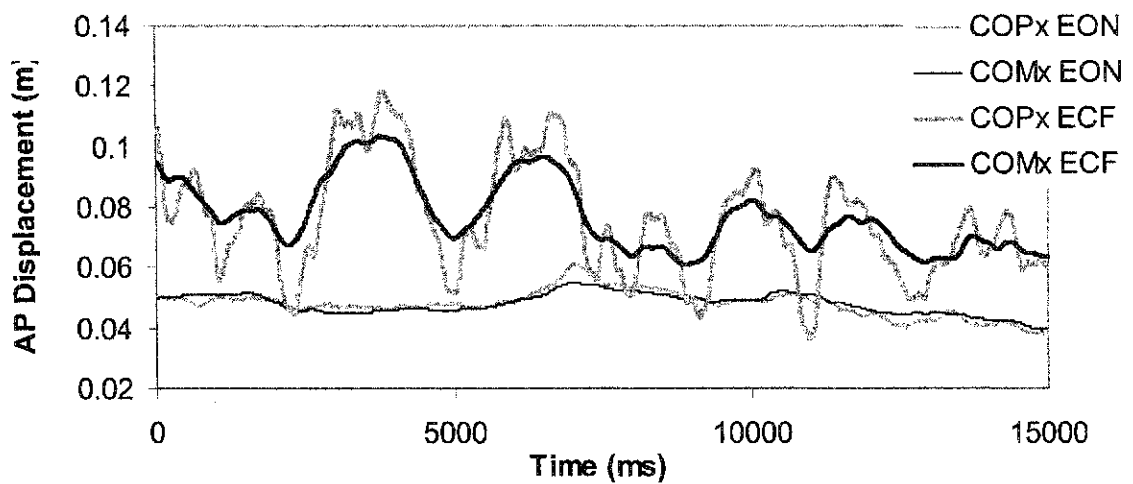
Figure 1.7: The relationship between the COP and COM in the AP direction of a 21 year old female participant during 15s of feet together, eyes closed quiet standing on a foam surface (ECF) versus quiet standing with eyes open on a normal surface (EON). Note that greater sway occurs in the ECF conditions compared to the EON.

However, the contributions from each sensory input are not weighted equally. When the CNS combines all the information, it assigns different weights to the contributions of each sense dependent upon the condition of the system. The SOT enables researchers to investigate the prioritization of inputs under different situations of postural control. Results from this research indicate that the somatosensory system has the largest influence on the CNS (Fig. 1.7) (Shumway-Cook & Woollacott, 2001; Baloh et al., 1994; Colledge et al., 1994; Wolfson et al., 1992; Nashner & Woollacott, 1979). However, when the somatosensory input is inappropriate or absent, the visual and vestibular systems are able to compensate. Therefore, the CNS shows flexibility in the hierarchy of sensory inputs. If the correct input is restored, the CNS reassigns the original contribution weights (Hay et al., 1996; Teasdale et al., 1991).

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b)   Central Nervous System

The primary function of the CNS in postural control is the integration of sensory and motor systems. In this process, the CNS organises and processes information from the senses and directs musculoskeletal responses. In the CNS, the control of movement obeys the principles of organisation. It is arranged hierarchically such that more complex tasks are produced and regulated in higher CNS centres (Nicholls, Martin, Wallace, & Fuchs, 2001).

Much of the incoming somatosensory information from muscle, joint, and cutaneous receptors is processed at the spinal cord level; however, information regarding the trunk and limbs is also transmitted to the sensory cortex and the cerebellum via two major afferent pathways: 1) the dorsal column - medial lemniscal system and 2) the anterolateral system (Heimer, 1995).   The dorsal column pathway is primarily responsible for information regarding muscle, tendon, and joint sensitivity and is particularly important for fine touch discrimination. Although the anterolateral pathway also relays information on crude touch, it specialises in thermal and nociceptive detection. Visual information travels via the optic nerve through the lateral geniculate nucleus of the thalamus and arrives at the primary visual cortex and higher order visual cortex (Nolte, 1988).   Other CNS targets for visual information include the superior colliculus and the pretectal region (Nolte, 1988). Vestibular information from both the otoliths and the semicircular canals is transmitted via the vestibulocochlear cranial nerve to the base of the medulla (Heimer, 1995). The vestibular nucleus is subdivided into four parts (i.e. medial, lateral, superior, and inferior vestibular nuclei) which project to the cerebellum, the reticular formation, the thalamus, and the cerebral cortex (Heimer, 1995).

Once all of the sensory information has been received and processed by the CNS, appropriate actions are relayed to the musculoskeletal system via descending systems. The

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descending systems conform to a hierarchy that consists of three main levels: 1) spinal reflexes, 2) interneuronal networks of the brainstem, and 3) higher order neurons in the cerebellum, cerebral cortex, and basal ganglia (Nicholls, et al., 2001).

Spinal reflexes are fast, automatic responses that do not require supraspinal input to function. Afferent fibres from the muscle spindle propagate to the spinal cord where they make direct or indirect (via interneuronal) connections to efferent fibres. These spinal reflexes operate through reciprocal innervation where motor neurons excite agonist muscles to contract and inhibit antagonist muscles from contracting (Ghez, 1991). One such example of a spinal reflex is the 'flexor reflex'. When nociceptors detect pain, there is an activation of the flexor muscles and an inhibition of the extensor muscles of the affected limb that results in withdrawal from the painful stimulus.

Also, the spinal cord is thought to be responsible for producing coordinated movements such as gait (Nicholls et al., 2001). This hypothesis is supported by experiments performed on cats with transected spinal cords (Rossignol, Chau, Brustein, Belanger, Barbeau, & Drew, 1996). When their body weight is supported while exposed to a treadmill, the cats are still capable of locomotion at various speeds. This research has been extended to human spinal cord injuries where similar results have been found (Wirz, Colombo, & Dietz, 2001; Dietz, Wirz, & Jensen, 1997). In fact, treadmill-walking or locomotor training may improve the potential recovery of walking in these patients (Wirz et al., 2001; Behrman & Harkema, 2000; Van de Crommert, Mulder, & Duysens, 1998; Dietz et al., 1997).

The descending systems of the brainstem play an important role in the maintenance of upright posture. Although there are a number of descending pathways responsible for the control of movement, only two pathways play a major role in the control of posture: 1) the lateral vestibulospinal tract and 2) the lateral reticulospinal tract (Ghez, 1991). As the

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name implies, the lateral vestibulospinal pathway originates in the lateral vestibular nucleus and descends through the spinal cord. Stimulation of this circuit produces reciprocal innervation of trunk and limb muscles, thereby increasing the muscle tone of the antigravity muscles (Heimer, 1995). This pathway plays a key role in maintaining erect posture and in reactive postural control. The second descending system is the lateral reticulospinal circuit that originates at the medullary reticular formation and projects down to the spinal cord (Heimer, 1995). The lateral reticulospinal tract has parallel functions with the lateral vestibulospinal tract. In fact, the vestibulospinal pathway can indirectly influence the spinal cord through connections with the reticular formation (Ghez, 1991). In addition to the maintenance of upright stance, the reticulospinal system is able to produce a wide range of coordinated movements as well (Heimer, 1995).

Motor control is accomplished by various higher order brain structures such as the cerebral cortex (primary motor cortex, premotor cortex, and supplementary motor area), the basal ganglia, and the cerebellum. Although all of these structures play a role in anticipatory postural control, particularly the planning and programming of voluntary movements, only the cerebellum is primarily responsible for the maintenance of upright stance. Different regions of the cerebellum are involved with different areas of postural control. The fastigial nucleus has connections to the vestibular nuclei and the reticular formation. Thereby, the cerebellum can exert a direct influence on the vestibulospinal and reticulospinal tracts responsible for the regulation of posture and reactive postural control (Mori, Matsui, Mori, Nakajima, & Matsuyama, 2000; Mori, Matsui, Kuze, Asanome, Nakajima, & Matsuyama, 1998). The flocculonodular lobe also projects to the vestibular nuclei, where it controls the axial muscles to maintain equilibrium control (Nolte, 1988). The lateral hemisphere nucleus projects to the cerebral cortex where it aids in the planning and execution of voluntary

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movements such as reaching (Ivry, Keele, & Diener, 1988). The vermis and the intermediate lobes function to correct postural deviations that may accompany voluntary movements (Dichgans & Fetter, 1993). These cerebellar regions also modulate muscle tone (Ghez, 1991). Because of the diverse functioning of the cerebellum, lesions may produce a wide range of deficits, dependent upon the location of the lesion within the cerebellum (Dichgans, & Fetter, 1993).

c) Musculoskeletal System

The musculoskeletal system executes the CNS commands for balance preservation. The skeletal system provides support for the weight of the body and acts as a site of attachment for the muscles. Muscles generate forces responsible for producing the stabilizing moments that control the movement of COM. The lower leg plantarflexors/dorsiflexors and the hip abductors/adductors are the main muscles used to control balance during upright stance. During anticipatory and reactive postural control, postural muscles often contract in centrally prescribed sequences or synergies. These synergies produce postural strategies that prepare or recover balance from impending disturbances. It is the effective and efficient contraction of the muscles that control balance.

C. Aging

Aging is a heterogeneous process that universally affects all humans. As we age, disease and injury become more prevalent (Horak, 1987). Injury and illness, such as cerebrovascular disease, Parkinson's disease, and other neurological disorders, impair balance abilities. However, in the absence of age-related diseases, healthy older adults also demonstrate deficits in postural control. High incidences of falls in the elderly imply that older adults have poorer balance and stability than younger adults (Sattin, 1992).

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Falls are a primary aetiology of morbidity and mortality in older adults. Approximately 30 to 50 percent of older adults, aged 65 years and over, experience a fall at least once a year (Tinetti & Williams, 1997; Suzuki et al., 1997). Falls lead to inactivity, injuries, and death, in addition to a loss of confidence and independence (Coni et al., 1992). Current demographic trends indicate an increasing aging population accompanied by increased life expectancies (Statistics Canada, 1999; Prince, Corriveau, Hebert, & Winter, 1997; Tideiksaar, 1997; Winter, 1995). In fact, Canada's elderly population is projected to rise from 12.3 percent of the total population in 1998 to 22.6 percent by the year 2031 (National Advisory Council on Aging, 1999; Northcott, 1997). These trends stress the importance of fall prevention to maintain mobility in the elderly.

1. Aging and Postural Control

The age-related deterioration in postural control can be observed in both static and dynamic balance tests. Numerous posturographic studies have illustrated age-related changes to the typical postural sway characteristics (Simoneau et al., 1999; Perrin, Jeandel, Perrin, & Bene, 1997; Hill & Vandervoort, 1996; Baloh, Spain, Socotch, Jacoboson, & Bell, 1995; Colledge et al., 1994; Baloh et al., 1994; Baloh et al., 1994; Hytonen, Pykko, Aaalto, & Starck, 1993; Patla, Frank, & Winter, 1992; Maki, Holliday, & Fernie, 1990; Stelmach, Teasdale, DiFabio, & Phillips, 1989; Hayes, Spencer, Riach, Lucy, & Kirshen, 1984; Overstall, Exton-Smith, Imms, & Johnson, 1977; Sheldon, 1963) (Fig. 1.8). These age-related changes include increases in the amplitude and frequency of postural sway in the AP and ML dimensions. Older adults have a larger magnitude, speed, area, and variability of sway than younger adults (Simoneau et al., 1999; Perrin et al., 1997; Hill & Vandervoort, 1996; Baloh et al., 1995; Colledge et al., 1994; Baloh et al., 1994; Baloh et al., 1994; Hytonen

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et al., 1993; Patla et al., 1992; Maki et al., 1990; Stelmach et al., 1989; Overstall et al., 1977; Sheldon, 1963). Furthermore, the frequency range and mean power frequency of sway is higher in older compared to younger adults (Tang & Woollacott, 1996; Patla et al., 1992; Maki, Holliday, & Topper, 1990; Hayes et al., 1984). These changes suggest older adults have difficulty controlling their sway because the COM is allowed to drift further toward the limits of stability, thus requiring larger stabilizing moments to be generated in order to maintain upright stance. This hypothesis is reinforced by studies investigating the postural sway of older fallers and nonfallers. Older fallers show even larger discrepancies in the measures of postural sway compared to nonfallers and younger adults (Vellas, Wayne, Romero, Baumgartner, & Garry, 1997; Tang & Woollacott, 1996; Baloh et al., 1995; Baloh et al., 1994). Furthermore, fallers demonstrate a disproportionately larger impairment in the ML compared to the AP direction (Maki & McIlroy, 1996; Maki, Holliday, & Topper, 1994). This finding is of particular consequence to older adults because research suggests that increased postural sway in the ML direction may be more threatening to balance because many falls in the elderly involve a lateral movement (Maki et al., 1994).
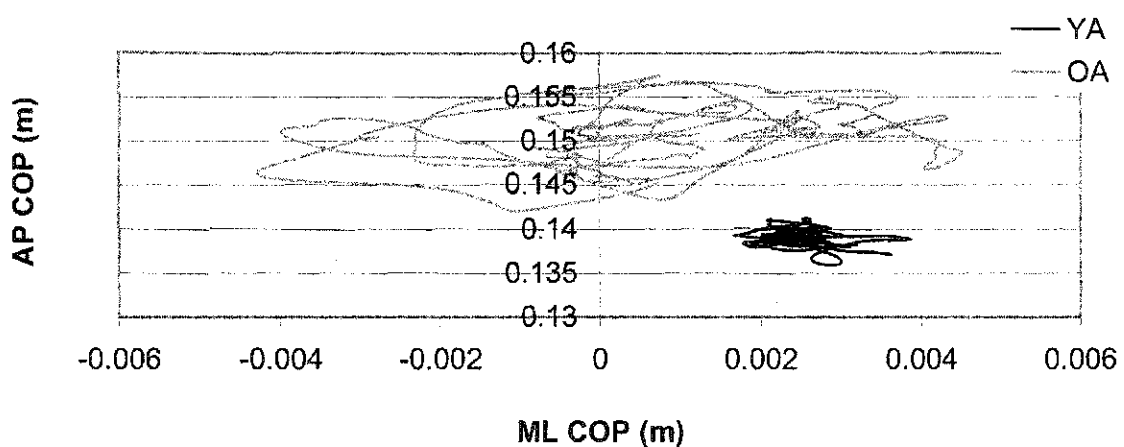
Figure 1.8: Stabilogram of a 25 year old male (YA) and a 75 year old male (OA) during 15s of quiet standing with eyes open. Note that older adults show greater AP and ML postural sway than younger adults.

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Reactive balance tests have revealed important age-related neuromuscular modifications to postural control. Past research using the translating platform paradigm has established significant differences in the temporal and spatial organisation of postural synergies between younger and older adults (Tang & Woollacott, 1996; Woollacott et al., 1986). In older adults, onsets of muscle activation are delayed by approximately 7 ms in calf muscles and are even more delayed in thigh muscles, with reports as high as 29 ms later than younger adults (Woollacott et al., 1986). Furthermore, the sequence of activation is disordered (proximal-to-distal) in older adults (Tang & Woollacott, 1996; Studenski, Duncan, & Chandler, 1991; Woollacott et al., 1986). In addition, older adults exact a generalized muscle activation strategy in which extraneous muscles are activated to collaboratively respond to the disturbance (Manchester et al., 1989). Often, there is increased cocontraction of antagonist muscles that may result in increased joint stiffness and a reduced range of motion (Winter et al., 1998; Tang & Woollacott, 1996; Maki & McIlroy, 1996; Manchester et al., 1989). Overall, impaired timing, coordination, and magnitude of postural response exist in the elderly (Wolfson et al., 1992; Woollacott et al., 1986). Furthermore, these abnormal muscle activity patterns are more prevalent in older adults with a history of falling compared to those that have not fallen (Studenski et al., 1991).

In response to perturbations, older adults show preference for a different postural response strategy than younger adults. In response to low magnitude perturbations, younger adults typically select the ankle strategy while older adults utilise the hip strategy (Manchester et al., 1989; Horak et al., 1989). Thus, older adults flex/extend at the hip joint generating large, potentially dangerous, horizontal shear forces in an effort to maintain balance (Manchester et al., 1989). When permitted to use "feet-in-place" or "change in support"

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strategies, older adults are more likely to adopt a compensatory stepping mechanism than younger adults regardless of perturbation size (Hall & Jensen, 2001; Cumming, Salkeld, Thomas, & Szonyi, 2000; Maki & McIlroy, 1996). Interestingly, this stepping strategy occurs even when the COM-BOS relationship suggests that balance recovery could be accomplished without stepping. Jensen and colleagues (2001) have suggested that older adults may not be as adept at attenuating perturbation-induced accelerations in segments cranial to the hip joint compared to their younger counterparts. Therefore, older adults experience larger horizontal accelerations of the head that may cause the perception of a larger-amplitude perturbation than actually exists. Furthermore, although both younger and older adults respond to AP perturbations with a single step in the sagittal plane, older adults take additional steps in the frontal plane (Maki & McIlroy, 1996; McIlroy & Maki, 1993). Maki and McIlroy (1996; 1993) hypothesise that it is these lateral components of sway that challenge stability in older adults. In fact, it may be the difficulty in the control of the ML load/unload mechanism following external perturbations that causes the high incidence of falls in the elderly.

Not only do the elderly differ in the ways they recover balance from a disturbance, they also show differences in the ways they prepare for a balance disturbance. Onset latencies of postural and focal muscles are delayed in older adults compared to younger adults (Woollacott, 1989), and the latency between the onset of the postural and the onset of the prime mover muscles in older adults is longer. This finding suggests that more time may be required to stabilize the body for the movement. Researchers suspect that although older adults are fully capable of performing most voluntary movements, more time to complete the task is required compared to younger adults (Alexander, 1994). To investigate this hypothesis, Man'kovskii and colleagues (1980) instructed younger and older adults to flex

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one leg at the knee while using the other leg for support at self-selected and fast movement speeds. The findings revealed an increase in the onset latency of the postural muscles and a subsequent decrease in the onset latency between the postural and focal muscles. These results suggest that when rushed, older adults may be at risk for falling as they perform a task prior to adequately accommodating for the impending disturbance.

When reaching or leaning, older adults restrict performance based on their perceptions of stability. Functionally, we are unable to make use of our entire dimensions of the BOS because the CNS only permits displacement of the COM within a certain range of the BOS. This prescribed area is termed the Functional Base of Support (FBOS), and the unusable portion of the BOS is termed the safety margin. In younger adults, the FBOS accounts for approximately 60 percent of the AP dimension of the BOS; with advancing age, the FBOS decreases to approximately 40 percent, concomitantly increasing safety margins in the elderly (Kozak, Ashton-Miller, & Nyquist, 1997; King, Judge, & Wolfson, 1994; Blaszczyk, Lowe, & Hansen, 1994; Blaszczyk et al., 1994; Duncan et al., 1990; Lee & Deming, 1988; Murray, Seireg, & Sepic, 1975). It is speculated that this age-related decline in FBOS may be the result of decreased confidence in balance ability (Robinovitch & Cronin, 1999). This strategy may serve as a protective mechanism that minimises the displacement of the COM within the BOS and reduces the likelihood that the COM will exceed the limits of stability.

2. Age-related Changes in the Systems of Postural Control

Although the cause of falling is complex and multifaceted, research has established the age-related deterioration in postural control as a primary cause. Older adults incur deficits in all systems of postural control. This section provides an overview to the

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anatomical and physiological changes in the sensory, CNS, and motor systems (Woollacott, 2000; Maki & McIlroy, 1996; Alexander, 1994).

a) Age-related Changes in the Sensory Systems

The detection of accurate sensory information decreases with age. These decrements are present in all three sensory subsystems: the somatosensory, visual, and vestibular systems; however, the influence of sensory system function to the postural instability in the elderly remains unclear (Woollacott, 2000; Hay et al., 1996; Teasdale et al., 1991).

(1) Somatosensory Systems

Age-related changes in the somatosensory system include increased thresholds of vibratory and tactile sensitivity, particularly in the lower limbs (Kenshalo, 1986; Skinner, Barrack, & Cook, 1984; Brocklehurst, Robertson, & James-Groom, 1982). The decline in these functions is evidenced by a decrease in the quantity and quality of Meissner end organs and Pacinian corpuscles involved in vibratory sensation and a decrease in the fine touch and pressure sensation responsible for cutaneous sensitivity (Potvin, Syndulko, Tourtellotte, Lemmon, & Potvin, 1980). There is a functional loss of mechanoreceptors and a loss of up to 10 percent of the sensory fibres resulting in peripheral neuropathy (Kenshalo, 1986) and an increased reliance on the other two sensory inputs, particularly vision (Sundermier, Woollacott, Jensen, & Moore, 1996; Colledge et al., 1994; Teasdale et al., 1991). Figure 1.9 illustrates the effect of somatosensory loss on the postural sway of a healthy older adult.

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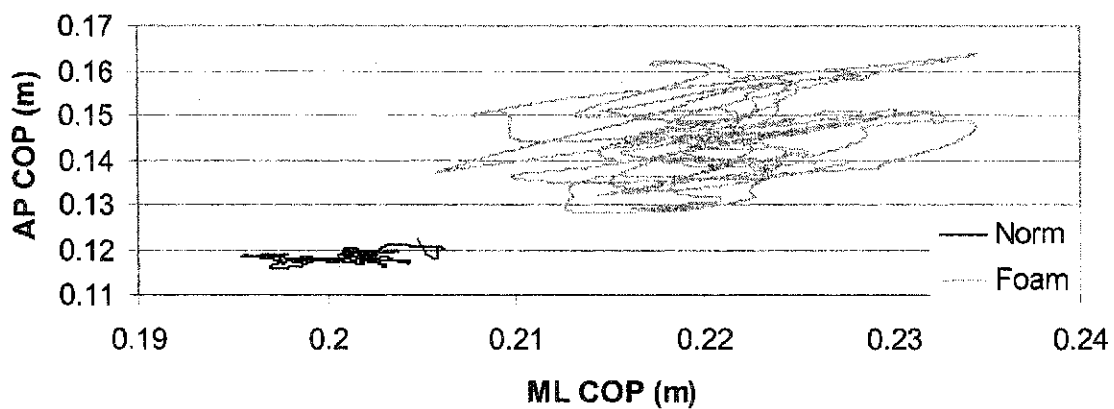
Figure 1.9: Stabilogram of a 77 year old male during 15s of quiet standing on a normal surface (Norm) and a compliant foam surface (Foam). Note that greater sway occurs in the foam compared to the normal support surface conditions.

(2) Vision

Older adults rely heavily on vision for balance (Gill et al., 2001; Perrin et al., 1997; Sundermier et al., 1996; Hill & Vandervoort, 1996; Hytonen et al., 1993; Teasdale et al., 1991; Ring, Matthews, Nayak, & Isaacs, 1988). The visual system experiences deficits in focal (foveal) and ambient (peripheral) vision with advancing age (Kosnik, Winslow, Kline, Rasinski, & Sekuler, 1988). The foveal visual system deficits involve reductions in visual acuity, visual field, depth perception, and contrast sensitivity at intermediate and high spatial frequencies; however, the sole presence of focal vision largely decreased postural stability among older adults (Manchester et al., 1989; Pitts, 1982). Age-associated changes in the peripheral visual system include decreased perceptual ability of motion, increased thresholds for self-motion detection, and decreased contrast sensitivity at low spatial frequencies (Tang & Woollacott, 1996; Paulus, Straube, & Brandt, 1984). When only peripheral vision is permitted, older adults stabilized sway almost as much as when full visual field feedback was provided (Manchester et al., 1989). Thus, peripheral visual field is crucial to stabilisation in

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the elderly, and the decrements in low frequency contrast sensitivity is of critical importance to balance as it reduces the ability to detect and discriminate obstacles in the environment (Alexander, 1994; Manchester et al., 1989). Figure 1.10 depicts the increase in the postural sway of a healthy older adult when vision is compromised.
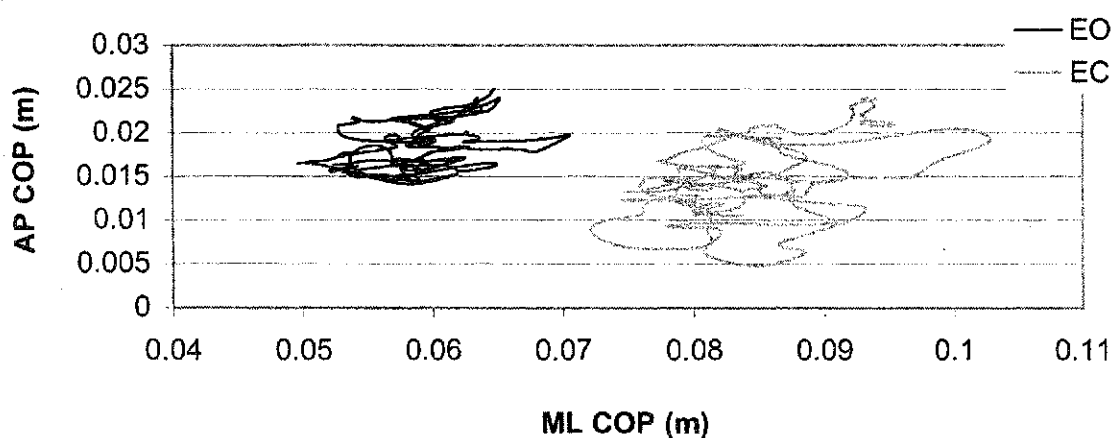
Figure 1.10: Stabilogram of a 73 year old male during 15s of quiet standing with eyes open (EO) and eyes closed (EC). Note that greater sway occurs in the eyes closed versus the eyes open conditions.

(3) Vestibular System

With aging, there is a progressive loss of up to 40 percent of labyrinthine hair cells, vestibular ganglion cells, and nerve fibres in the vestibular system (Sloane, Baloh, & Honrubia, 1989; Rosenhall, 1973). Researchers still debate the importance of age-induced vestibular deficits to postural control. The removal of vestibular input shows little effect on balance in sensory organization tests, and therefore, suggests it is not a prominent factor associated with falls (Alexander, 1994; Brocklehurst et al., 1982).

Results of sensory organization studies have also demonstrated an important role for the vestibular system in the modulation of sensory inputs (Nashner, 1982). When

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conflicting sensory inputs are provided, older adults demonstrate significant impairment in balance performance (Woollacott et al., 1986). This finding suggests that age-associated decreases in vestibular system function may result in decreased stability not because of poor detection of sensory information, but because of a reduced ability to modulate the other sensory inputs.

b) Age-related Changes in the Central Nervous System

Aging causes progressive, universal, and irreversible impairments in the integrative processing of the CNS. Functionally, the elderly show a general slowing of information processing and decreases in nerve conduction velocity of the peripheral nerves to the CNS (Maki & McIlroy, 1996; Stelmach et al., 1989; Stelmach & Worringham, 1985). Anatomically, age-related changes in the CNS constitute a loss of neurons and dendrites, axonal degeneration, and reduced dendritic branching (Schaumberg, Spencer, & Ochoa, 1983). Physiologically, there is impaired cerebral metabolism, reduced cerebral perfusion, and altered transmitter metabolism (Lipsitz & Goldberger, 1992).

Many researchers believe that decreased CNS functioning, not peripheral neuropathy, causes decreased postural stability in the elderly (Hay et al., 1996; Colledge et al., 1994). Teasdale and colleagues (1991) conducted an ingenious study to investigate this possibility. Conditions of sensory deprivation and reinsertion of sensory information were presented to younger and older adults. As predicted, the removal of correct sensory input resulted in decreased stability in younger and older adults. Interestingly, the balance of older adults was disturbed by the reinsertion of sensory input as well. Further research has substantiated these findings, theorizing that a slowing of central integrative processing and

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not sensory system deterioration is the cause of postural instability in the elderly (Teasdale & Simoneau, 2001; Hay et al., 1996; Teasdale et al., 1991).

c) Age-related Changes in the Musculoskeletal System

Age-induced impairments in the musculoskeletal system decrease the ability to execute an effective and appropriate motor response. The three most prominent age-associated changes in the musculoskeletal system involve a reduced range of motion, a decrease in muscle strength, and a slowing of muscle contraction (O'Brien, Culham, & Pickles, 1997; Wolfson, Judge, Whipple, & King, 1995; Studenski et al., 1991; Iverson, Gossman, Shaddeau, & Turner, 1990; Aniansson, Grimby, Hedberg, Rundgren, & Sperling, 1978). As stated by Alexander (1994), there is "less strength available to move a stiffer joint through a more limited range of motion". The decreased range of motion suggests increased joint stiffness and is supported by recordings of EMG activity (Prince et al., 1997). The age-associated reduction in muscle strength (i.e. the amount of force a muscle can produce) is most dramatic in the lower extremities with decreases of up to 40 percent between 30 and 80 years of age (Lamoureux, Sparrow, Murphy, & Newton, 2001; Hurley, 1995; Aniansson, Hedberg, & Henning, 1986). In a study by Whipple and collaborators (1987), elderly fallers were unable to produce the same peak moment and power at the ankle and the knee as the nonfallers. This finding was substantiated by another study that revealed reduced muscle strength in older fallers compared to older nonfallers (Daubney & Culham, 1999). Whipple and colleagues (1987) also revealed that older adults had slower muscular responses to external perturbations compared to their younger counterparts. Research by Thelen and colleagues (Thelen, Ashton-Miller, Schultz, & Alexander, 1996) revealed that the delay in onset of moment generation was not due to decline in central processing function but most

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likely, a result of age-dependent changes in muscle contraction mechanics. As suggested by other researchers, these changes may be due to the age-related decrease in the size and number of muscle fibres and motor neurons (Vandervoort, 2002; Lexell, 1995; Grimby, 1995). Whipple and colleagues (Whipple et al., 1987) proposed that a predisposition to falling may be due to the combination of reduced muscle strength and the slow activation of lower extremity muscles.

It should be noted that deficits in muscular function are not the primary cause of postural instability in older adults (Brown, Sinacore, & Host, 1995). Neither upright standing nor perturbed standing require maximal muscle strength or a large range of motion (Maki & McIlroy, 1996; Alexander, 1994). In fact, older adults are well within their muscular capacity to recover balance. The fact that falls still occur implies that postural instability may be due, at least in part, to factors other than the age-related deficits observed in the musculoskeletal system. As forwarded by Heyley and colleagues (1998), the decline in muscle function may foster a fear of falling that could result in increased fall risk.

D. Falling

The age-related deteriorations in the systems of postural control have been identified as a chief source of postural instability. The reduced stability results in a higher fall-risk in the elderly (Vellas et al., 1997; Tang & Woollacott, 1996; Baloh et al., 1995; Baloh et al., 1994; Overstall et al., 1977). These falls are a frequent problem that can cause moderate to severe injuries such as hip fractures (Salkeld et al., 2000; Wilkins, 1999a; Tinetti, Speechley, & Ginter, 1988). Fall-related injuries may lead to long-term physical disabilities and possibly even death. In fact, 15 percent of those fallers who sustained fractures die each year (Manning, Neistadt, & Parker, 1997).

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1. Fear of Falling

Aside from the physical impacts of falling, there are psychological consequences as well. Many fallers, as well as many nonfallers, suffer from anxiety of falling again (Cumming et al., 2000; Chandler, Duncan, Sanders, & Studenski, 1996; Downton & Andrews, 1990; Silverton & Tideiksaar, 1989). This anxiety has been termed fear of falling or ptophobia (Bhala, O'Donnell, & Thoppil, 1982). Fear of falling is a pervasive condition that may be more severe than the actual fall occurrence itself because it leads to reduced activity, diminished confidence, and ultimately, a complete loss of independence (Salkeld et al., 2000; Howland et al., 1998; Cumming & Nevitt, 1994; Timiras, 1994; Arfken, Lach, Birge, & Miller, 1994; Howland et al., 1993; Black, Maki, & Fernie, 1993; Grimley Evans, 1992; Tinetti et al., 1988; Murphy & Isaacs, 1982). These consequences have severe implications for the quality of life of older adults, especially considering recent findings that indicate older adults fear a loss of independence more than their own mortality (Salkeld et al., 2000).

2. The Effects of Fear of Falling on Postural Control

Postural instability is one prominent outcome of fear of falling (McAuley et al., 1997; Myers et al., 1996; Tinetti, Mendes de Leon, Doucette, & Baker, 1994). Past research has used a clinical approach to the investigation of postural control in elderly adults who have a fear of falling. Classic balance testing involved performance on proven balance scales such as the Get Up and Go test, the Performance Oriented Assessment of Mobility, the Berg Balance Scale, the Functional Reach, the Falls Efficacy Scale, and the Activities Specific Balance Confidence Scale (Nakamura, Holm, & Wilson, 1998; Myers et al., 1996). In some of these studies, balance performance was further quantified through static or dynamic posturography. The results indicated that older adults with a fear of falling perform poorer

on these balance scales compared to older adults without a fear of falling (McAuley et al., 1997; Franzoni et al., 1994; Tinetti et al., 1990). Fearful older adults also demonstrated increased magnitude and velocity of postural sway compared to their nonfearful counterparts (Baloh et al., 1995; Baloh et al., 1994).

Due to the ethical constraints, the effect of fear of falling on the ability to overcome challenging postural tasks is difficult to quantify. Without a measure of fear, it is unclear whether these older adults actually felt anxiety about maintaining their balance. Thus to explore the effect of anxiety or arousal, Maki and Whitelaw (1993) exposed healthy, younger adults to low, moderate, and high forward and backward perturbations to investigate the effects of expectation and arousal on postural response. Galvanic skin conductance (GSC) and heart rate were used to measure physiological arousal. GSC and heart rate are standard measures of arousal because they can be used to describe changes in the sympathetic autonomic nervous system (ANS) in response to anxiety (Critchley, 2002; Ashcroft, Guimares, Wang, & Deakin, 1991). For example, GSC measures the conductive properties of the skin that change in response to the amount of perspiration being secreted onto the surface of the skin (Critchley, 2002; Critchley, Elliott, Mathias, & Dolan, 2000; Boucsein, Baltissen, & Euler, 1984). Heart rate measures the number of heartbeats per minute. Under heightened levels of arousal, the ANS causes increases in heart rate and increases in sweat secretion, corresponding to increases in GSC levels (Kettunen, Ravaja, Naatanen, Keskivaara, & Keltikangas-Jarvinen, 1998). Previous research has found these measures of arousal to be correlated with anxiety (Kettunen et al., 1998; Ashcroft et al., 1991). The testing protocol involved the delivery of low or high perturbations following the presentation of a visual cue that indicated whether a low, high, or random perturbation would occur. Although information regarding the direction of the perturbation was never

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given prior to the trial, participants leaned forward prior to each perturbation suggesting forward platform translations that induce backwards sway, were more challenging to balance recovery. This preparatory leaning mechanism was more prominent prior to a larger or a random balance disturbance. The results revealed significant positive correlations between arousal and perturbation size and between arousal and degree of anticipatory response (i.e. forward leaning). Although these findings suggest a relationship between fear of falling and postural accommodations, no cause-effect relationship between arousal and proactive responses can be deduced from this experiment. In other words, it is impossible to discern whether anxiety caused forward leaning or whether forward leaning caused increased anxiety.

To overcome this limitation, Maki and McIlroy (1997) conducted another experiment that investigated the effects of attention and arousal on the postural control of healthy, younger adults during static stance. In this experiment, younger adults were asked to perform secondary cognitive tasks that specifically affected attention, arousal, or both while maintaining upright stance. The results revealed that arousal was positively associated with anticipatory adjustments, supporting previous findings by Maki and Whitelaw (1993). Maki and McIlroy (1997) also discovered evidence of stiffness control as demarcated by increased coactivation of postural lower leg muscles (i.e. Tibialis Anterior m. and Gastrocnemius m.) during conditions of heightened arousal. These findings suggest a fear of falling may cause stiffness that limits range of motion about the ankle joint. This experiment utilised a moderate arousal-evoking task that did not threaten balance; therefore, it is unclear whether these results would be replicated if manipulations of postural threat rather than cognitive tasks were used to increase arousal.

Although the previous studies induced anxiety (Maki & McIlroy, 1997; Maki & Whitelaw, 1993), the tasks did not threaten balance in a manner similar to fear of falling.

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Therefore, Maki and researchers (Maki, Holliday, & Topper, 1991) attempted to introduce a fear of falling into older adults through the progressive increase in balance task difficulty. Older adults were asked to maintain balance on two force plates during static and dynamic balance conditions with eyes open and closed. Older adults with a fear of falling had increased postural sway in the static eyes closed trial and significantly poorer balance on one-legged trials than nonfearful older adults. This research is limited as older adults with and without balance problems were included in the analysis. Thus, underlying balance disorders present in the testing population may have confounded the results attributed to a fear of falling.

Although Maki and collaborators (1991) increased the threat to stability, the tasks still did not induce an anxiety similar to that produced by a fear of falling. To overcome this limitation, current research has focused on the control of posture under environmental contexts that alter the potential consequences of a loss of balance (Adkin et al., 2002; Brown et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997). These protocols require participants to stand at the edge of an elevated platform (Adkin et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997) or walk along an elevated narrow walkway (Brown et al., 2002). The underlying assumption is that these manipulations increase balance anxiety in a manner similar to a fear of falling.

Brown and Frank (1997) were the first experimenters to investigate postural control using this postural threat paradigm. Their purpose was to examine the effects of increased postural threat on postural recovery in healthy, younger adults. Participants were exposed to perturbations at the edge of a platform at two heights: low and high. The results revealed younger adults adopted proactive and reactive strategies that minimized the range of COM

displacement during conditions of increased postural threat. The observed anticipatory adjustments involved adopting a more conservative, i.e. posterior starting position during high height conditions. The threatening conditions also altered their reactive strategies by reducing the range of COM displacement and decreasing the time to reach peak COM velocity. These findings imply that younger adults adopt conservative postural strategies that maintain whole body movement within a smaller range and reduce the probability of a fall when the consequences of a loss of balance are most severe.

Following the work of Brown and Frank (1997), Carpenter and colleagues (1999) investigated the effects of increased postural threat on the postural control of healthy, younger adults during upright stance. Participants stood quietly in four conditions of threat: on a force plate located at the middle or the edge of a platform (indicating stepping constraint) set at a low or high height (i.e. low-unconstrained, low-constrained, high-unconstrained, high-constrained; LUC, LC, HUC, HC). Although there was no effect for constraint, manipulation of height did affect postural control. Under conditions of high postural threat, they observed changes in postural sway characteristics. These changes included reduced amplitude, reduced variability, and increased frequency of sway. Based on these findings, we may infer that fear of falling results in a more conservative posture. These findings support a model for stiffness control during increased postural threat.

To validate the proposed threat-induced stiffness strategy, Carpenter and researchers (2001) repeated the previous study but included EMG activity of the postural muscles and calculated a stiffness coefficient. The results confirmed the earlier hypothesis; under conditions of greatest postural threat, younger adults adopted a stiffness strategy as evidenced by the increase in the coefficient of stiffness. Furthermore, stiffness coincided with backward leaning strategies, increased frequency and decreased magnitude of postural

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sway, and reduced displacement of the COM. Consistent with a stiffness hypothesis, increased agonist/antagonist muscle cocontraction was also observed in the most threatening condition. Interestingly, however, regression analysis revealed that only increased activity of the Tibialis Anterior muscle could be correlated with increased stiffness.

Adkin and colleagues (2000) extended the work of Carpenter and collaborators (1999) to determine whether changes in the postural control of healthy, younger adults were scaled to the level of postural threat. Participants were asked to stand as still as possible on a force plate atop a platform set at three height conditions: low, medium, and high. In the most threatening condition, participants adopted a more posterior body position and reduced variability and increased frequency of postural sway. These postural accommodations changed linearly with respect to postural threat except body position, where a much larger posterior shift in body position was observed from medium to high conditions. These findings support previous research findings that report younger adults increase stiffness to achieve conservative postural adjustments that would serve well to reduce the likelihood of a fall during more threatening situations (Carpenter et al., 1999).

Although a fair body of knowledge regarding the effects of postural threat on postural threat among younger adults had developed (Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999), no studies had examined the potential effects of postural threat on anticipatory postural control. Furthermore, there was no research to substantiate whether the postural threat paradigm actually induced balance anxiety among participants. Thus, Adkin and coworkers (2002) examined the effects of postural threat on measures of physiological arousal and anticipatory postural control during the performance of a quasi-static rise-to-toes task among healthy younger adults. The results indicated that the conditions of greatest postural threat were significantly associated with increased levels of

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galvanic skin conductance (i.e. physiological arousal) and self-reported anxiety and were sufficient to produce conservative adaptations to balance. In particular, younger adults made similar modifications to their body position by adopting a backward lean in the most threatening condition. Furthermore, the rate and magnitude of COP displacement associated with anticipatory postural adjustments (APAs) were altered by postural threat. The authors suggest that the observed slower and smaller APAs were generated as a means to restrict the displacement and acceleration of the COM in an effort to reduce the risk of falling under threatening conditions.

Work to this point has demonstrated that the CNS demonstrates a cautious approach to balance control during static, proactive or reactive balance tasks; however, these studies have focused solely on the effects of postural threat on younger adults. These findings cannot be generalized to the elderly population whom are more susceptible to falling and a fear of falling. Thus, Brown and colleagues (Brown, Sleik, Polych, & Gage, 2002; Brown & Sleik, 2002) extended the work to date to investigate the postural control of older adults under threatening conditions. The testing protocol required participants to perform secondary cognitive tasks while maintaining upright stance on an elevating platform at four positions of threat: LUC, LC, HUC, and HC. Similar to previous studies (Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997), the results revealed that all participants adopted a more posterior body position with reduced area of sway under the most threatening condition. Thus, both younger and older adults exhibited similar conservative balance patterns under conditions of postural threat.

Brown and collaborators (2002) extended these static task results to the dynamic task of gait. The purpose of this research was two-fold: 1) to determine whether previous fear of falling results could be extrapolated to dynamic balance tasks, and 2) to establish whether

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younger and older adults responded to the postural threat in the same way. In this experiment, participants were asked to walk along pathways of varying postural threat. As in previous research, these walkways were manipulated by height (i.e. low or high) and constraint (i.e. wide or narrow pathway widths). Although all participants adopted a more conservative gait pattern in the most threatening condition as evidenced by slower gait velocities, decreased stride lengths, and increased double limb stance times, age differences were found in the joint kinematic and EMG data. Under the most threatening conditions, older adults reduced joint range of motion, particularly at the knee and hip and increased distal muscle activation of the calf muscles. These investigators hypothesise that these modifications conform to a stiffness strategy that may enable older adults to adopt a more cautious gait.

3. Conclusions

Falling is a prevalent and devastating problem in the elderly; however, falls can be prevented through the identification and treatment of risk factors that underlie imbalance. Past research has established a multitude of age-related deficits in all of the subsystems of postural control. For example, because of somatosensory deterioration, older adults are less able to perceive environmental conditions; thus, they should be advised against fluffy carpets that further reduce tactile sensitivity. Another example would be to advise older adults to engage in some form of exercise or physical activity to overcome the age-related decrements in muscle strength and flexibility. But older adults are not just susceptible to physical deficits, psychosocial factors such as fear of falling, also increase the risk for falling. Although many researchers have established a decrease in stability associated with fear of falling, few researchers have sought to quantify the mechanical and neuromuscular effects

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induced by environmental contexts that alter the potential consequences of a fall. Furthermore, the existing knowledge on the effects of postural threat is largely limited to responses of younger adults that cannot be generalized to the elderly. With the growing number of older adults and the associated disparities caused by falling, fall prevention should be directed at physical, as well as psychological, indicators of fall-risk.

E. Objectives of the Thesis

The purpose of this research is to investigate the mechanical and neuromuscular consequences of increased postural threat in healthy, younger and older adults. Two studies were conducted to examine this objective. The research questions were designed to examine the effects of balance during static and anticipatory postural control. Study 1: What are the mechanical and neuromuscular consequences of postural threat on the maintenance of upright stance among healthy younger and older adults? Study 2: What are the mechanical and neuromuscular consequences of postural threat on the regulation of postural control during preparatory and focal movement phases of a voluntary forward reach among healthy younger and older adults?

1. Environmental Manipulation of Postural Threat

In these experiments, postural threat was imposed by a manipulation of height and position on an elevating platform. The conditions of postural threat involved low or high elevations and unconstrained or constrained stepping conditions on the platform (LUC, LC, HUC, and HC) (see Fig. 2.1). The height variable increased the consequences of a loss of balance as a fall from 1.2m would have more severe implications than a fall from 0.17m. The position variable constrained the selection of postural strategy as a step could not occur in the edge conditions. The lowest threat condition, LUC, constitutes low elevation and low constraint while the most threatening condition, HC, involves high elevation and high constraint.

2. Study 1 Predictions

We hypothesised that both younger and older adults will demonstrate more conservative postural accommodations in balance-threatening conditions. We proposed that this conservative posture, in accordance with Winter's (1998) stiffness control model, would be evidenced in all three levels of biomechanical measurement (i.e. kinematic, kinetic, and neuromuscular) during threatening conditions. Therefore, we expect a more posterior body position, reduced amplitude and variability of postural sway, and increased coactivation of agonist postural muscles (i.e. joint stiffness). We further predicted that older adults would be more affected by increased postural threat such that greater conservative accommodations would occur in older compared to younger adults.

3. Study 2 Predictions

We expected younger and older adults would decrease their overall range of reach under the most threatening conditions. We predict this functional change will be the result of conservative changes to reach strategy. We propose that participants will restrict horizontal displacement of the hip and reduced hip flexion in tenuous conditions to achieve more posterior body positions and limited movement of the COM both prior to and during a forward reach. We also hypothesise that during increased postural threat, there will be a larger latency between the onset of postural and the onset of focal muscles to ensure adequate preparation occurs prior to the upcoming disturbance of the intended movement. We expect this delay will likely be due to the CNS making larger adjustments and allowing more time for preparation to ensure a loss of balance will not occur. Again, we proposed that older adults would demonstrate larger modifications to reaching under postural threat than younger adults.

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II.   STUDY 1:   AGE-RELATED EFFECTS OF POSTURAL THREAT ON THE REGULATION OF UPRIGHT STANCE:

A. Introduction

Falls are a leading source of mortality and morbidity in the elderly, occurring in approximately one-third of adults over 65 years of age (Suzuki et al., 1997; Tinetti & Williams, 1997; Province et al., 1995; Perry, 1982). Older adults who have fallen may suffer moderate to severe injuries including fractures, soft tissue damage, and head or spinal cord injury that have long-term physical disabilities and possibly even death (Wilkins, 1999b; Nakamura et al., 1998; Sattin, 1992). Although only 15 percent of all falls require medical assistance (Vellas, Cayla, Bocquet, dePemille, & Albarede, 1987), falls are a contributing factor in 40 percent of all nursing home admissions (Kellogg International Work Group on the Prevention of Falls by the Elderly, 1987).

In addition to the physical injuries associated with a fall, psychological consequences also prevail. For example, many older adults suffer from an incapacitating anxiety regarding their balance abilities (Maki et al., 1991). Fear of falling, or a diminished confidence regarding balance abilities (Tinetti et al., 1990), is a pervasive and debilitating condition that may result in self-imposed activity restriction, further physical deterioration, and ultimately, a complete loss of independence (Yardley & Smith, 2002; Murphy, Williams, & Gill, 2002; Howland et al., 1998; Tinetti et al., 1994; Atfken et al., 1994; Howland et al., 1993; Tinetti et al., 1988). Fear of falling is such a widespread phenomenon that it affects an estimated 50 to 60 percent of elderly fallers (Yardley, 1998; Tinetti et al., 1990; Downton & Andrews, 1990; Tinetti et al., 1988). However, although often equated with the 'post-fall syndrome' (Murphy & Isaacs, 1982), fear of falling is present in nonfallers as well (Yardley, 1998; Chandler et al., 1996; Tinetti et al., 1994; Tinetti et al., 1988). In fact, 30 to 50 percent of community-

dwelling older adults who have not fallen do express a fear of falling (Downton & Andrews, 1990; Tinetti et al., 1988).

It is not unwarranted to expect that fear of falling would influence postural control. In fact, previous research has confirmed that an association does exist between fear of falling and postural instability among community-dwelling and institutionalized elderly cohorts (McAuley et al., 1997; Myers et al., 1996; Baloh et al., 1995; Tinetti et al., 1994; Franzoni et al., 1994; Baloh et al., 1994; Maki et al., 1991; Tinetti et al., 1988). However, because of the cross-sectional nature of the majority of work to date, these studies cannot present a concise cause-effect relationship between fear of falling and postural control. Consequently, investigators have been unable to conclude whether fear of falling causes balance impairments or balance impairments cause fear of falling.

To further elucidate the relationship between fear of falling and postural control, recent research efforts have examined whether the central nervous system (CNS) alters the regulation of balance and locomotion under environmental manipulations that alter the potential consequences of a fall (Adkin et al., 2002; Brown et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997). Using this paradigm, individuals stand at the edge of an elevated platform (Adkin et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997) or walk along an elevated narrow walkway (Brown et al., 2002). The impending threat associated with the manipulation in environmental context has been shown to heighten physiological arousal (Adkin et al., 2002; Brown et al., 2002); the inference is that the environmental manipulation imposes situation-specific anxiety regarding balance ability (Critchley, 2002; Ashcroft et al., 1991), such as that which may occur when there is a fear of falling.

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A number of studies have demonstrated that the CNS imposes conservative modifications to postural control under threatening conditions, and that the accommodations that emerge under these contexts would serve well to reduce the probability of a fall occurrence (Adkin et al., 2002; Brown et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997). For example, participants adopt tighter control over upright standing as postural threat increases. This adaptation is evidenced by reduced variability and increased frequency of postural sway in the condition of greatest postural threat (Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999). Reduced variability and increased frequency of sway are congruent with a strategy of stiffness control at the ankle joint, a finding that was recently confirmed by Carpenter and colleagues (2001). Winter (2001; 1998) proposed that ankle joint stiffening is accomplished through the cocontraction of agonist/antagonist muscle pairs. The resulting effect is a tighter regulation of COM control as evidenced by a decreased time lag between COM and COP movements. This accommodation serves to reduce the permitted range of COM displacement and, thus, may minimize the probability of a loss of balance following a disturbance (Houk & Rymer, 1981).

To date, we know that when postural threat increases, the CNS regulates postural control by increasing ankle stiffness (Carpenter et al., 2001). This finding, however, is based on tests conducted on young adults (Adkin et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997). Postural instability is a problem inherent to the elderly and not younger cohorts. This fact presents the possibility that age-dependent differences may exist for the effects of postural threat on the regulation of upright standing. Thus, the purpose of this study was to investigate the effects of postural

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threat on the mechanical and neural regulation of upright standing among younger and older adults.

We hypothesised that during conditions of greatest postural threat, younger and older adults would adopt more conservative postural strategies that minimise the possibility of a loss of balance. Similar to previous research, we expected this conservative posture to correspond with Winter's stiffness control model (Winter et al., 2001; Winter et al., 1998) as denoted by a reduction in whole body (COM) movement, an increase in the frequency and a decrease in amplitude and variability of postural sway, and an increase in coactivation of antagonist postural muscles. Finally, based on previous work in our laboratory (Brown et al., 2002), we predicted that older adults would be differentially influenced by postural threat. More conservative postural accommodations to increased threat, such as increased backward leaning, decreased amplitude and variability and increased frequency of sway, and increased agonist/antagonist muscle cocontraction were expected to be observed in older adults compared to younger adults.

B. Methods

1. Participants

Fifteen younger (YA; 8 females and 7 males; mean age 22.00 ± 2.17 years) and fifteen older adults (OA; 10 females and 5 males; mean age 69.98 ± 5.35 years) voluntarily participated in this study. All adults were free from neurological and orthopaedic disorders that may affect postural control. A neurologist conducted extensive medical examinations on all older adults to confirm eligibility. The neurological screen was comprised of a standard series of sensorimotor tests of function, an electronystagmogram to exclude vestibular pathologies, and a Mini-Mental State Evaluation to confirm cognitive status. All subjects were informed of the testing protocol prior to signing a consent form in accordance with guidelines from the Human Research Ethics Committee at the University of Lethbridge (Appendix 1).

Prior to testing, all participants completed a Falls History questionnaire and a Self Perceptions of Balance (SPB) questionnaire. The Falls History questionnaire was composed of three questions: 1) fear of heights (yes or no), 2) fear of falling (ranked on a Likert scale from 1 [not afraid] to 10 [very afraid]), and 3) time of most recent fall (in months) (Appendix 2). The purpose of the Falls History questionnaire was to determine the frequency of falling and fear of falling in the testing population. The SPB questionnaire was composed of two items: 1) the Gait Efficiency Scale (GES; (McAuley et al., 1997)) (Appendix 3) and 2) the Activities Specific Balance Scale (ABC; (Powell & Myers, 1995)) (Appendix 3). The purpose of the SPB questionnaire was to quantify perceived confidence on balance and performance of activities of daily living. Both of these items are validated scales that demonstrated

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excellent internal consistency (GES: $\alpha=0.99$ and ABC: $\alpha=0.95$) (McAuley et al., 1997; Powell & Myers, 1995).

2. Postural Threat

A hydraulic lift platform (1.2m x 1.8m, Pentalift, Guelph, ON) was used to manipulate the environmental context and alter the level of postural threat. Participants were tested at two platform heights, low (0.43m) and high (1.4m), and at two locations on the platform, differentiated by the imposed constraint to forward stepping. These locations were in the middle of the platform (0.91m from the edge) and at the edge of the platform indicating a stepping constraint. Thus, four conditions of postural threat were used: 1) low height in the middle of the platform, stepping unconstrained (LUC), 2) low height at the edge of the platform, stepping constrained (LC), 3) high height in the middle of the platform, stepping unconstrained (HUC), and 4) high height at the edge of the platform, stepping constrained (HC). The LUC condition represents the condition of least postural threat while the HC condition represents the condition of greatest postural threat. Figure 2.1 is a schematic illustration of the four postural threat conditions.

The presentation order of the postural threat conditions was block-randomized using a Latin-square design to minimise potential order effects. Four possible combinations of threat conditions were randomly assigned to participants: 1) LUC, LC, HUC, HC, 2) LC, LUC, HUC, HC, 3) HUC, HC, LUC, LC, and 4) HC, HUC, LC, LUC. All combinations were performed by 4 YA and 4 OA except combinations 3 and 4, which were completed by 3 OA and 3 YA respectively.
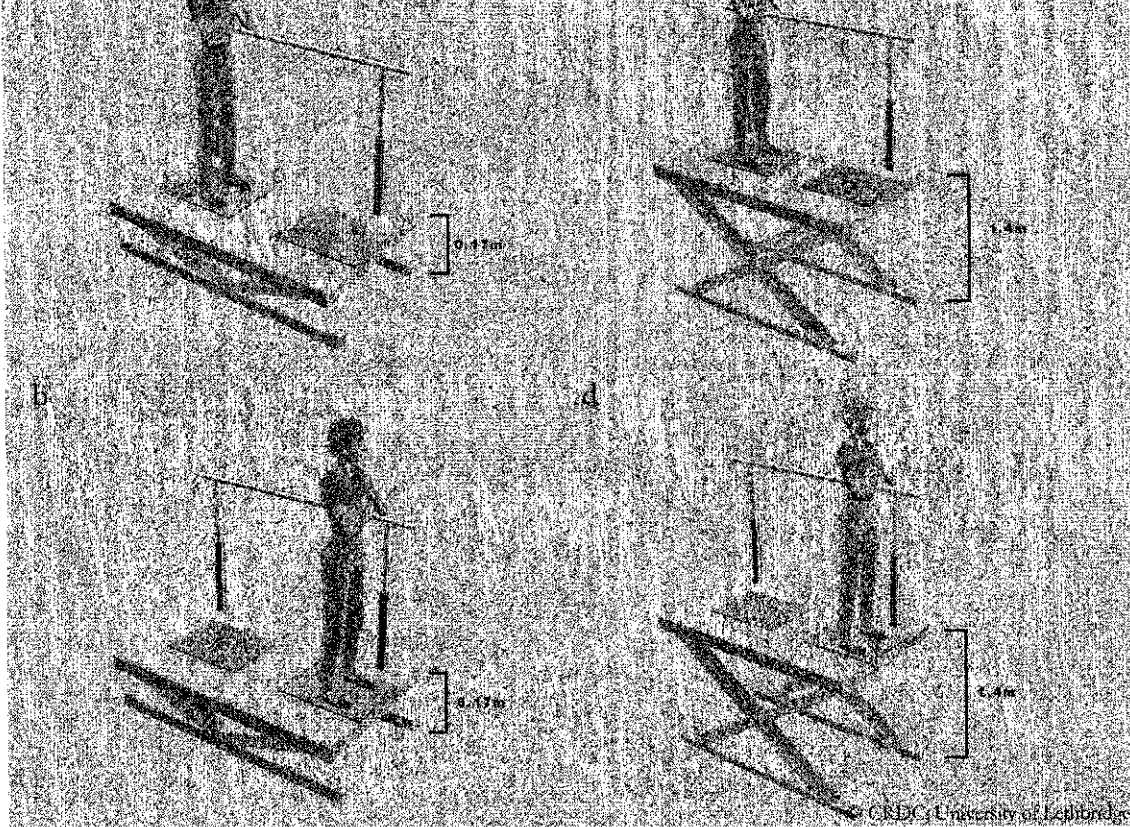
Figure 2.1: The experimental conditions for manipulating postural threat: a) Low-Unconstrained (LUC), b) Low-Constrained (LC), c) High-Unconstrained (HUC), d) High-Constrained (HC).   Note: Participants wore a safety harness and stood at the anterior edge of the platform.

3.   Protocol

Participants were asked to stand as still as possible for three 15s quiet standing trials in each of the four conditions of postural threat.   Participants stood with their arms crossed in front of their chest and their feet positioned flush with the front edge of the force plate and spaced at a comfortable distance apart.   Foot tracings were made to ensure foot placement remained constant throughout all four postural threat conditions.   To ensure participant safety, an overhead harness was worn throughout testing, and a spotter stood behind the participant at all times.

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4. Instrumentation and Data Conditioning

Galvanic skin conductance (GSC) was measured by finger cuffs, containing silver/silver-chloride electrodes from a BioDerm Skin Conductance Level Meter (UFI, Morro Bay, CA), attached to the middle phalanges of digits 3 and 4. GSC, a measure of sweat secretion, is an indicator of a physiological change in arousal (Critchley, 2002; Critchley et al., 2000). GSC data were recorded at a sampling frequency of 600 Hz for 15s.

Spherical reflective markers were attached bilaterally to 12 landmarks on the body including the head of the 5th metatarsal, the lateral malleolus of the fibula, the head of the fibula, the greater trochanter of the femur, the lateral condyle of the humerus, and the greater tubercle of the humerus (Fig. 2.2). Kinematic data were collected at a sampling frequency of 120 Hz using a six-camera motion analysis system (Peak Performance Technologies and Peak Motus 2000 software, Englewood, CO). Raw marker coordinate data were low-pass filtered with a dual pass 4th order Butterworth filter at a cut-off frequency of 3 Hz using custom-written algorithms (MatLab, The MathWorks, Natick, MA, US). Whole body centre of mass (COM) was calculated using a 4-segment model and referenced to the ankle joint. The calculation for COM is as follows:

$$COM = \frac{2m_1 + 2m_2 + 2m_3 + m_4}{m_1 + m_2 + m_3 + m_4}$$

where 1=foot, 2=shank, 3=thigh, and 4=head/arms/trunk (HAT).

16 Reflective Markers (black circles):

1. Medial Condyle of R. Ulna
2. Lateral Condyle of R. Humerus
3. Greater Tubercle of R. Humerus
4. Greater Tubercle of L. Humerus
5. Lateral Condyle of L. Humerus
6. Head of the R. 5th Metatarsal
7. Lateral Malleolus of R. Fibula
8. R. Heel (Posterior View Only)
9. R. Fibular Head
10. Greater Trochanter of R. Femur
11. Greater Trochanter of L. Femur
12. L. Fibular Head
13. L. Heel (Posterior View Only)
14. Lateral Malleolus of L. Fibula
15. Head of the L. 5th Metatarsal
16. Sacrum (Posterior View Only)

7 Surface EMG Electrodes (grey squares):

A. R. Tibialis Anterior m.
B. R Soleus m.
C. R. Rectus Femoris m.
D. R. Biceps Femoris m.
E. R. Rectus Abdominis m.
F. R. Erector Spinae m.
G. R Anterior Deltoid m.
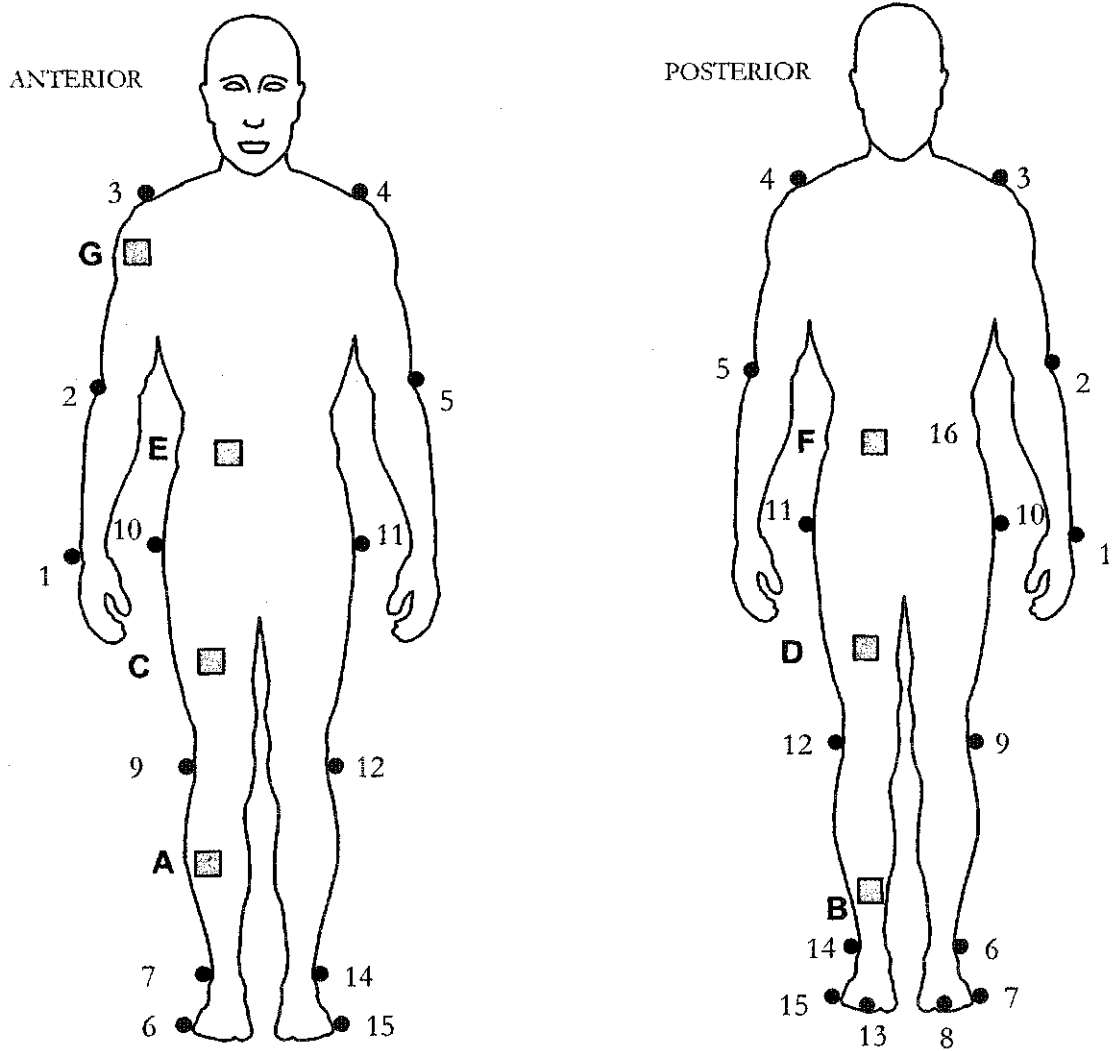
Figure 2.2: This schematic illustrates the placement of 16 reflective markers (black circles) and 12 EMG surface electrodes (grey squares).

Ground Reaction Force (GRF; i.e. F_z) and Moment of Force (Mf) data were collected at a sampling frequency of 600 Hz (Peak Performance Technologies, Englewood, CO) from two Bertec force plates (30x40x8cm) and amplified using a Bertec 6100 amplifier at a gain of 5 (Bertec Corporation, Columbus, OH). GRF and Mf were low-pass filtered with a dual pass 4th order Butterworth filter at a cut-off frequency of 5 Hz. Filtered data were then scaled to N and Nm values using factory set calibration values. Centre of pressure (COP) was computed from the following calculation:

$$COP_x = -M_y/F_z$$
$$COP_y = M_x/F_z$$

where x = anterior/posterior, y = medial/lateral.

An Octopus cable telemetry system (Bortec Electronics Inc., Calgary, AB) was used to collect six channels of electromyograhic (EMG) data. Surface electrodes were attached to the Tibialis Anterior m. (TA), the Soleus m. (SOL), the Rectus Femoris m. (RF), the Biceps Femoris m. (BF), the Rectus Abdominis m. (ABS), and the Erector Spinae m. (ES) on the right side of the body (Fig. 2.2). The EMG data were amplified using a Bortec amplifier (Bortec Electronics Inc., Calgary, AB) at a gain of 1000. All raw EMG data were full-wave rectified and low-pass filtered with a dual pass 4th order Butterworth filter at a cut-off frequency of 100 Hz. Prior to testing, participants sat quietly for 5s for baseline EMG collections. These data were used to normalize all muscle activity data to a ratio of resting EMG values.

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5. Measures of Interest

All three Falls History items were compiled to investigate whether YA and OA differed in fear of falling, time since last fall, and cause of fall. Average scores for the GES and the ABC items were calculated for YA and OA to assess any possible differences in their perceptions of balance and the ability to perform activities of daily living.

Mean GSC was calculated by averaging the galvanic skin response for the first 5s of each trial. Only the first 5s of data were used to prevent the potential effects of stimulus habituation. An increase in GSC was used to infer an increase in physiological arousal imposed by the environmental manipulation of postural threat.

Three variables were derived from the COM and COP time series data to describe the effect of postural threat on postural control in the anterior/posterior (AP) dimension. For each of the displacement profiles, a Mean, Range, and Standard Deviation were calculated. The mean position (Mean) was calculated by averaging the position of each variable across the 15s trial. The range of displacement (Range) was calculated as the difference between the maximum and the minimum position attained during the 15s trial. The variability of the position (SD) was calculated as the standard deviation of the position across the 15s trial. All variables were expressed as a percentage of base of support (BOS) dimensions relative to the location of the ankle marker to normalize for individual variation in foot size and position. In addition, sway velocity (Velocity) and mean power frequency (MPF) were also calculated from the COP data. The Velocity was calculated as the average COP velocity during the 15s by differentiating the COP displacement signal using the finite differences method. To calculate the MPF, a fast fourier transformation (FFT) was performed on the COP position signal to derive the power spectrum density function. The

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data were demeaned and a Hanning window was applied using DataPac2000 software (RUN Technologies Co., Laguana Hills, CA). The MPF, or the average frequency contained within the power spectrum, was calculated as follows:

$$MPF = \frac{\Sigma f * P(f)}{\Sigma P(f)}$$

where f = frequency and P(f) = power at each frequency.

All variables, except the Mean, were calculated in the medial/lateral (ML) dimension of postural control. The Mean was not calculated because the threat was imposed in the AP dimension only. The Range and SD were normalized to BOS width and expressed relative to the centre of stance width as the point of origin.

The normalized EMG data were integrated over a 10 second time interval between the 1st and the 11th second to obtain an estimate of muscle activity amplitudes (iEMG). Integration began at the 1st second rather than at time zero to allow for filter resonance. Anterior/posterior muscle activity ratios (APmar) were calculated by expressing the anterior muscle activity as a ratio of the posterior muscle activity for the primary muscles around the ankle (TA/SOL), the knee (RF/BF), and the hip (ABS/ES) joints. The selected measures of EMG activity permitted examination of the effect of postural threat on the magnitude of muscle activity and the amount of agonist/antagonist co-activity.

Due to technical difficulties, the GSC data were restricted to 12 younger and 5 older adults and the EMG data to 14 younger and 14 older adults; no data were excluded from the analysis of COM and COP measures.

6. Statistical Analysis

The Falls History, GES, and ABC results were analysed using independent Student's t-tests between younger and older adults. The frequency scores on the fear of heights

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question were analysed using a Chi Square test. Separate univariate mixed 3-way [Height (High/Low) X Constraint (Unconstrained/Constrained) X Age (YA/OA)] Repeated Measures Analyses of Variance (RM ANOVAs) were performed on the mean GSC data and the COP and COM mean position data. Separate mixed 3-way [Height X Constraint X Age] Repeated Measures Multivariate Analyses of Variance (RM MANOVAs) were performed on the remaining COP and COM measures. The COP variables included Range, SD, and MPF; the COM measures included Range and SD. The AP and ML variables were analysed separately. The iEMG and AP muscle activity ratios were analysed using separate mixed 3-way [Height X Constraint X Age] RM ANOVAs.

Univariate mixed 3-way [Height X Constraint X Age] RM ANOVAs were performed on significant effects found in the multivariate analyses. Post-hoc Student's t-tests were used to investigate significant interactions revealed by the univariate RM ANOVAs. The alpha criterion was set to 0.05 for all statistical analyses.

C. Results

The results from our statistical analyses are presented in Table 2.1. Descriptive measures (mean ± standard error) for Height and Constraint main effects are presented in Table 2.2. The effect of age on static postural control during non-threatening environmental contexts is already well documented in the literature (Simoneau et al., 1999; Perrin et al., 1997; Hill & Vandervoort, 1996; Baloh et al., 1995; Colledge et al., 1994; Baloh et al., 1994; Baloh et al., 1994; Hytonen et al., 1993; Patla et al., 1992; Maki et al., 1990) and is not presented in this paper. Our findings describe the effects of postural threat on the regulation of upright standing among younger and older adults. Our results indicated that there were no age-related effects of postural threat on arousal, COM, and COP. However, our findings did indicate that age-dependent effects of do emerge at the neuromuscular level.

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Table 2.1: Summary of statistical findings. Shaded cells contain rmMANOVA results. Open cells contain rmANOVA results. Level of significance is indicated by: *p<0.05, **p<0.01, ***p<0.001.

	H	C	A	AxH	AxC	HxC	AxHxC
GSC							
Mean	***	*					
AP COP	**	*	***			***	
Mean Position	**	***	*			***	
Range							
SD	*					*	
MPF	**		**			***	0.054
ML COP		***	0.060				
Range		***	0.059				
SD		***	*				
MPF		*	***				
AP COM	**	*					
Mean Position	*	**					
Range						*	
SD						*	
ML COM		**	*				*
Range		**	**				
SD		***	**				
iEMG A:P Ratios	**	***		*		*	
Ankle	**	***				*	
Knee	**	***	*	*		*	
Hip		0.058					
iEMG Muscles							
TA	*	**				*	
SOL	**	***					
RF	***	***		***		0.068	
BF							
ABS							
ES			*				

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Table 2.2: Summary of descriptive statistics (mean ± standard error) for height and constraint main effects. Data are collapsed across age groups.

	Low	High	Unconstrained	Constrained
GSC (µS)				
Mean	10.71 ± 1.02	14.40 ± 1.36	11.93 ± 1.16	13.19 ± 1.17
AP COP (% BOS)				
Mean Position	16.48 ± 0.99	15.13 ± 1.16	17.61 ± 0.98	14.01 ± 1.24
Range	6.60 ± 0.52	6.11 ± 0.27	6.50 ± 0.43	6.21 ± 0.32
SD	1.52 ± 0.13	1.33 ± 0.07	1.43 ± 0.11	1.42 ± 0.09
MPF	0.305±0.016	0.355±0.020	0.317 ± 0.017	0.343 ± 0.018
ML COP (% BOS)				
Range	2.32 ± 0.20	2.37 ± 0.18	2.17 ± 0.16	2.52 ± 0.19
SD	0.49 ± 0.04	0.51 ± 0.04	0.45 ± 0.03	0.54 ± 0.04
MPF	0.397±0.016	0.371±0.014	0.399 ± 0.014	0.369 ± 0.015
AP COM (% BOS)				
Mean Position	16.93 ± 0.99	15.15 ± 1.24	17.70 ± 1.14	14.37 ± 1.24
Range	5.65 ± 0.47	5.49 ± 0.31	5.72 ± 0.41	5.43 ± 0.33
SD	1.47 ± 0.13	1.30 ± 0.07	1.40 ± 0.10	1.37 ± 0.09
ML COM (% BOS)				
Range	1.75 ± 0.16	1.92 ± 0.16	1.72 ± 0.14	1.95 ± 0.16
SD	0.43 ± 0.04	0.47 ± 0.04	0.41 ± 0.04	0.49 ± 0.04
iEMG A:P Ratios				
Ankle	0.36 ± 0.07	0.66 ± 0.14	0.28 ± 0.06	0.74 ± 0.15
Knee	0.65 ± 0.06	0.74 ± 0.05	0.61 ± 0.05	0.77 ± 0.06
Hip	0.52 ± 0.04	0.55 ± 0.04	0.56 ± 0.04	0.52 ± 0.03
iEMG (mV)				
TA	12.59 ± 0.75	20.00 ± 3.06	11.55 ± 0.47	21.04 ± 3.27
SOL	64.36 ± 5.64	56.77 ± 5.01	68.69 ± 5.46	52.45 ± 5.50
RF	20.61 ± 1.82	23.35 ± 2.03	20.11 ± 1.77	23.85 ± 2.13
BF	37.50 ± 3.67	34.62 ± 2.51	37.40 ± 3.23	34.72 ± 2.96
ABS	10.92 ± 0.19	10.95 ± 0.22	10.93 ± 0.21	10.94 ± 0.20
ES	25.28 ± 2.09	24.11 ± 1.95	23.81 ± 1.86	25.58 ± 2.18

1. Participant Data

Younger and older adults did not differ on their self-reported ability to perform activities of daily living ($p>0.05$). The results of independent Student's t-tests indicated that there were no significant differences between younger and older adults on the GES and ABC scales of the SPB questionnaire ($t(28)=-1.34$, $p=0.191$ and $t(28)=0.30$, $p=0.764$ respectively). Furthermore, no significant differences were found in perceived fear of falling or fear of heights between younger and older adults ($t(28)=-0.03$, $p=0.764$ and $X^2(1, N=30)=1.29$, $p=0.256$ respectively). A number of participants had to be excluded from the final question on the Falls History Questionnaire, time since last fall, because they did not report ever having fallen; therefore, the analysis was restricted to six younger and seven older adults. Results revealed that younger adults fell more recently than older adults (1.92 months versus 40.29 months; $t(11)=-2.71$, $p=0.020$). However, all falls in younger adults were precipitated by hazardous activities where the risk of falling was greater (e.g. rollerblading).

2. Effects of the Imposed Postural Threat

a) Arousal Data

The imposed postural threat successfully increased levels of arousal in younger and older adults (Fig. 2.3). The results from the 2x2x2 RM ANOVA indicated significant main effects for Height and Constraint ($F(1,15)=21.18$, $p=0.000$ and $F(1,15)=4.84$, $p=0.044$ respectively). All participants demonstrated increases in GSC during high compared to low conditions (14.03µS versus 10.71µS) and constrained compared to unconstrained (13.19µS versus 11.93µS) conditions. Although the interaction between Height and Constraint was not significant ($F(1,15)=2.73$, $p=0.119$), visual inspection of the data revealed that the GSC showed a 26% increase from the LUC to the HC condition ($t(29)=-3.73$, $p=0.002$; Fig. 2.3).

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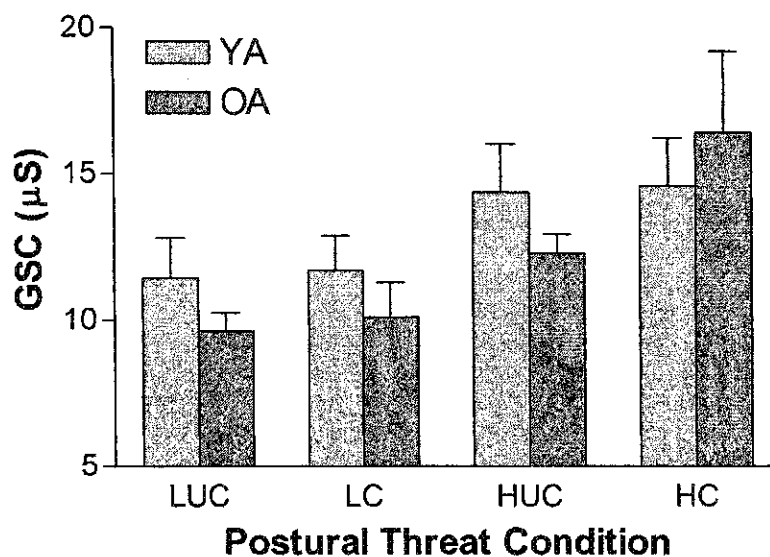
Figure 2.3: Galvanic skin conductance for younger and older adults across four conditions of postural threat. Note that physiological levels increase as threat increases.

b) <u>Centre of Pressure and Centre of Mass Data</u>

(1) Anterior-Posterior Direction

Significant main effects for Height and Constraint emerged in COP Mean Position (Fig. 2.4a; $F(1,28)=4.60$, $p=0.041$ and $F(1,28)=17.22$, $p=0.000$ respectively) and COM Mean Position (Fig. 2.4b; $F(1,28)=6.33$, $p=0.018$ and $F(1,28)=9.57$, $p=0.004$ respectively) for all participants. Specifically, a more posterior position of the COP was observed in the high compared to the low (16.48 versus 15.13 %BOS) and in the constrained compared to the unconstrained conditions (17.61 versus 14.01 %BOS). Likewise, participants adopted a more posterior COM position in the high compared to the low (16.93 versus 15.15 %BOS) and in the constrained compared to the unconstrained conditions (17.70 versus 14.37 %BOS). Participants demonstrated a 28% posterior shift in both COP and COM mean positions from the LUC to the HC conditions.

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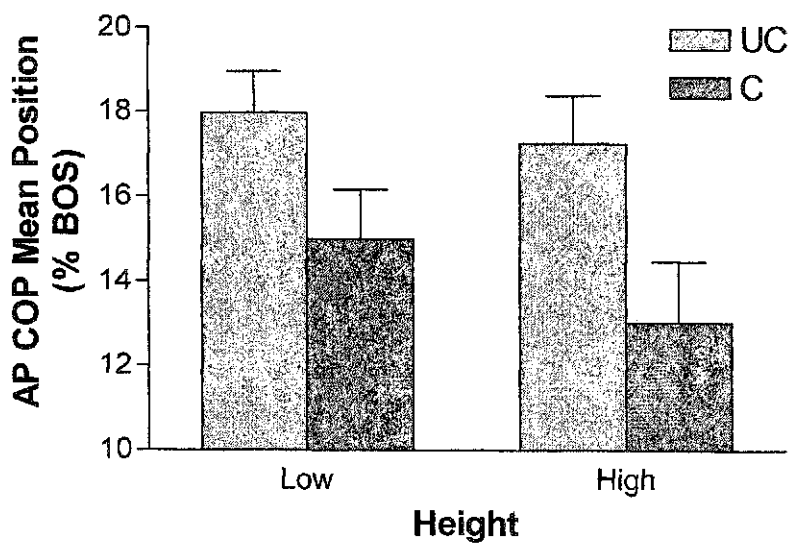
Fig. 2.4.1

Fig. 2.4.2
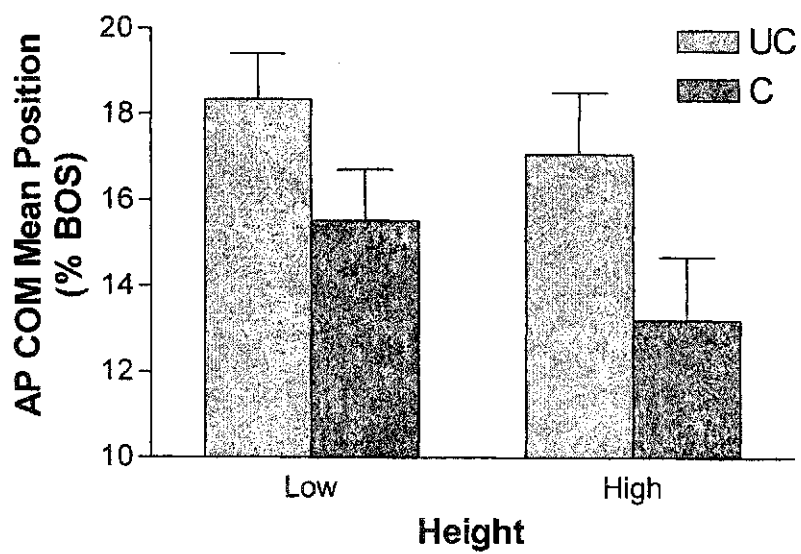
Figure 2.4: Mean position of 1) COP and 2) COM under four conditions of postural threat. Note that mean positions were more posterior in the high versus low conditions and in the constrained and unconstrained conditions.

Significant multivariate interactions between Height and Constraint confirmed that the imposed postural threat significantly influenced COP and COM kinematics for all participants ($F(3,26)=6.82$, $p=0.002$, $\Lambda=0.560$ and $F(2,27)=3.64$, $p=0.040$, $\Lambda=0.787$ respectively). Follow-up univariate tests confirmed that significant Height by Constraint interactions were supported in measures describing the variability and frequency content of the COP (SD: Fig. 2.5.1; $F(1,28)=4.17$, $p=0.050$ and MPF: Fig. 2.5.2; $F(1,28)=21.62$, $p=0.000$) and in range and variability measures of the COM (Range: Fig. 2.6.1; $F(1,28)=6.98$, $p=0.013$ and SD: Fig 2.6.2; $F(1,28)=6.77$, $p=0.015$). The COP measures showed significantly lower variability and increased frequency in the HC compared to the other three conditions (SDCOP$_{LC}$: $t(29)=3.14$, $p=0.004$; MPFCOP$_{LUC}$: $t(29)=2.96$, $p=0.006$, MPFCOP$_{LC}$: $t(29)=4.60$, $p=0.000$, MPFCOP$_{HUC}$: $t(29)=3.67$, $p=0.001$). Likewise, the displacement of the COM showed significantly reduced range and variability in the HC condition compared to the LC and HUC conditions (RangeCOM$_{LC}$: $t(29)=2.46$, $p=0.020$, RangeCOM$_{HUC}$: $t(29)=2.18$, $p=0.038$ and SDCOM$_{LC}$: $t(29)=3.20$, $p=0.003$, SDCOM$_{HUC}$: $t(29)=2.20$, $p=0.036$.

Moreover, a multivariate main effect for Height was found in measures of COP and COM for all participants ($F(3,26)=3.38$, $p=0.033$, $\Lambda=0.719$ and $F(2,27)=7.61$, $p=0.002$, $\Lambda=0.640$ respectively). Also, a significant multivariate main effect for Constraint emerged in COP measures as well ($F(3,26)=2.98$, $p=0.050$, $\Lambda=0.744$). Although significant main effects for Height were supported in the COP MPF ($F(1,28)=9.78$, $p=0.004$), no COM measures indicated a significant main effect for Height. As well, a main effect for Constraint approached significance in COP MPF ($F(1,28)=3.42$, $p=0.075$). Smaller MPF values were observed in the high compared to the low (0.355 versus 0.305 Hz) and in the constrained

compared to the unconstrained conditions (0.343 versus 0.314 Hz).

Fig. 2.5.1
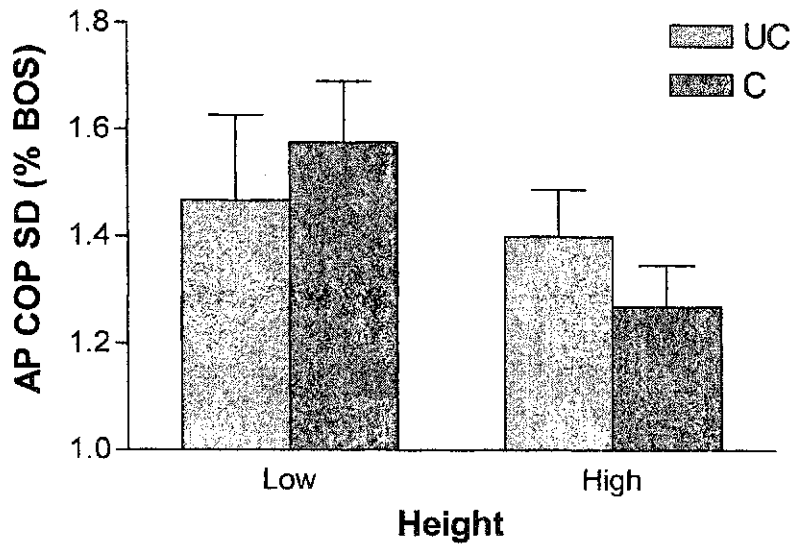
Fig. 2.5.2
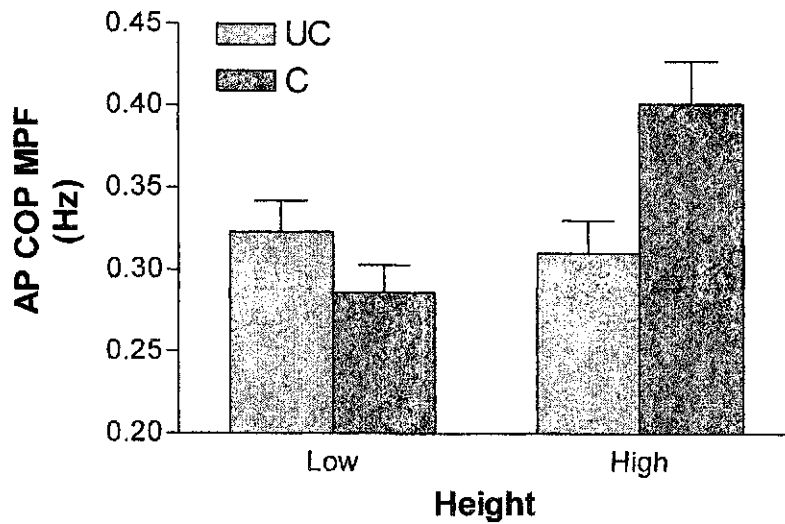
Figure 2.5: COP measures of 1) standard deviation (SD) and 2) mean power frequency (MPF) under four conditions of postural threat. Note that significant HxC interactions revealed decreased SD and increased MPF in the most threatening condition compared to the other conditions of threat.

Fig. 2.6.1
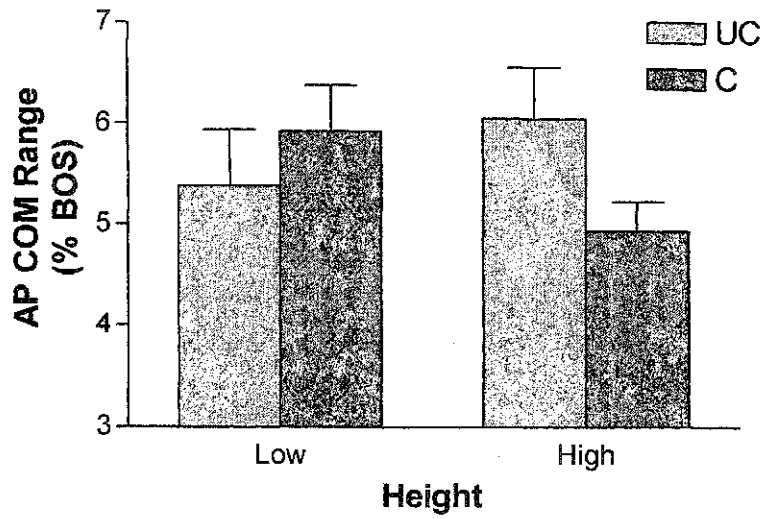
Fig. 2.6.2
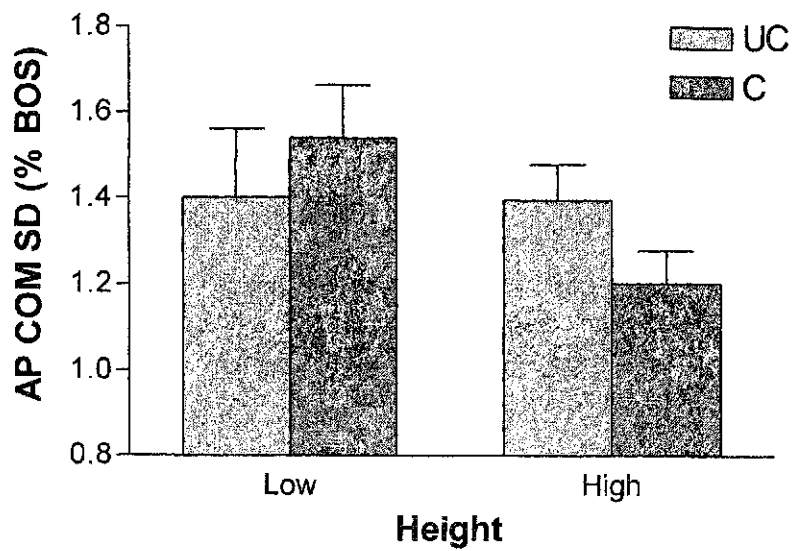
Figure 2.6: COM measures of 1) range and 2) standard deviation (SD) under four conditions of postural threat. Note that significant HxC interactions revealed decreased SD and increased MPF in the most threatening condition compared to the other conditions of threat.

(2) Medial-Lateral Direction

A significant main effect for Constraint emerged in the multivariate analyses for COP $(F(3,26)=6.35,\ p=0.001,\ \Lambda=0.496)$ and COM $(F(2,27)=6.55,\ p=0.005,\ \Lambda=0.673)$ measures. Follow-up comparisons revealed that the COP Range $(F(1,28)=19.75,\ p=0.000)$ and COP SD $(F(1,28)=23.44,\ p=0.000)$ were larger and COP MPF $(F(1,28)=5.73,\ p=0.024)$ was smaller in constrained compared to unconstrained conditions. Greater range (2.17% versus 2.52%) and variability (0.45% versus 0.54%) and reduced frequency (0.40Hz versus 0.37Hz) of COP displacement were observed in constrained compared to unconstrained conditions. Likewise, the COM Range $(F(1,28)=9.84,\ p=0.004)$ and COM SD $(F(1,28)=13.57,\ p=0.001)$ were significantly greater in constrained compared to unconstrained conditions (1.72% versus 1.95% and 0.41% versus 0.49% respectively).

c) Muscle Activity Data

(1) Ankle

Postural threat altered the Anterior/Posterior muscle activity ratios (APmar) of the ankle. There was a significant interaction between Height and Constraint in the APmar of the ankle (Fig. 2.7.1; $F(1,25)=6.13,\ p=0.020$). The highest APmar were observed in the HC compared to the other three conditions of postural threat (APmar$_{LUC}$: $t(26)=4.01,\ p=0.000$, APmar$_{LC}$: $t(26)=3.04,\ p=0.005$, APmar$_{HUC}$: $t(26)=3.74,\ p=0.001$). In fact, Ankle APmar showed a 320% increase from LUC to HC conditions (0.24 versus 1.00). A Height by Constraint interaction was supported in the TA muscle only (Fig. 2.7.2; $F(1,25)=6.13$, $p=0.020$) with a larger amplitude of activity when the postural threat was the greatest, i.e. HC, compared to the other three conditions (TA$_{LUC}$: $t(26)=3.01,\ p=0.006$, TA$_{LC}$: $t(26)=2.65$,

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p=0.014, TA$_{HUC}$: t(26)=2.89, p=0.008; Fig. 2.7.2). In fact, TA muscle activities showed a 149% increase from LUC to HC conditions (11.27mV versus 28.06mV).

Furthermore, significantly larger APmar were observed in the high versus the low (F(1,25)=11.76, p=0.002) and in the constrained versus the unconstrained (F(1,25)=16.17, p=0.000) conditions. Comparison of individual muscles revealed that participants had significantly greater iEMG levels in the TA muscle in the high versus the low conditions (F(1,25)=7.64, p=0.011) and in the constrained versus the unconstrained conditions (F(1,25)=9.79, p=0.004). In contrast, the amplitude of SOL muscle activity was significantly lower in the high compared to the low conditions (F(1,25)=10.24, p=0.004) and in the constrained compared to the unconstrained conditions (F(1,25)=22.12, p=0.000).

Fig. 2.7.1
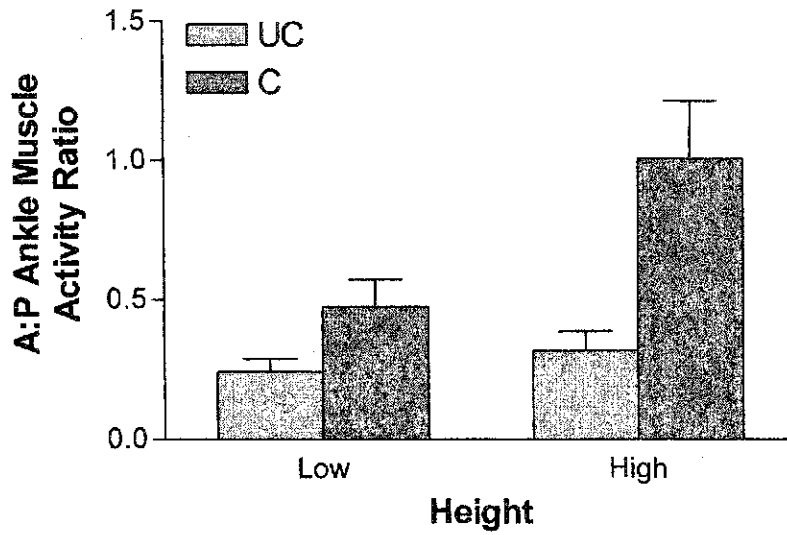
Fig. 2.7.2
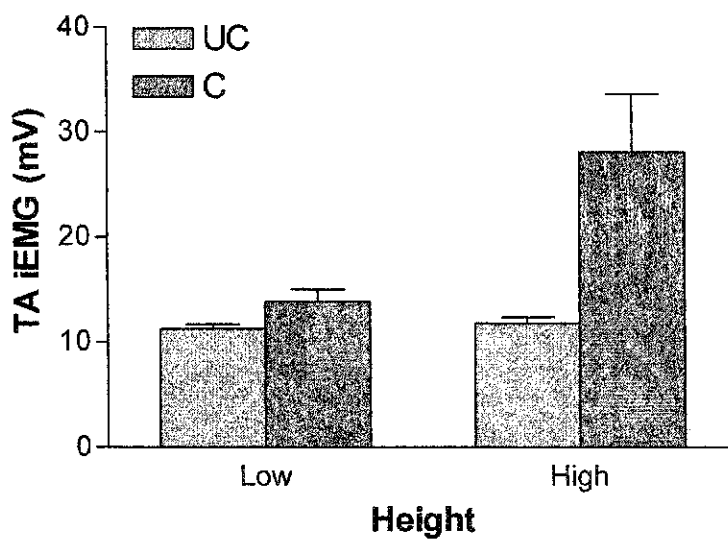
Figure 2.7: These graphs illustrate the 1) anterior/posterior muscle activity ratio (APmar) of the ankle joint and 2) muscle activity of the Tibialis Anterior (TA) under four conditions of postural threat. Note that significant HxC interactions revealed increased APmar of the ankle joint and increased TA iEMG in the HC compared to the LUC, LC, and HUC conditions.

(2) Knee

The Anterior/Posterior muscle activity ratio (APmar) around the knee joint was influenced by postural threat. A significant Height by Constraint interaction (Fig. 2.8; $F(1,25)=4.89$, $p=0.036$) revealed a significantly higher APmar in the HC compared to the other three conditions of postural threat (APmar$_{LUC}$: $t(26)=4.08$, $p=0.000$, APmar$_{LC}$: $t(26)=2.89$, $p=0.008$, APmar$_{HUC}$: $t(26)=3.63$, $p=0.001$). A 41% increase in Knee muscle activity ratios occurred from LUC to HC conditions (0.61 versus 0.86).
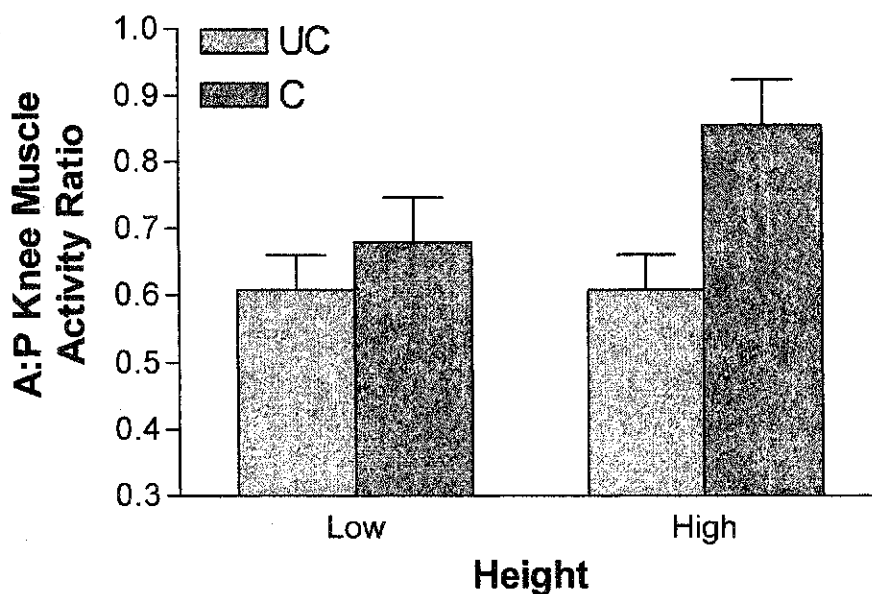
Figure 2.8: This graph illustrates the anterior/posterior muscle activity ratio (APmar) of the knee joint under four conditions of postural threat. Note that a significant HxC interaction revealed increased APmar of the knee joint in the HC compared to the LUC, LC, and HUC conditions.

Moreover, significantly larger APmar were found in the high versus the low ($F(1,25)=9.40$, $p=0.005$) and in the constrained versus the unconstrained ($F(1,25)=14.55$, $p=0.001$) conditions. Follow-up analysis of the independent knee muscles revealed significant main effects for Height and Constraint in the iEMG of the RF muscle.

Participants showed larger RF muscle activity in the high versus the low conditions ($F(1,25)=15.11$, $p=0.001$) and in the constrained versus the unconstrained conditions ($F(1,25)=14.68$, $p=0.001$). In contrast, the amplitude of BF muscle activity remained unchanged between high and low conditions ($F(1,25)=1.87$, $p=0.184$) and constrained and unconstrained conditions ($F(1,25)=2.27$, $p=0.144$).

A significant Height by Age interaction emerged in the APmar of the knee joint (Fig. 2.9.1; $F(1,25)=7.61$, $p=0.011$). Older adults demonstrated significantly larger APmar in the high compared to the low conditions (0.89 versus 0.72; $t(26)=3.89$, $p=0.002$) while the younger adults exhibited similar ratios in both height conditions (0.58 versus 0.57; $t(26)=-0.23$, $p=0.821$). Follow-up analysis of the individual knee muscle revealed that the significant Height by Age interaction occurred in the iEMG muscle activity of the RF muscle only (Fig. 2.9.2; $F(1,25)=13.59$, $p=0.001$). The significant difference in RF muscle activity was found between the different height conditions among older adults only ($t(12)=3.76$, $p=0.003$). In high conditions, older adults had larger magnitudes of RF muscle activity compared to low conditions (28.29mV versus 22.95mV). Younger adults maintained the same amount of RF muscle activity in the low and high conditions (18.28mV versus 18.42mV; $t(13)=0.306$, $p=0.764$).

Fig. 2.9.1
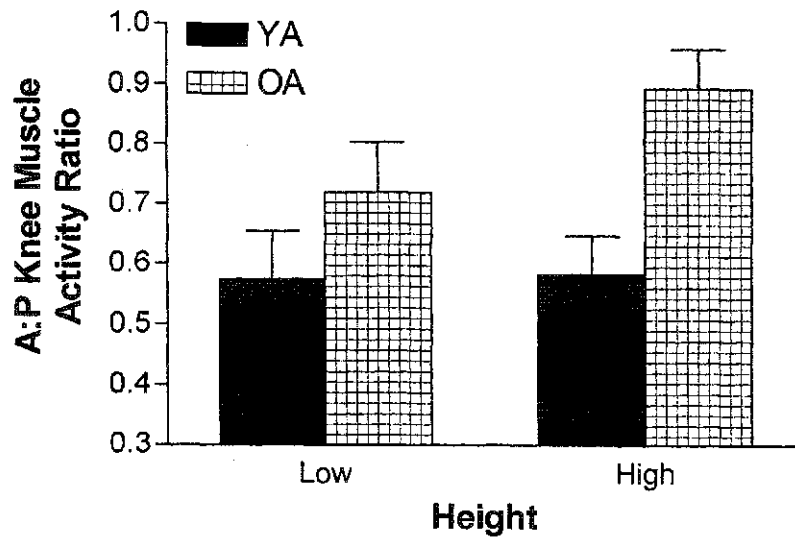
Fig. 2.9.2
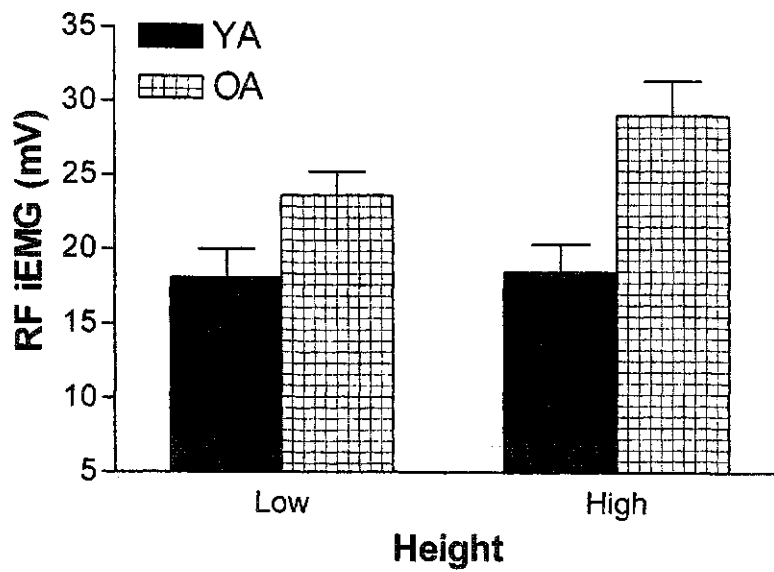
Figure 2.9: This graph illustrates the anterior/posterior muscle activity ratio (APmar) of the knee joint for younger and older adults under high and low conditions of postural threat. Note that older adults had larger APmar of the knee joint and greater RF muscle activity in the high compared to the low conditions while younger adults remained constant across height conditions.

(3) Hip

There were no significant effects or interactions associated with the APmar or the iEMG of the hip muscles.

D. Discussion

The purpose of this study was to identify mechanical and neural modifications to upright standing under conditions of increased postural threat. Because fear of falling is prominent and debilitating among the elderly, and not among younger adults, we were specifically interested in determining whether the postural accommodations were age-dependent. Our results confirmed that the imposed postural threat successfully heightened physiological arousal and altered the regulation of upright stance in healthy younger and older adults. All participants utilized a backward leaning strategy and changed the regulation of posture by reducing variability and increasing frequency of postural sway in the condition of greatest postural threat. Participants also demonstrated greater levels of agonist/antagonist muscle cocontraction due to increased activity in the anterior postural muscles. We have interpreted these findings to mean that participants adopt a more conservative body position, a tighter control of posture, and increased coactivation of the agonist/antagonist muscle pairs of the ankle as threat increases. These adaptations imply that an ankle stiffening strategy emerged for the control of upright standing in response to greater levels of postural threat. No age differences in the kinematics of upright standing emerged for the effect of postural threat. We interpreted these findings to indicate that older adults have maintained the capacity to adapt to their environment. Interestingly, older adults did show different neuromuscular adaptations to postural threat than younger adults. These differences were evidenced by increased amplitude of activity and greater agonist/antagonist cocontraction of lower extremity muscles. This finding presents the possibility that aging may alter the mechanism by which stiffness is achieved.

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1. Mechanical Consequences of Postural Threat

The inverted pendulum model for postural control dictates that the body operates as a single, rigid segment that rotates around the ankle joints (Winter et al., 2001; Winter et al., 1998; Winter, 1995; Winter et al., 1990). To preserve balance, the COM must be maintained within the BOS, prescribed by the dimensions of the feet during quiet stance. The position of the COM is regulated by the COP and movement of the COP directs the movement of the COM (Winter, 1995; Winter et al., 1990). In this model, discrepancy between COP and COM movement is highly correlated with the horizontal acceleration of the COM (Winter, 1995; Winter et al., 1990). As in previous work in this area, our findings are interpreted according to the inverted pendulum model for postural control.

In our experimental paradigm, the greatest postural threat was imposed by positioning participants at the edge of an elevated platform that does not afford a forward step in the event of a loss of balance. It is now known that when the potential consequences of a fall are more severe, the CNS employs a tighter rein of control over posture (Adkin et al., 2002; Brown et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997). In our work, as in the work of others (Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999), tighter control of posture is evidenced by decreased amplitude and variability of sway and increased frequency of sway. The inverted pendulum model of quiet stance dictates that a decrease in the variability of sway, accompanied by an increase in sway frequency, reflects increased stiffness at the ankle joint (Winter et al., 1998). According to Winter and colleagues (2001; 1998), an adjustment in ankle stiffness is a strategy adopted by the CNS to passively control movement of the COM. Thus, during increased postural threat, the CNS responds to COM movements by generating smaller

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COP displacements more frequently in an effort to restrict the COM movement to a smaller area.

Consistent with previous findings (Adkin et al., 2002; Brown et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997), our work showed that the position of the COP and COM were displaced posteriorly in balance-threatening conditions. We propose that a backward shift in COM position creates a larger safety region between the COM position and the anterior edge of the platform. This enlarged safety zone permits a greater range of COM displacement before a potentially injurious forward fall may occur in the event of a perturbation.

Although our findings for the AP control of balance are consistent with prior research, the effect of postural threat on the ML control of balance is much less understood. Previous research on the topic has provided conflicting results. ML balance either showed improvement (Adkin et al., 2000; Carpenter et al., 1999) or did not change (Carpenter et al., 2001) under conditions of postural threat. Interestingly, our work revealed that ML postural control was adversely affected when environmental conditions constrained the forward stepping strategy. It is possible that the AP stepping constraint may increase the attention directed to AP control of balance, consequently leading to a concomitant decline in available attention to ML balance control. This hypothesis presents the possibility that attention directed to AP and ML dimensions of posture are controlled independently under environmental contexts that selectively influence only one dimension of sway. This notion is particularly relevant to older adults who already have a reduced attentional capacity (Lajoie, Teasdale, Bard, & Fleury, 1996; Teasdale et al., 1993) and decreased ML balance abilities that coincide with increased fall-risk (Maki et al., 1994). Further research should investigate the effects of a ML postural threat to the regulation of quiet stance.

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2. Neural Consequences of Postural Threat

Winter (1998) argues that stiffness is achieved by the cocontraction of agonist/antagonist muscle pairs that increase muscle tone in the muscles around the ankle joint. Coactivation serves to provide passive regulation of COM movement without imposing any modification to body orientation. However, to achieve a backward displacement of COM, larger amplitudes of activity in anterior muscle groups may be required to produce effective muscle torque. Thus, in our study, it is impossible to determine whether a change in muscle activity reflects the requirements for a stiffer system, or, as forwarded by Carpenter and colleagues (2001), whether the observed changes in muscle activity are a secondary effect of the threat-induced modification of a backward lean. Indeed, the threat-induced backward leaning strategy may also predispose a change in the characteristic quiet standing muscle activity patterns. However, the tighter control of posture observed in the kinematic data support the use of a stiffness strategy under conditions of increased threat; therefore, we propose that the concomitant increases antagonist muscle coactivity must be, at least in part, a CNS mechanism for achieving a stiffer system. Further research is necessary to elucidate the causal nature of muscle activity and stiffness.

3. Is a stiffness strategy beneficial to postural control?

In our study, we have interpreted a tighter control over posture and agonist/antagonist cocontraction to imply that stiffness increased when postural threat increased. Carpenter et al. (2001) provided support for this inference. The behavioural modifications to increased postural threat provided a more conservative posture in both younger and older adults. In the most threatening conditions, the COM was directed

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posteriorly, and the control of the COM was restricted to a smaller area. We consider these postural accommodations to be conservative because the probability of a forward fall is minimised when the consequences of a loss of balance are most severe. Furthermore, these modifications enable a broader scope of postural compensations to potential balance disturbances, such as backward or lateral stepping, rather than the more frequently adopted forward stepping response should a disturbance occur (Carpenter et al., 1999; McIlroy & Maki, 1993). Although stiffness appears to be beneficial to static postural control, further research is necessary to determine whether the effects of these compensations are also effective during gait and during voluntary and reactive balance tasks. These activities may be particularly demanding for older adults who are already hindered by balance deficits.

4. Why do older adults show different threat-induced modifications to the control of balance than younger adults?

Our findings suggest that older adults require larger amounts of muscle activity to achieve the same conservative control observed in younger adults under conditions of increased postural threat. In particular, older adults demonstrate greater activity of the anterior knee muscles and greater levels of coactivation around the knee joint than younger adults. Older adults may utilise greater levels of RF muscle activity to achieve a backward lean with the same posterior body position as younger adults in the condition of greatest postural threat. The increased cocontraction of knee muscles may reflect a need for more proximal control of posture, in addition to the existing control of posture at the ankle joints, to effectively control COM movement (Gill et al., 2001; Jensen, Brown, & Woollacott, 2001). In older adults, the CNS may increase agonist/antagonist muscle coactivity at both the ankle and the knee to increase stiffness at both joints as a means to accomplish the same tighter regulation of posture observed in younger adults.

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5. Conclusions

Younger and older adults demonstrate mechanical modifications to the regulation of upright stance under conditions of increased postural threat. In response to increased postural threat, physiological arousal is heightened and a conservative posture is adopted. It appears that the purpose of this postural strategy is to minimise the probability of a potentially injurious fall. These findings support the hypothesis of a threat-induced stiffness strategy. The observed coactivation of agonist/antagonist muscle pairs suggests a possible mechanism for ankle stiffness. Interestingly, although all participants showed greater levels of agonist muscle coactivity at the ankle joint, older adults also demonstrated increased cocontraction of agonist/antagonist muscles around the knee. We suggest that the additional stiffening of the knee joint provides a more proximal control of balance to further reduce movement of the COM. These differences in the underlying neural control of posture may enable older adults to achieve the same postural adaptations observed in younger adults. Although a stiffness strategy appears to be a conservative mechanism for postural control, the potential for imbalance during balance recovery following perturbations such as in obstacle avoidance remains unclear.

III. STUDY 2: AGE-RELATED EFFECTS OF POSTURAL THREAT ON POSTURAL CONTROL DURING THE PREPARATORY AND FOCAL MOVEMENT PHASES OF A FORWARD REACHING TASK

A. Introduction

Self-initiated movements associated with everyday activities of daily living (ADLs), such as reaching forward, present a significant threat to balance. Reaching forward is a particularly challenging task because the centre of mass (COM) is also displaced forward and toward the limits of stability. To accommodate for this potentially destabilizing event, the central nervous system (CNS) exacts a highly prescribed sequence of muscle activation, termed a postural synergy, prior to the focal movement (Frank & Earl, 1990; Cordo & Nashner, 1982; Bouisset & Zattara, 1981; Belen'kii et al., 1967). Preparatory distal-to-proximal activation of the posterior leg muscles generates a stabilizing, counter clockwise torque to oppose the upcoming destabilizing, clockwise torque that is generated when the arm is raised for the forward reach (Stapley, Pozzo, Cheron, & Grishin, 1999). The activation of the posterior postural muscles causes a backward displacement of the centre of pressure (COP) for the purpose of controlling the impending forward movement of the COM. These preparatory actions occur prior to the focal movement, and therefore, are termed anticipatory postural adjustments (APAs).

With advancing age, deficits in anticipatory postural control occur. Older adults exhibit spatial and temporal disruption of the postural synergies required for APAs. Man'kovskii and colleagues (1980) found that when fast movements were required, the postural and focal muscles were initiated almost simultaneously in older adults, causing falls in a significant number of trials. However, disruption in the timing of APA is not restricted to fast movements. Older adults exhibit delayed activation of postural muscles during the execution of voluntary movements under slow and self-selected speeds, as well (Inglin &

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Woollacott, 1988; Man'kovskii, Mints, & Lysenyuk, 1980). Inglin and Woollacott (1988) have suggested that the delayed onsets of prime mover muscles were likely a compensatory strategy to permit more time for the CNS to prepare for the intended movement, thereby decreasing the likelihood of a fall.

Despite this proposed compensatory mechanism, reaching is one of the leading causes of injurious falls in the elderly (Nevitt, Cummings, & Hudes, 1991). Not surprisingly, older adults perceive reaching as one of the most challenging activities of daily living to perform without falling (Lachman, Howland, Tennstedt, Jette, & Peterson, 1998; Manning et al., 1997; Powell & Myers, 1995; Tinetti et al., 1990). This difficulty is exacerbated when reaching while atop a step stool or ladder (Powell & Myers, 1995). Possibly, older adults may be more aware of the potentially injurious consequences of a fall and therefore may experience an anxiety or a reduced confidence in their ability to perform the task without falling.

Reduced confidence in mobility tasks, or a fear of falling, afflicts an estimated 60 percent of the elderly, fallers and nonfallers alike (Yardley & Smith, 2002; Legters, 2002) (Myers et al., 1996; Chandler et al., 1996; Tinetti et al., 1994; Maki et al., 1991; Tinetti et al., 1988), and although a fear of falling protects older adults from engaging in dangerous situations, more extreme cases can cause an unnecessary curtailment of activities that may result in further balance impairments and eventually, a loss of independence and quality of life (Cumming et al., 2000; Lachman et al., 1998; Howland et al., 1998; Vellas et al., 1997; Raina, Dukeshire, & Lindsay, 1997; Timiras, 1994; Arfken et al., 1994; Howland et al., 1993; Tinetti et al., 1988).

Because fear of falling has been associated with postural instability, recent research efforts have examined how the CNS may alter the regulation of postural control under

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environmental contexts that modify the potential consequences of a fall (Adkin et al., 2002; Brown et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997). The protocol for these experiments requires that participants stand at the edge of an elevated platform (Adkin et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997) or walk along an elevated narrow walkway (Brown et al., 2002). The assumption is that the environmental manipulation imposes a threat to balance that heightens balance anxiety such as when there is a fear of falling (Adkin et al., 2002; Brown et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997).

A number of studies using the postural threat paradigm have demonstrated that the CNS exerts conservative control over posture and that the adaptations observed minimise the probability of a loss of balance (Adkin et al., 2002; Brown et al., 2002; Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Frank, 1997; Polych & Brown, Study 1). For example, during quiet standing, participants adopted a leaning strategy that directs the centre of mass (COM) away from the potential threat. Furthermore, participants exhibited a tighter control over balance as evidenced by decreased variability and increased frequency of postural sway (Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Brown & Polych, to be submitted). In addition, when quiet stance was perturbed, participants reduced the COM movement and velocity under conditions of greatest postural threat (Brown & Frank, 1997). Moreover, Brown and colleagues (2002) demonstrated that under threatening conditions, participants, particularly older adults, modified gait patterns and demonstrated slower gait speeds and increased time spent in double-limb stance time.

Recent research has extended these findings to investigate the effect of postural threat on anticipatory postural control among younger adults during a rise-to-toes task

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(Adkin et al., 2002). Participants demonstrated reduced COP displacement and velocity in the APA under the most threatening conditions. Furthermore, although the relative timing between postural and focal muscle activity remained constant, the onset of each component was significantly delayed. Interestingly, the number of failed attempts to rise onto toes increased in the condition of greatest postural threat. Adkin and researchers (2002) proposed that unsuccessful trial attempts were most likely due to the decrease in APA rate and magnitude, which is needed to assist in movement initiation. The authors suggested that although these adaptations would serve well to reduce the likelihood of a fall, the successful completion of the task was compromised. This finding implies that postural threat is sufficient to influence the preparatory and focal movement components of a voluntary movement.

These findings of Adkin et al. (2002) provide insight into CNS regulation of anticipatory postural control, as well as the regulation of focal movements under the conditions of postural threat. However, these findings are limited to younger adults and cannot be generalized to the elderly population. Indeed, older adults frequently fall and experience a fear of falling. Furthermore, although the rise-to-toes task provides the opportunity to investigate preparatory and focal movement components, this task is limited in functional utility, especially in older adults. Therefore, our work sought to extend these findings to older adults during a functional forward reach task. Forward reaching was used because it is a common activity of daily living prone to fearful perceptions in older adults (Powell & Myers, 1995). Therefore, the purpose of this study was to examine the effects of postural threat on the mechanical and neuromuscular regulation of postural control during preparatory and focal movement phases of a voluntary forward reach among younger and older adults.

We hypothesised that all participants would exhibit postural adaptations that minimize the probability of a fall but may adversely affect reach performance under conditions of greatest postural threat. In the most threatening conditions, we expected that initially, participants would prepare for the intended movement by adopting more posterior body positions as found in previous quiet stance research. Contrary to Adkin's results (2002), we expected increased magnitude and duration of APA during conditions of greatest postural threat because of task-differences in the purpose of the APA. In the rise-to-toes task, the APA destabilizes the body to initiate the movement. However, although recent research has suggested similar functioning of the APA during reaching (Stapley et al., 1999; Stapley, Pozzo, & Grishin, 1998), most researchers believe the primary function of the APA during arm movements is to provide a stabilizing influence to counter the destabilization of the arm movement (Cordo & Nashner, 1982; Bouisset & Zattara, 1981; Belen'kii et al., 1967).

During the focal movement phase, we predicted a smaller absolute position, range, and velocity of COP and COM movement as postural threat increases. We expect that these adaptations would correspond to restricted horizontal and angular movements of the hip, as well as earlier muscle onset latencies and larger amplitudes of muscle activity.

Finally, we hypothesized that older adults would be more affected by postural threat than younger adults. We predicted older adults would demonstrate more conservative accommodations to postural threat in both preparatory and focal movement phases of the reach. Similar to previous findings by Polych and Brown (Study 1), we expected that older adults would adopt a different postural strategy to achieve a forward reach than younger adults under conditions of greatest postural threat.

B. Methods

1. Participants

Fourteen younger (YA; 7 females and 7 males; mean age 22.14 ± 2.18 years) and eleven older adults (OA; 8 females and 3 males; mean age 69.69 ± 5.16 years) voluntarily participated in this study. All adults were free from neurological and orthopaedic disorders that may affect postural control. A neurologist conducted extensive medical examinations on all older adults to confirm eligibility. The neurological screen was comprised of a standard series of sensorimotor tests of function, an electronystagmogram to exclude vestibular pathologies, and a Mini-Mental State Evaluation to confirm cognitive status. All subjects were informed of the testing protocol prior to signing a consent form in accordance with guidelines from the Human Research Ethics Committee at the University of Lethbridge (Appendix 1).

Prior to testing, all participants completed a Falls History questionnaire and a Self Perceptions of Balance (SPB) questionnaire. The Falls History questionnaire was composed of three questions: 1) fear of heights (yes or no), 2) fear of falling (ranked on a Likert scale from 1 [not afraid] to 10 [very afraid]), and 3) time of most recent fall (in months) (Appendix 2). The purpose of the Falls History questionnaire was to determine the frequency of falling and fear of falling in the testing population. The SPB questionnaire was composed of two items: 1) the Gait Efficiency Scale (GES; (McAuley et al., 1997)) (Appendix 2) and 2) the Activities Specific Balance Scale (ABC; (Powell & Myers, 1995)) (Appendix 3). The purpose of the SPB questionnaire was to quantify perceived confidence on balance and performance of activities of daily living. Both of these items are validated scales that demonstrated excellent internal consistency (GES: $\alpha=0.99$ and ABC: $\alpha=0.95$).

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2. Postural Threat

Similar to Study 1, a hydraulic lift platform (1.2m x 1.8m, Pentalift, Guelph, ON) was used to manipulate the environmental context and alter the level of postural threat. Participants were tested at two platform heights, low (0.43m) and high (1.4m), and at two locations on the platform, differentiated by the imposed constraint to forward stepping. These locations were in the middle of the platform (0.91m from the edge) and at the edge of the platform. Thus, four conditions of postural threat were used: 1) low height in the middle of the platform, stepping unconstrained (LUC), 2) low height at the edge of the platform, stepping constrained (LC), 3) high height in the middle of the platform, stepping unconstrained (HUC), and 4) high height at the edge of the platform, stepping constrained (HC). The LUC condition represents the condition of least postural threat while the HC condition represents the condition of greatest postural threat. (See Figure 2.1 for a schematic illustration of the four postural threat conditions).

The presentation order of the postural threat conditions was block-randomized using a Latin-square design to minimise potential order effects. Four possible combinations of threat conditions were randomly assigned to participants: 1) LUC, LC, HUC, HC, 2) LC, LUC, HUC, HC, 3) HUC, HC, LUC, LC, and 4) HC, HUC, LC, LUC. All combinations were performed by 3 YA and 3 OA except combination 1 which was completed by 4 YA and 3 OA and combination 4 that was completed by 4 YA and 2 OA.

3. Protocol

Participants began each trial in a position of quiet stance with their arms placed at their sides. Shoes were removed, and feet were positioned flush with the front edge of the force plate and spaced at a comfortable distance apart. Foot tracings were made to ensure

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foot placement remained constant throughout all four postural threat conditions. After 2 seconds of quiet standing, participants were instructed to reach as far forward as possible with their right arm at a self-selected speed while maintaining a fixed BOS. The position of maximum reach was maintained for 3 seconds before returning to a position of quiet stance. Three consecutive trials were performed for each condition of postural threat. To ensure participant safety, an overhead harness was worn throughout testing, and a spotter stood behind the participant at all times.

4. Instrumentation and Data Conditioning

Galvanic skin conductance (GSC) was measured by finger cuffs, containing silver/silver-chloride electrodes from a BioDerm Skin Conductance Level Meter (UFI, Morro Bay, CA), attached to the middle phalanges of digits 3 and 4. GSC, a measure of sweat secretion, is an indicator of a physiological change in arousal (Maki & McIlroy, 1996). As per the work of Critchey et al. (2002; 2000) and Ashcroft et al. (1991), we have used a change in physiological arousal to indicate a change in anxiety. GSC data were recorded at a sampling frequency of 600 Hz and converted into units of µS.

Reflective markers were attached bilaterally to 12 landmarks on the body including the head of the 5th metatarsal, the lateral malleolus of the fibula, the head of the fibula, the greater trochanter of the femur, the lateral condyle of the humerus, and the greater tubercle of the humerus (see Figure 2.2). Kinematic data were collected at a sampling frequency of 120 Hz using a six-camera optoelectric motion analysis system (Peak Performance Technologies and Peak Motus 2000 software, Englewood, CO). Raw marker coordinate data were low-pass filtered with a dual pass 4th order Butterworth filter at a cut-off frequency of 3 Hz using custom-written algorithms (MatLab, The MathWorks, Natick, MA, US).

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Whole body centre of mass (COM) in the anterior-posterior (AP) dimension was calculated using a 7-segment model (Figure 3). The calculation for COM is as follows:

$$COM = \frac{m_1{}^*l_1 + m_2{}^*l_2 + m_3{}^*l_3 + m_4{}^*l_4 + m_5{}^*l_5 + m_6{}^*l_6 + m_7{}^*l_7}{m_1 + m_2 + m_3 + m_4 + m_5 + m_6 + m_7}$$

where 1=foot, 2=shank, 3=thigh, and 4=head/arms/trunk (HAT).

Ground Reaction Force (GRF; i.e. F_z) and Moment of Force (Mf) data were collected at a sampling frequency of 600 Hz (Peak Performance Technologies, Englewood, CO) from two Bertec force plates (30x40x8cm) and amplified using a Bertec 6100 amplifier at a gain of 5 (Bertec Corporation, Columbus, OH). GRF and Mf were low-pass filtered with a dual pass 4th order Butterworth filter at a cut-off frequency of 5 Hz. Filtered data were then scaled to N and Nm values using factory set calibration values. Centre of pressure (COP) in the AP plane was computed from the following calculation:

$$COP_x = -M_y / F_z$$

where x = anterior/posterior and y = medial/lateral.

An Octopus cable telemetry system (Bortec Electronics Inc., Calgary, AB) was used to collect six channels of electromyograhic (EMG) data. Surface electrodes were attached to the Tibialis Anterior m. (TA), the Soleus m. (SOL), the Rectus Femoris m. (RF), the Biceps Femoris m. (BF), the Rectus Abdominis m. (ABS), and the Erector Spinae m. (ES) on the right side of the body (see Figure 2.2). The EMG data were amplified using a Bortec amplifier (Bortec Electronics Inc., Calgary, AB) at a gain of 1000. Prior to testing, participants sat quietly for 5s for baseline EMG collections. These data were used to normalize all muscle activity data to resting EMG values. All raw EMG data were full-wave rectified and low-pass filtered with a dual pass 4th order Butterworth filter at a cut-off frequency of 100 Hz.

5. Measures of Interest

a) Participant Data

All three Falls History items were compiled to investigate whether YA and OA differed in their perceptions of falling or their fear of falling. Average scores for the GES and the ABC items were calculated for YA and OA to assess any possible differences in their perceptions of balance and the ability to perform activities of daily living.

b) Arousal

Mean GSC was calculated by averaging the galvanic skin response for the first 5s of each trial. An increase in GSC was used to infer an increase in physiological arousal, and thus an increase in anxiety, imposed by the environmental manipulation of postural threat.

c) Reach Kinematics

The effect of postural threat on the performance of a reach was characterized by the maximum displacement and velocity of the wrist marker achieved during the reach. Maximum reach displacement (ReachDist) was calculated as the peak position of the wrist marker, relative to the participant's arm length, during the reach. The maximum reach velocity (ReachVel) was calculated as the maximum velocity obtained during the reach by differentiating the displacement profile using the finite differences method.

d) Centre of Mass and Centre of Pressure

The COM and COP displacement profiles were analyzed during each phase of the reaching movement: 1) preparatory phase and 2) focal movement phase. These phases were demarcated by an event, peak backward displacement of the COP, that separated the anticipatory postural control component from the voluntary movement component. Prior

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to this event, the CNS was preparing the body for the upcoming movement. After this event, the CNS is actually performing the forward reaching task.
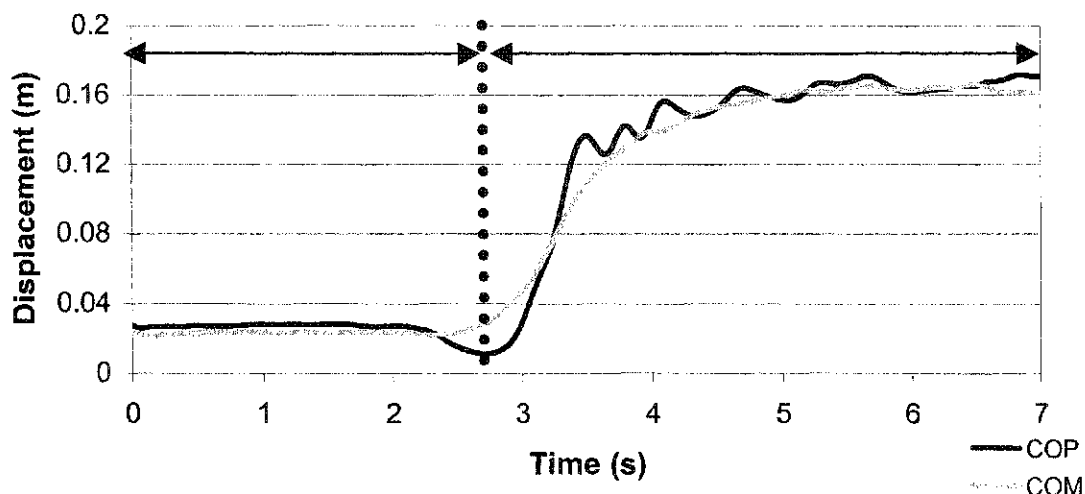
Figure 3.1: Phase determination during a reach trial, demarcated by peak backward position of the COP. The preparatory phase occurs prior to the event and the focal movement phase occurs following the event.

(1) Preparatory Phase

The linear mean start position (Start) was calculated as the average displacement during the beginning 500ms of the trial (prior to reach). Start was expressed as a percentage of base of support (BOS) dimensions relative to the location of the ankle marker to normalize for individual variation in foot size and position.

The preparatory backward movement phase of the COP (COP APA) was characterized by the calculation of range and duration variables. APA Range was calculated as the difference in magnitude between the COP position at the onset of change and the maximum backward COP position prior to the reach. APA Duration was calculated as the time difference between the onset of backward COP movement and the time of peak

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backward COP position. These onsets were determined through visual inspection of the COP displacement profile.

(2) Focal Movement Phase

Three variables were derived from the COM and COP time series data to describe the effect of postural threat on the anterior/posterior (AP) kinematic measures during the focal movement phase. For each of the displacement profiles, peak position (Peak), range (Range), and peak velocity (PeakVel) were calculated. Peak was calculated as the maximum excursion during the reach. Range was calculated as the difference between the Peak and the Start. Peak and Range variables were expressed as a percentage of BOS dimensions. PeakVel was calculated as the maximum velocity obtained during the 15s by differentiating the COP displacement signal using the finite differences method.

e) Joint Kinematics

Joint kinematic data were assessed to observe any potential changes in reach strategy due to the imposed threat. The linear displacement of the shoulder, hip, and ankle and the angular displacement of the hip were selected to describe postural strategy during reaching. The angular hip data were derived from the linear motion analysis data of the hip using custom-written algorithms (MatLab, The MathWorks, Natick, MA, US). These variables were also temporally divided into the preparatory and focal movement components of the forward reach.

(1) Preparatory Phase

The start position (Start) of the shoulder and hip were calculated as the mean position during the beginning 500ms of the trial (prior to reach).

(2) Focal Movement Phase

Peak position (Peak) and range of displacement (Range) variables were calculated for measures of the linear shoulder, linear hip, and angular hip. Peak was calculated as the maximum position obtained during the reach, and Range was calculated as the difference between the Peak and the Start.

f) Electromyography

Muscle onsets were derived from a custom-written computer algorithm (MatLab, The MathWorks, Natick, MA, US) that selected the first point at which the normalized EMG activity exceeded the mean baseline activity plus two standard deviations and remained above this threshold for a minimum of 50ms. This conservative approach has been used by previous researchers (Adkin et al., 2002) to ensure true activation of the muscle, given the nature of surface EMG signals. The mean baseline activity was calculated as the average activity during the first 500ms of the trial (prior to reach). These onsets were verified through visual inspection. The muscle onset latencies were then expressed relative to the time of peak COM velocity to estimate the time required by the CNS to brake the forward movement of the COM. In addition, estimates of muscle activity were obtained through the integration of normalized EMG data (iEMG) over a 1 second time interval beginning at the time of muscle onset. All EMG data were calculated for the focal movement phase only.

Due to technical difficulties, the GSC data were restricted to 11 younger and 4 older adults, the muscle latency data to 11 younger and 8 older adults, and the iEMG data to 13 younger and 9 older adults; all 14 younger and 11 older adults contributed to the kinematic data.

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6. Statistical Analysis

The Falls History, GES, and ABC questionnaires were analysed using independent Student's t-tests except the frequency scores on the fear of heights question which were analysed using a Chi Square test. Mixed 3-way Repeated Measures Multivariate Analyses of Variance (RM MANOVAs) [Height (Low/High) X Constraint (Unconstrained/Constrained) X Age (YA/OA)] were performed on all kinematic measures during the focal movement phase independently. Univariate mixed 3-way Height X Constraint X Age repeated measures analyses of variance (RM ANOVAs) were performed on all remaining measures of interest. Those data that did not conform to a normal distribution were logarithmically transformed.

Univariate mixed 3-way Height X Constraint X Age RM ANOVAs were performed on significant effects found in the multivariate analysis. Post-hoc Student's t-tests were used to investigate significant interactions revealed by the univariate RM ANOVAs. The alpha criterion was set to 0.05 for all statistical analyses.

C. Results

The results from our statistical analyses are presented in Table 3.1. Descriptive measures (mean ± standard error) for Height, Constraint, and Age main effects are presented in Table 3.2. Our findings describe the effects of postural threat on the preparatory and focal movement phases of a forward reach among younger and older adults. Our findings indicate that a postural threat that induced anxiety caused modifications to the preparatory and focal components of a reach in all participants; however, older adults appeared to be more affected by threat than younger adults. In particular, younger and older adults adopted different kinematic reaching strategies in response to postural threat.

1. Participant Data

The results of independent Student's t-tests indicated that there were no significant differences between younger and older adults on the GES and ABC scales of the SPB questionnaire ($t(23)=-0.97$, $p=0.340$ and $t(23)=0.32$, $p=0.752$ respectively). Furthermore, younger and older adults did not differ in measures of perceived fear of falling or fear of heights ($t(23)=0.44$, $p=0.665$ and $X^2(1,\ N=25)=0.65$, $p=0.420$ respectively). The final measure on the Falls History Questionnaire, time since last LOB, was only completed by six younger and six older adults. Analysis of these participants revealed that younger adults fell more recently than older adults (1.92 months versus 37.00 months; $t(10)=-2.35$, $p=0.041$) However, all falls in younger adults could be attributed to participation in hazardous activities where the risk of falling was greater (e.g. hiking or rollerblading).

Table 3.1: Summary of statistical findings during 1) the preparatory phase and 2) the focal movement phase of a forward reach. Shaded cells contain rmMANOVA results. Open cells contain rmANOVA results. Level of significance is indicated by: *p<0.05, **p<0.01, ***p<0.001.

Table 3.1.1

PREPARATORY PHASE							
	H	**C**	**A**	**AxH**	**AxC**	**HxC**	**AxHxC**
GSC							
Mean	***	**		*	*	**	**
COP Start	*			*		***	
COM Start		*		*			
Linear Hip Start	***		*			**	
Linear Shoulder Start	*		**				
COP APA Duration							
COP APA Magnitude							

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Table 3.1.2

	H	C	A	AxH	AxC	HxC	AxHxC
FOCAL MOVEMENT PHASE							
Reach Distance							
Mean	***	***		*	*	***	*
COP	***	***				***	
Peak Position	***	***				***	
Range	***	***				***	
Peak Velocity		0.074				*	
COM	**	***		**	*	***	*
Peak Position	***	***			**	***	
Range	**	***			**	***	
Peak Velocity				*		***	*
Linear Hip	***		***			**	
Peak Position	***		***			***	
Range	***		***				
Linear Shoulder	*		0.067				
Peak Position			0.068				
Range				*			
Angular Hip		*	**			*	
Peak Position		0.066					
Range		**				***	***
Muscle Onset Latencies							
SOL			**			**	0.077
BF	**					*	
ES			0.059			**	
iEMG Muscles							
SOL	***	***				***	
BF							
ES		***					

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Table 3.2: Summary of descriptive statistics (mean ± standard error) for height and constraint main effects. Data are collapsed across age groups.

	Low	High	Unconstrained	Constrained
GSC (µS)				
Mean	12.61 ± 1.17	16.07 ± 1.40	13.24 ± 1.29	15.44 ± 1.24
Reach Distance (m)				
Mean	0.622±0.018	0.573 ± 0.018	0.620 ± 0.018	0.578 ± 0.018
COM (% BOS)				
Start Position	16.59 ± 1.60	15.34 ± 1.57	17.02 ± 1.36	14.91 ± 1.36
Peak Position	60.29 ± 2.49	53.40 ± 2.30	61.92 ± 2.23	51.77 ± 2.29
Range	43.73 ± 2.20	38.03 ± 2.03	44.90 ± 2.04	36.86 ± 2.07
Peak Velocity (m/s)	0.119 ±0.021	0.116 ± 0.014	0.121 ± 0.017	0.114 ± 0.017
COP (% BOS)				
Start Position	17.54 ± 1.60	14.91 ± 1.59	18.32 ± 1.64	14.13 ± 1.43
Peak Position	50.07 ± 2.49	40.95 ± 2.30	50.41 ± 2.23	40.62 ± 2.29
Range	32.54 ± 2.20	26.03 ± 2.03	32.09 ± 2.04	26.48 ± 2.07
Peak Velocity (m/s)	0.234 ±0.021	0.225 ± 0.014	0.238 ± 0.017	0.22 ± 0.017
COP APA (Magnitude in m; Duration in s)				
APA Magnitude	0.017 ±0.001	0.017 ± 0.001	0.017 ± 0.002	0.017 ± 0.001
APA Duration	0.57 ± 0.02	0.62 ± 0.03	0.57 ± 0.022	0.61 ± 0.027
Linear Shoulder (m)				
Start Position	0.31 ± 0.016	0.29 ± 0.013	0.29 ± 0.016	0.30 ± 0.012
Peak Position	0.70 ± 0.019	0.69 ± 0.018	0.70 ± 0.019	0.69 ± 0.017
Range	-0.39 ± 0.011	-0.40 ±0.013	-0.40 ± 0.011	-0.395 ±0.012
Linear Hip (m)				
Start Position	0.30 ± 0.019	0.29 ± 0.017	0.30 ± 0.018	0.30 ± 0.018
Peak Position	0.26 ± 0.022	0.22 ± 0.02	0.25 ± 0.022	0.24 ± 0.02
Range	-0.077±0.007	-0.092 ±0.006	-0.084 ± 0.008	-0.084±0.006
Angular Hip (θ)				
Peak Position	45.24 ±2.803	43.52 ± 3.203	46.08 ± 2.951	42.68 ± 3.05
Range	47.84 ±2.602	47.32 ± 2.841	48.93 ± 2.671	46.24 ± 2.74
Muscle Onset Latencies (s)				
SOL	0.38 ± 0.041	0.42 ± 0.041	0.39 ± 0.042	0.41 ± 0.039
BF	0.36 ± 0.059	0.57 ± 0.074	0.44 ± 0.067	0.49 ± 0.057
ES	0.36 ± 0.059	0.46 ± 0.074	0.40 ± 0.056	0.42 ± 0.069
iEMG (mV)				
SOL	1.10 ± 0.063	1.03 ± 0.056	1.11 ± 0.061	1.02 ± 0.057
BF	0.87 ± 0.041	0.86 ± 0.035	0.88 ± 0.039	0.85 ± 0.037
ES	0.75 ± 0.032	0.76 ± 0.039	0.77 ± 0.033	0.74 ± 0.037

2. Effects of Postural Threat

a) Arousal Data

Environmental manipulation of the consequences of a loss of balance heightened physiological arousal (Fig. 3.2). Follow-up analysis of a significant interaction between Height and Constraint ($F(1,13)=10.53$, $p=0.006$) revealed greater levels of arousal in the most threatening condition compared to the other conditions for all participants (GSC_{LUC}: $t(14)=-3.17$, $p=0.007$, GSC_{LC}: $t(14)=-2.68$, $p=0.018$, GSC_{HUC}: $t(14)=-2.15$, $p=0.050$). Arousal increased by 46% from the LUC to the HC conditions. Furthermore, significant main effects for Height ($F(1,13)=20.68$, $p=0.001$) and Constraint ($F(1,13)=14.85$, $p=0.002$) indicated heightened arousal in the high versus the low (12.61 vs. 16.07μS) and the constrained compared to the unconstrained (13.24 vs. 15.44 μS) conditions.
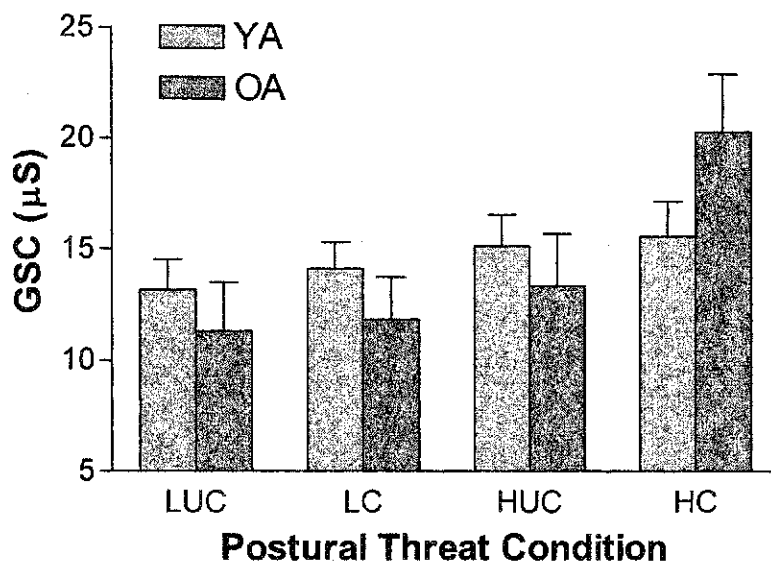
Figure 3.2: Galvanic skin conductance under four conditions of postural threat. Note that a significant HxCxA interaction indicated that levels of physiological arousal increase as postural threat increases, particularly in older adults.

A significant Height X Constraint X Age interaction $F(1,13)=14.34$, $p=0.002$) revealed that postural threat differentially altered arousal in younger and older adults. In the post-hoc analysis, neither younger $(F(1,10)=0.400$, $p=0.541)$ nor older $(F(1,3)=8.31$, $p=0.063)$ adults showed a significant interaction between Height and Constraint. Rather, significant interactions between Height and Age $(F(1,13)=5.53, p=0.035)$ and Constraint and Age $(F(1,13)=7.04, p=0.020)$ emerged and revealed significantly greater arousal levels among younger adults in the high compared to the low $(t(10)=-2.95, p=0.015)$ and among older adults in the constrained compared to the unconstrained $(t(3)=-2.98, p=0.059)$ conditions. Between the LUC and HC conditions, arousal levels increased by 18% in younger and 79% in older adults.

b) Reach Kinematics

Postural threat significantly influenced reaching kinematics (Fig. 3.3). A significant interaction between Height and Constraint was found in reach distance $(F(1,23)=20.64$, $p=0.000)$. Post-hoc analysis revealed significantly shorter reach distances in the HC condition compared to the other three conditions (ReachDist$_{LUC}$: $t(24)=4.69$, $p=0.000$, ReachDist$_{LC}$: $t(24)=4.70$, $p=0.000$, ReachDist$_{HUC}$: $t(24)=4.28$, $p=0.000$). Participants reduced reach distance by 15% from the LUC to the HC condition. Moreover, main effects for Height $(F(1,25)=26.92, p=0.000)$ and Constraint $(F(1,25)=21.47, p=0.000)$ indicated that reach distance decreased significantly in the high compared to the low (0.622 vs. 0.573m) and the constrained compared to the unconstrained (0.620 vs. 0.578m) conditions.
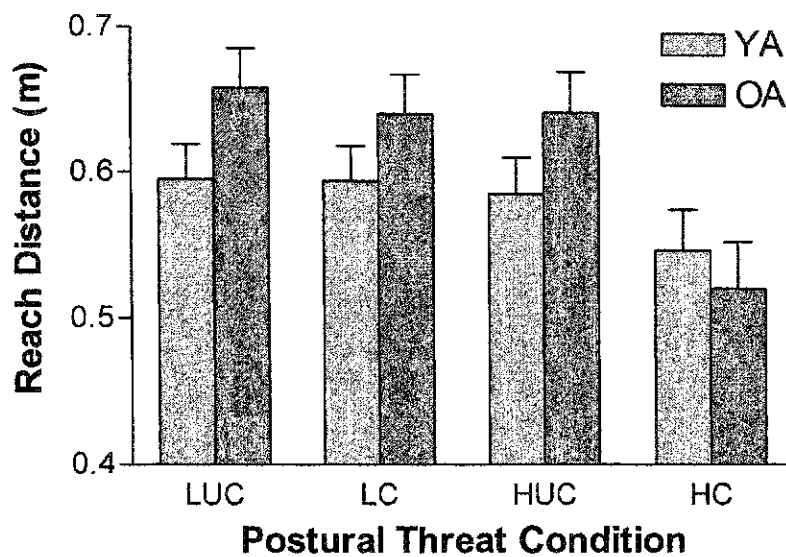
Figure 3.3: Reach distance of younger and older adults under four conditions of postural threat. Note that a significant HxCxA interaction indicated that participants, particularly older adults, did not reach as far in conditions of increased postural threat.

Age-related effects of postural threat emerged in measures of reaching kinematics (Fig. 3.3). A significant 3-way interaction for Height, Constraint, and Age ($F(1,23)=4.32$, $p=0.049$) emerged for measures of reach distance. Follow-up analyses revealed a significant Height X Constraint interaction within older adults only ($F(1,10)=19.46$, $p=0.001$ respectively). Older adults did not reach as far in the HC compared to the LUC, LC, and HUC conditions (ReachDist$_{LUC}$: $t(10)=4.36$, $p=0.001$, ReachDist$_{LC}$: $t(10)=4.22$, $p=0.002$, ReachDist$_{HUC}$: $t(10)=4.44$, $p=0.001$). In fact, older adults demonstrated a 21% decrease in reach distance from the HC compared to the LUC conditions while younger adults decreased reach distance by only 8%. Furthermore, significant Height X Age ($F(1,23)=4.45$, $p=0.046$) and Constraint X Age ($F(1,23)=6.85$, $p=0.015$) interactions emerged. Follow-up t-tests revealed that both younger ($t(13)=3.38$, $p=0.005$) and older ($t(10)=3.76$, $p=0.004$) adults

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did not reach as far in the high versus the low and that older adults reached less far in the constrained versus unconstrained conditions ($t(10)=4.19, p=0.002$).

c) Kinematics of the Preparatory Phase of a Reach

(1) Centre of Pressure and Centre of Mass

Postural threat affected the start position of the COP and the COM (Fig. 3.4). Significant main effects for Constraint emerged in the StartCOP (Fig. 3.4.1; $F(1,23)=12.42$, $p=0.002$) and StartCOM locations (Fig. 3.4.2; $F(1,23)=10.05, p=0.004$) and indicated a more posterior position in the constrained versus the unconstrained conditions for both COP (18.32 vs. 14.13 %BOS) and COM (17.02 vs. 14.91 %BOS) positions. Although a significant interaction between Height and Constraint was not found, participants maintained COP and COM positions that were 35% and 20% more posterior respectively in the HC compared to the LUC conditions.

Significant interactions between Height and Age did emerge for the StartCOP ($F(1,23)=7.13$, $p=0.014$) and StartCOM ($F(1,23)=7.05$, $p=0.014$) positions, however, no significant effects were found in the post-hoc analysis. Older adults did demonstrate posterior shifts in COP and COM start position from low to high conditions (COP: $t(10)=2.07, p=0.066$; COM: $t(10)=2.11$, $p=0.061$) that approached statistical significance while younger adults no change in these measures across height conditions (COP: $t(13)=-1.60, p=0.127$; COM: $t(13)=-1.31, p=0.212$).

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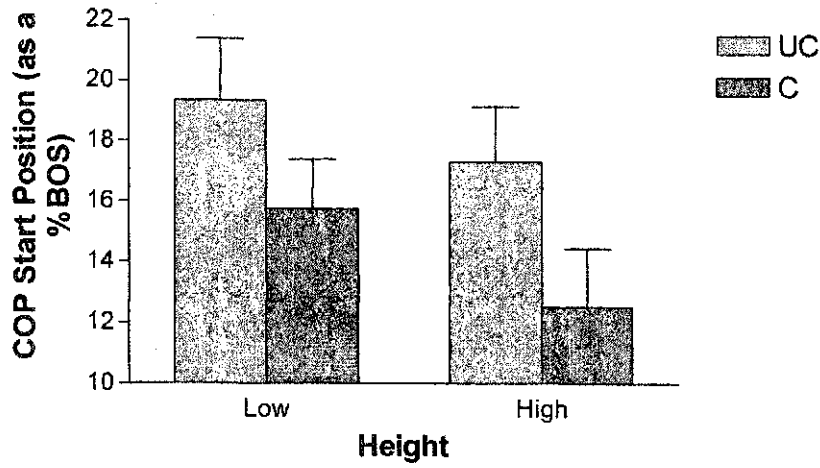
Fig. 3.4.1

Fig. 3.4.2
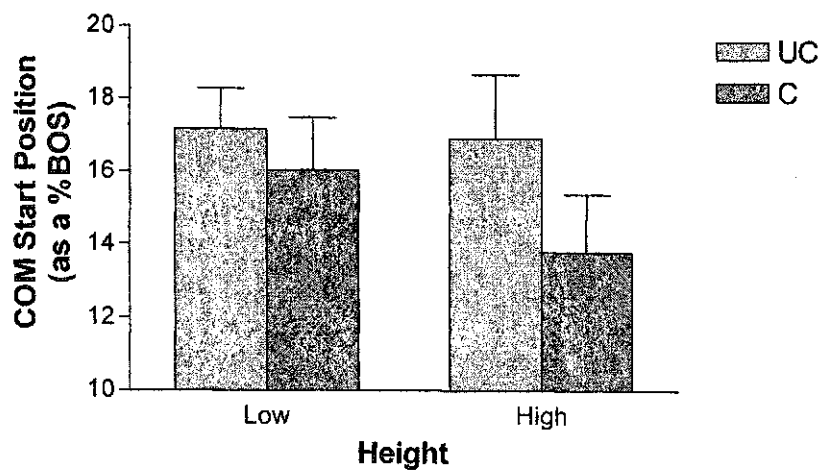
Figure 3.4: Start position of 1) COP and 2) COM as a percentage of base of support under four conditions of postural threat. Note that a significant HxC interaction revealed that the most posterior start positions were observed in the conditions of greatest postural threat.

(2) Linear Joint Kinematics

A significant Height x Constraint interaction was revealed in the starting position of the hip (Fig 3.5; $F(1,23)=8.22$, $p=0.004$); more posterior positions of the hip were found in the HC compared to the LUC, LC, and HUC conditions for all participants (LinearStartHip$_{LUC}$: $t(24)=4.05$, $p=0.000$, LinearStartHip$_{LC}$: $t(24)=6.46$, $p=0.000$, LinearStartHip$_{HUC}$: $t(24)=2.31$, $p=0.030$). In fact, the linear start position of the hip was 6% more posterior in the HC compared to the LUC condition. Significant main effects for Height and Age were also identified in the start position of the hip ($F(1,23)=51.77$, $p=0.000$ and $F(1,23)=7.34$, $p=0.013$ respectively) and the shoulder ($F(1,23)=5.34$, $p=0.030$ and $F(1,23)=8.20$, $p=0.009$ respectively). The initial positions of both joints were more posterior in the high compared to the low conditions and among younger compared to older adults.
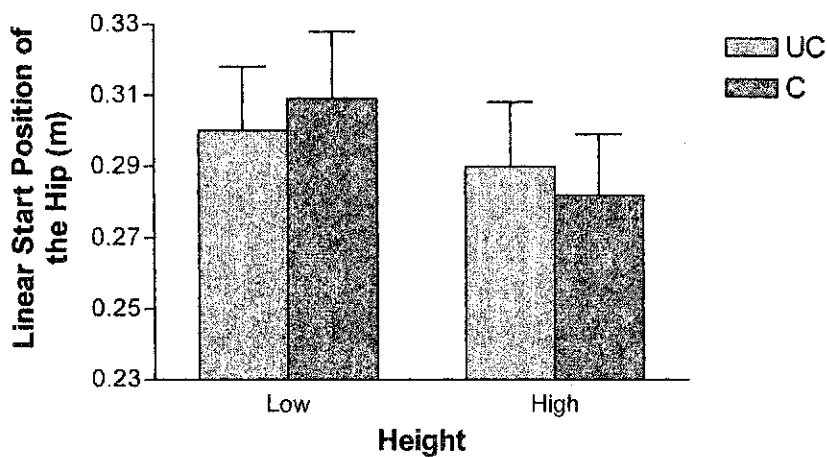
Figure 3.5: Linear start position of the hip joint under four conditions of postural threat. Note Note that a significant HxC interaction revealed that the linear start position of the hip was more posterior in the HC compared to the other conditions of postural threat.

(3) Anticipatory Postural Adjustments of the Centre of Pressure

Postural threat did not influence the magnitude or duration of the backward displacement of the COP. However, visual inspection of the data did suggest a possible trend that indicated APA duration increased as postural threat increased.

d) Kinematics of the Focal Movement Phase of a Reach

(1) Centre of Pressure and Centre of Mass

COP and COM reach kinematics were altered under conditions of postural threat (Fig. 3.6 and 3.7 respectively). Results from the multivariate analysis revealed significant interactions between Height and Constraint for the COP ($F(3,21)=7.22$, $p=0.002$, $\Lambda=0.492$) and COM ($F(3,21)=12.67$, $p=0.000$, $\Lambda=0.356$) variables. Follow-up univariate analysis indicated that the Height x Constraint interaction was supported by all three COP measures (PeakCOP ($F(1,23)=8.07$, $p=0.009$), RangeCOP ($F(1,23)=17.51$, $p=0.000$), and PeakVelCOP ($F(1,23)=6.83$, $p=0.016$) and all three COM measures (Peak COM ($F(1,23)=36.86$, $p=0.000$), RangeCOM ($F(1,23)=14.15$, $p=0.001$), and PeakVelCOM ($F(1,23)=16.02$, $p=0.001$). In the most threatening condition, participants demonstrated restricted peak COP and COM positions (PeakCOP$_{LUC}$: $t(24)=4.82$, $p=0.000$, PeakCOP$_{LC}$: $t(24)=4.26$, $p=0.000$, PeakCOP$_{HUC}$: $t(24)=5.65$, $p=0.000$; PeakCOM$_{LUC}$: $t(24)=5.97$, $p=0.000$, PeakCOM$_{LC}$: $t(24)=5.36$, $p=0.000$, PeakCOM$_{HUC}$: $t(24)=7.15$, $p=0.000$) and smaller ranges of COP and COM

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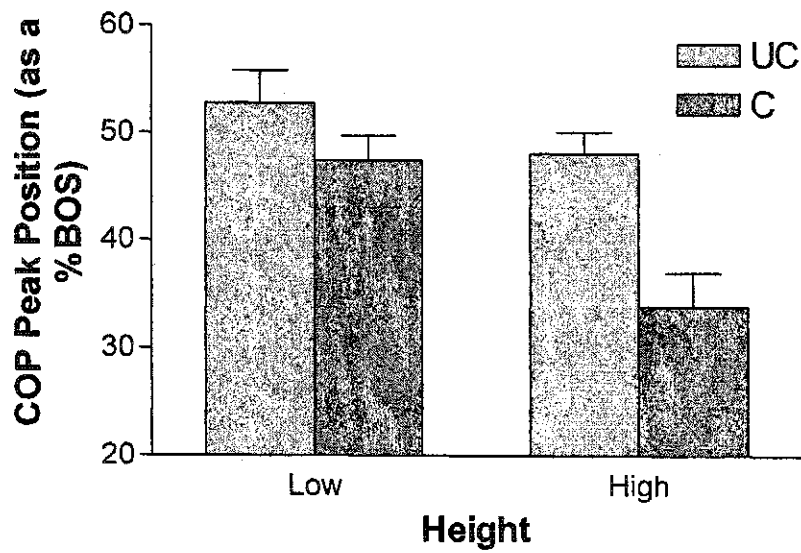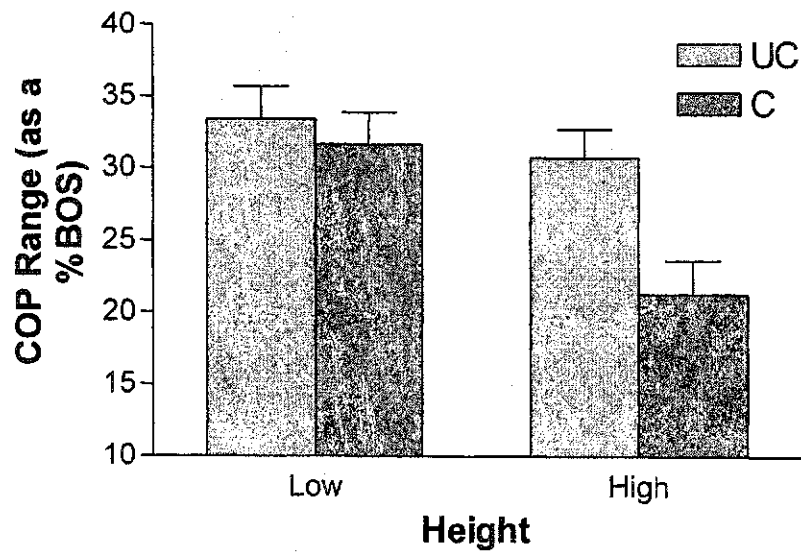
Fig. 3.6.1

Fig. 3.6.2

Fig. 3.6.3
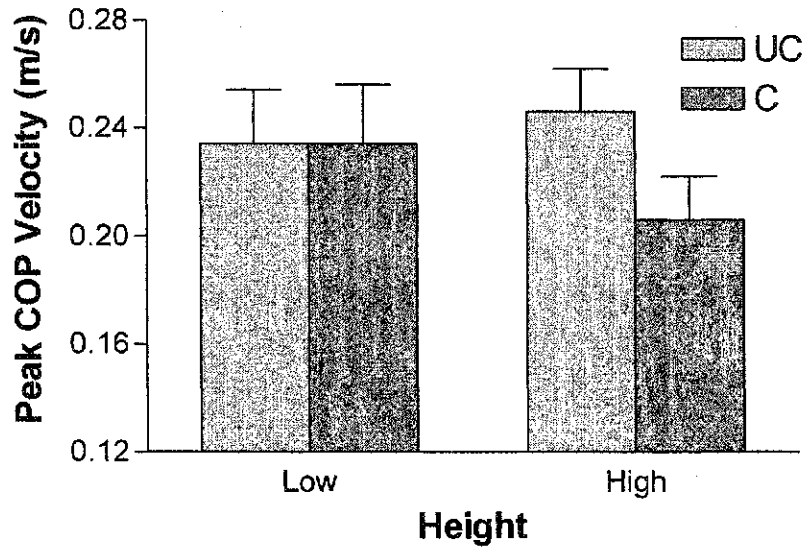
Figure 3.6: These graphs illustrate the COP 1) peak position 2) range, and 3) peak velocity on four conditions of postural threat. Note that significant HxC interactions indicate decreases in COP kinematics in the most threatening conditions.

displacement (RangeCOP$_{LUC}$: $t(24)=6.21$, $p=0.000$, RangeCOP$_{LC}$: $t(24)=5.26$, $p=0.000$, RangeCOP$_{HUC}$: $t(24)=6.16$, $p=0.000$; RangeCOM$_{LUC}$: $t(24)=5.69$, $p=0.000$, RangeCOM$_{LC}$: $t(24)=4.32$, $p=0.000$, RangeCOM$_{HUC}$: $t(24)=5.02$, $p=0.000$) compared to the other three conditions. Slower peak COP and COM velocities were also found in the HC compared to less threatening conditions (PeakVelCOP$_{LUC}$: $t(24)=3.11$, $p=0.005$; PeakVelCOM$_{LC}$: $t(24)=2.51$, $p=0.019$, PeakVelCOM$_{HUC}$: $t(24)=3.35$, $p=0.003$).

Fig. 3.7.1
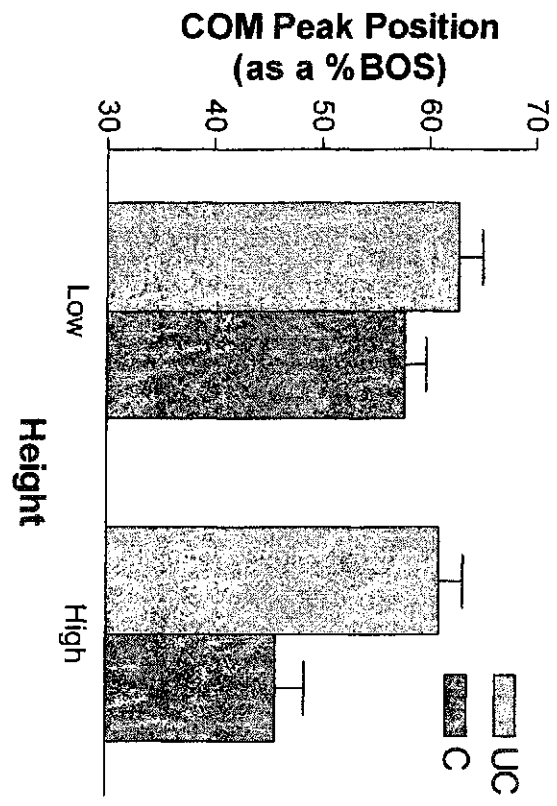
Fig. 3.7.2
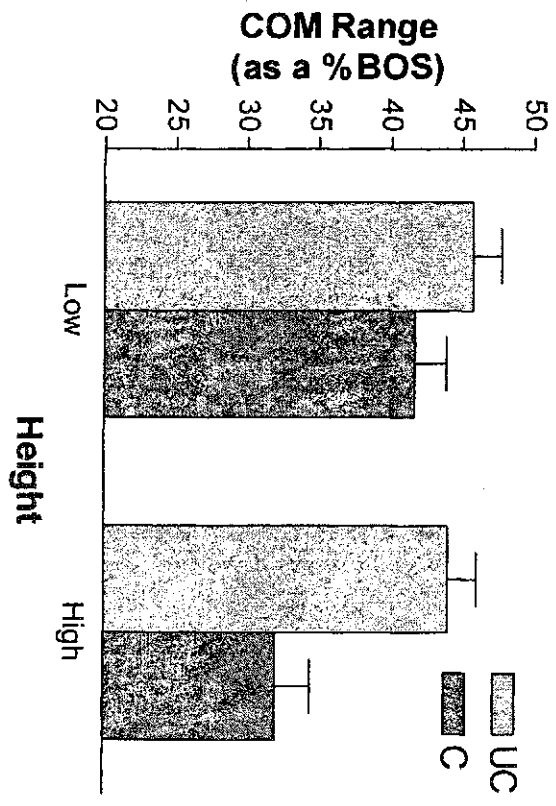
Fig. 3.7.3
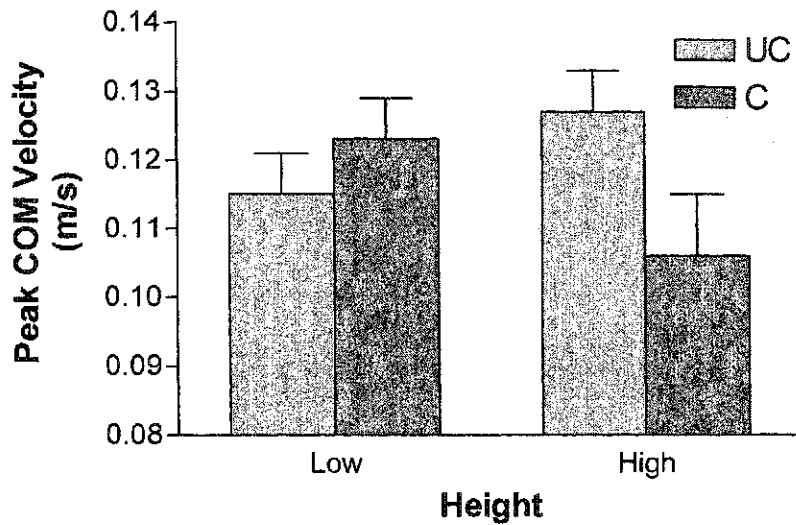
Figure 3.7: These graphs illustrate the COM 1) peak position 2) range, and 3) peak velocity on four conditions of postural threat. Note that significant HxC interactions indicate decreases in COM kinematics in the most threatening conditions.

Significant multivariate main effects for Height and Constraint emerged in the COP ($F(3,21)$=8.05, p=0.001, Λ=0.465 and $F(3,21)$=12.75, p=0.000, Λ=0.354 respectively) and COM ($F(3,21)$=4.99, p=0.009, Λ=0.584 and $F(3,21)$=20.36, p=0.000, Λ=0.256 respectively) data. Univariate RM ANOVAs revealed restricted peak COP and COM positions and smaller ranges of COP and COM displacement in the high versus low (PeakCOP: $F(1,23)$=25.40, p=0.000, PeakCOM: $F(1,23)$=15.12, p=0.001, RangeCOP: $F(1,23)$=23.87, p=0.000, and RangeCOM: $F(1,23)$=12.50, p=0.002) and in the constrained versus unconstrained (PeakCOP: $F(1,23)$=24.46, p=0.000, PeakCOM: $F(1,23)$=64.83, p=0.000, RangeCOP: $F(1,23)$=40.56, p=0.000, and RangeCOM: $F(1,23)$=39.96, p=0.000) conditions.

Furthermore, significant Height X Age ($F(3,21)$=5.57, p=0.006, Λ=0.557) and Constraint X Age ($F(3,21)$=4.12, p=0.019, Λ=0.629) interactions were revealed in the COM but not COP data. Follow-up univariate analysis indicated a significant interaction between

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Height and Age in PeakVelCOM (Fig. 3.8; $F(1,23)=5.49$, $p=0.028$).
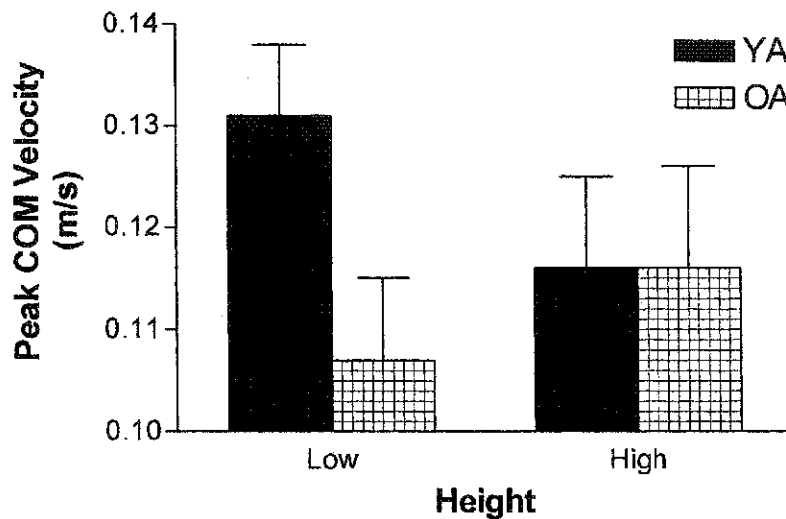
Figure 3.8: Peak COM velocity of younger and older adults in low and high conditions. Note that during high conditions, peak COM velocity is reduced, particularly in older adults.

Younger adults significantly reduced horizontal COM velocity from low to high conditions ($t(13)=3.09$, $p=0.009$). As well, older adults moved slower in low conditions than younger adults ($t(23)=2.09$, $p=0.048$). Signficant interactions between Constraint and Age emerged in the PeakCOM (Fig. 3.9.1; $F(1,23)=9.65$, $p=0.005$) and RangeCOM (Fig. 3.9.2; $F(1,23)=10.88$, $p=0.003$) measures.

Fig. 3.9.1
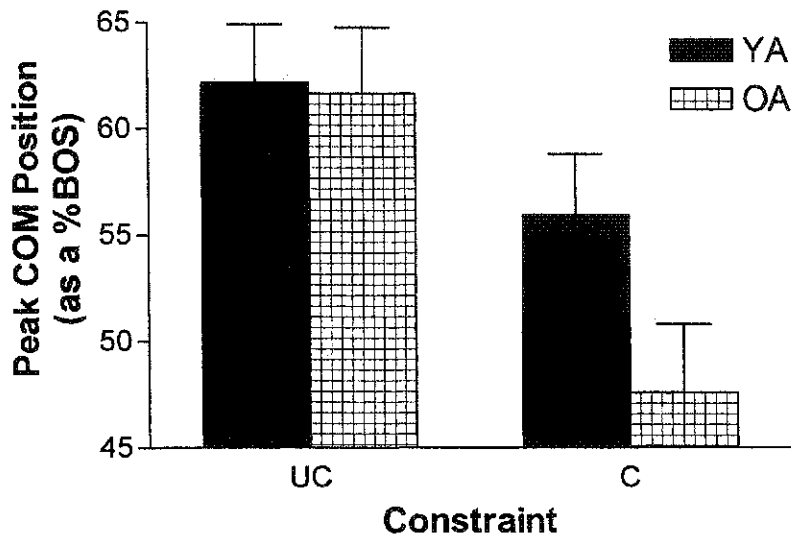
Fig. 3.9.2
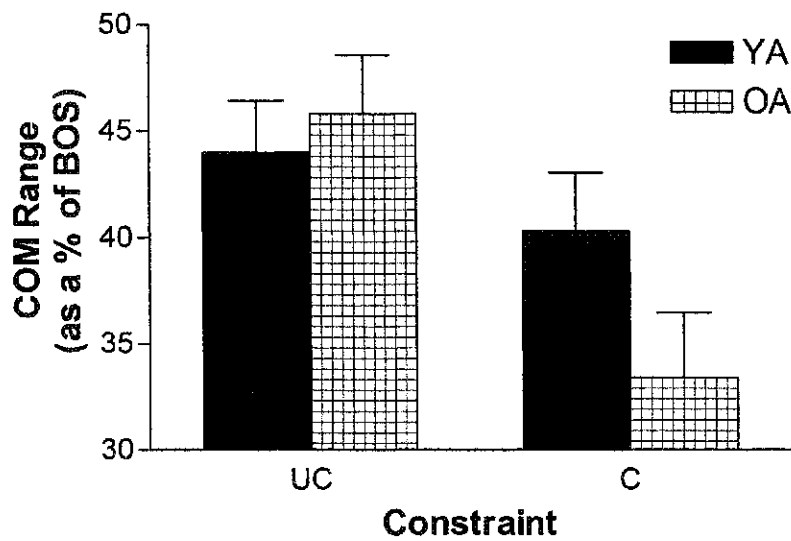
Figure 3.9: These graphs illustrate COM 1)peak position and 2) range among younger and older adults when in unconstrained and constrained positions. Note that older adults are more affected by the constraint than younger adults.

Both younger and older adults restricted their peak COM position (Younger: $t(13)=4.11$, $p=0.001$; Older: $t(10)=6.72$, $p=0.000$) and range of COM displacement (Younger: $t(13)=2.28$, $p=0.040$; Older: $t(10)=5.72$, $p=0.000$) in the constrained compared to the unconstrained conditions. The data also showed a trend for older adults to have more posterior peak COM positions than younger adults in the constrained conditions ($t(23)=1.95, p=0.064$).

(2) Linear and Angular Joint Kinematics

A significant interaction between Height and Constraint emerged in the linear (Fig. 3.10.1; $F(2,22)=7.11$, $p=0.004$, $\Lambda=0.607$;) and angular (Fig. 3.10.2; $F(2,22)=4.65$, $p=0.021$, $\Lambda=0.703$) hip displacement measures. Post-hoc analysis revealed significant Height X Constraint interactions in both the linear ($F(1,23)=14.65$, $p=0.001$) and angular ($F(1,23)=9.61, p=0.005$) peak positions. Participants attained a more posterior peak position and a smaller maximum point of hip flexion in the HC compared to the other three conditions (LinearHipPeak$_{LUC}$: $t(24)=4.18, p=0.000$, LinearHipPeak$_{LC}$: $t(24)=5.87, p=0.000$, LinearHipPeak$_{HUC}$: $t(24)=2.49$, $p=0.020$; AngularHipPeak$_{LUC}$: $t(24)=-2.76$, $p=0.011$, AngularHipPeak$_{LC}$: $t(24)=-2.82$, $p=0.009$, AngularHipPeak$_{HUC}$: $t(24)=-3.78$, $p=0.001$). Participants reduced horizontal peak hip position by 19% and reduced the maximum point of hip flexion by 11%.

Fig. 3.10.1
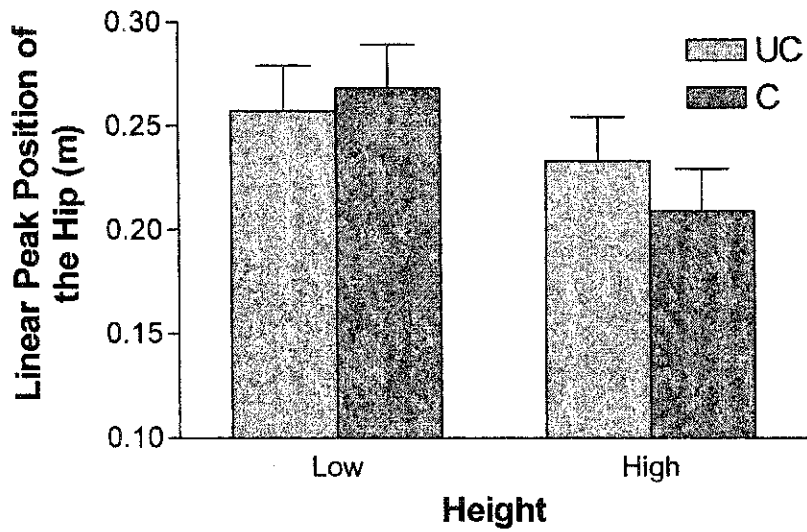
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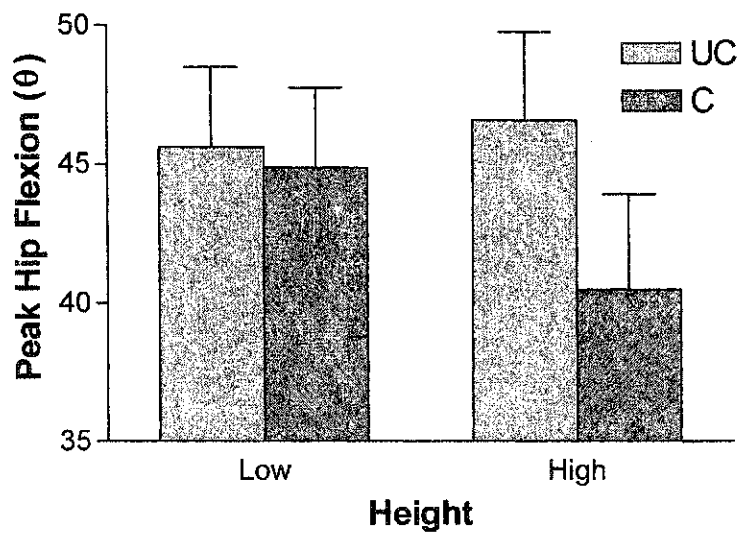
Figure 3.10: These graphs illustrate the 1) linear peak position of the hip and 2) the maximum point of hip flexion under four conditions of postural threat. Note that a significant HxC interaction indicates that both variables are reduced in the HC compared to the LUC, LC, and HUC conditions.

Multivariate main effects for Height $(F(2,22)=7.40$, $p=0.001$, $\Lambda=0.358)$ and Age $(F(3,23)=5.75, p=0.004, \Lambda=0.551)$ emerged for measures of linear hip displacement. Follow-up univariate analyses revealed that the effect of Height was supported in Peak $(F(1,25)=9.67, p=0.005)$ and Range $(F(1,25)=16.05, p=0.001)$ measures. The effect for Age was supported in Peak $(F(1,25)=16.32, p=0.000)$ measures. In the high conditions, all participants attained a smaller peak position and achieved a smaller range of displacement compared to the low conditions. No significant changes were observed in the linear displacement measures of the shoulder $(p>0.05)$.

In the angular hip measures, a multivariate main effect for Constraint $(F(2,22)=3.92$, $p=0.035$, $\Lambda=0.738)$ emerged. Participants reduced the maximum point of hip flexion $(F(2,22)=8.08, p=0.009)$ and the absolute range of hip flexion $(F(2,22)=3.72, p=0.066)$ in the constrained versus the unconstrained conditions. There were no significant interactions for age; however, individual subject analysis indicated that while the same number of younger (71%) and older (64%) adults reduced their maximum point of hip flexion in the HC compared to the LUC conditions, fewer older adults (36%) reduced the range of hip flexion beyond a 5% threshold minimum from LUC to HC conditions than younger adults (64%).

(3) Muscle Onset Latencies

Muscle onset latencies were significantly modified under conditions of postural threat. These changes were evidenced by a significant interaction between Height and Constraint in the SOL (Fig. 3.11.1; $F(1,17)=9.47$, $p=0.007$), BF (Fig. 3.11.2; $F(1,17)=4.46$, $p=0.050$;), and ES (Fig. 3.11.3; $F(1,17)=8.40$, $p=0.010$;) muscle onset latencies. Follow-up analyses revealed earlier muscle activation of SOL, BF and ES muscles in the HC compared to the other three conditions (SOL_{LC}: $t(18)=-2.074$, $p=0.053$; BF_{LUC}: $t(18)=-3.96$, $p=0.001$,

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BF$_{LC}$: $t(18)=-2.40$, $p=0.027$, BF$_{HUC}$: $t(18)=$, $p=0.$; ES$_{LC}$: $t(18)=-2.39$, $p=0.028$, ES$_{HUC}$: $t(18)=-2.02$, $p=0.058$). In fact, the onsets decreased by 14%, 65%, and 28% for SOL, BF, and ES muscles between the LUC and HC conditions. Moreover, a significant main effect for Height indicated earlier activation of the BF muscle in the high versus the low (0.363 vs. 0.572s) conditions. There was also a significant main effect for Age, indicating that older adults had a longer delay in SOL muscle activation ($F(1,17)=11.80$, $p=0.003$) than younger adults.

Fig. 3.11.1
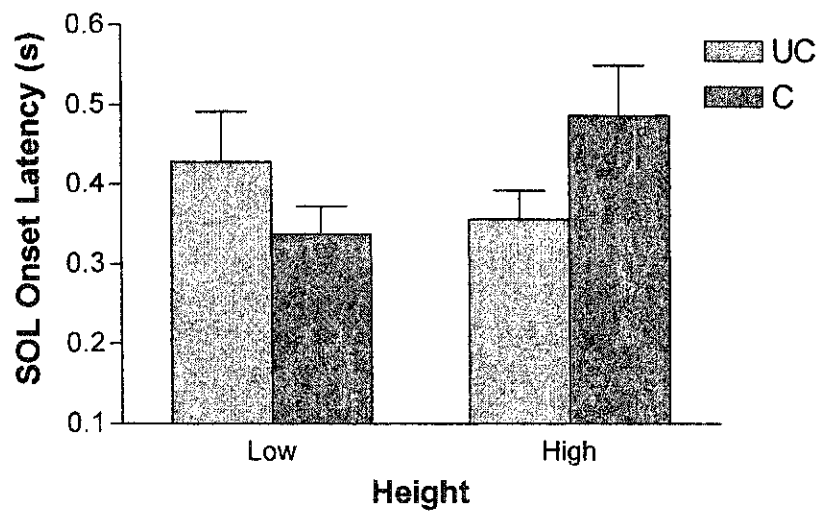
Fig. 3.11.1
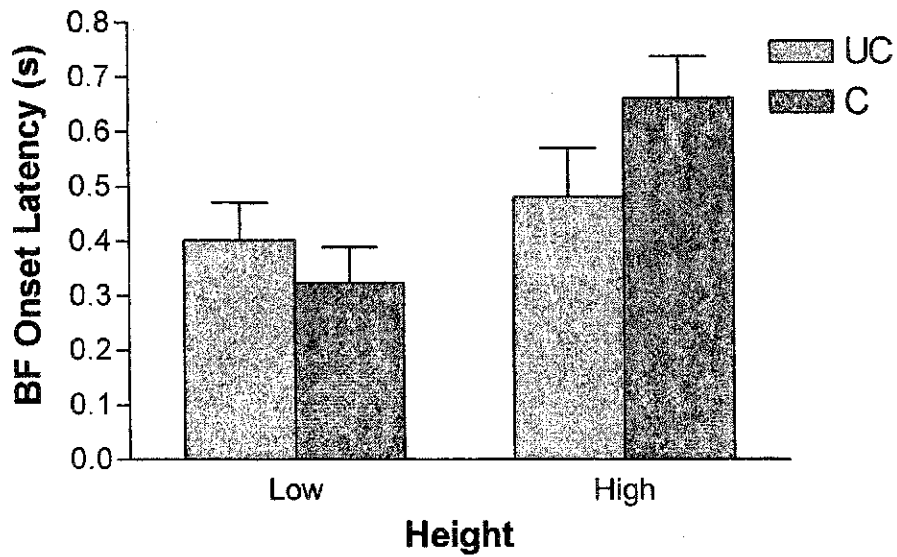
Fig. 3.11.2
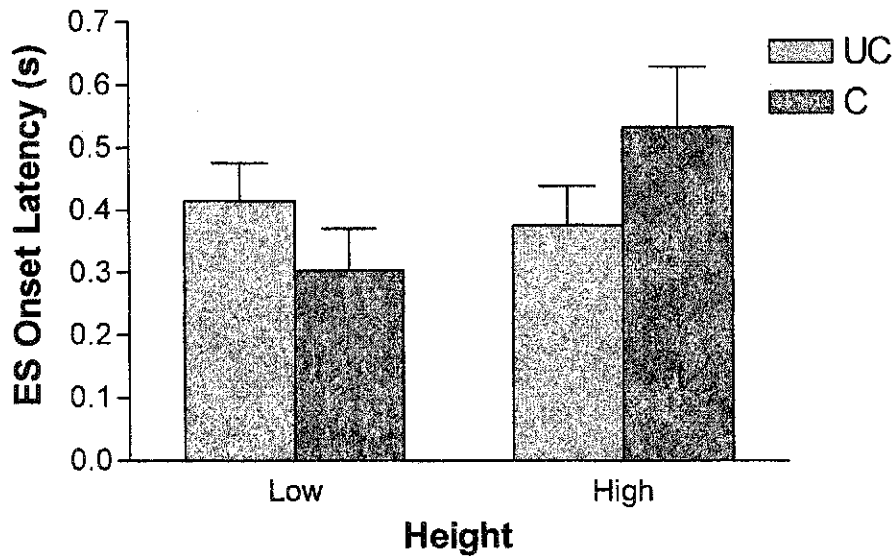
Figure 3.11: These graphs illustrate the muscle onset latencies of the 1) Soleus 2) Biceps Femoris and 3) Erector Spinae with respect to peak COM velocity under four conditions of postural threat. Note that larger values in onset latencies correspond to earlier activations. Thus, significant HxC interactions in the postural muscles indicate earlier activations in the most threatening conditions.

(4) Muscle Activity Levels (iEMG)

The imposed threat influenced the magnitude of muscle activity. A Height by Constraint interaction $(F(1,20)=15.92, p=0.001)$ revealed significant reductions in SOL muscle activity in the condition of greatest postural threat (Fig. 3.12; SOL_{LUC}: $t(21)=4.63$, $p=0.000$, SOL_{LC}: $t(21)=4.31$, $p=0.000$, SOL_{HUC}: $t(21)=6.13$, $p=0.000$). SOL activity was reduced by 15% between the LUC and HC conditions. Furthermore, the magnitude of SOL muscle activity was reduced in the high versus the low (1.104 vs. 1.026mV) and the constrained versus the unconstrained conditions (1.113 vs. 1.018mV) as evidenced by the significant main effects for Height $(F(1,20)=13.89, p=0.001)$ and Constraint $(F(1,20)=31.81, p=0.000)$. A significant main effect for Constraint $(F(1,20)=9.79, p=0.005)$ was also observed in the amplitude of ES muscle activity indicating reduced activity in the constrained compared to the unconstrained conditions (0.772 vs. 0.738mV).
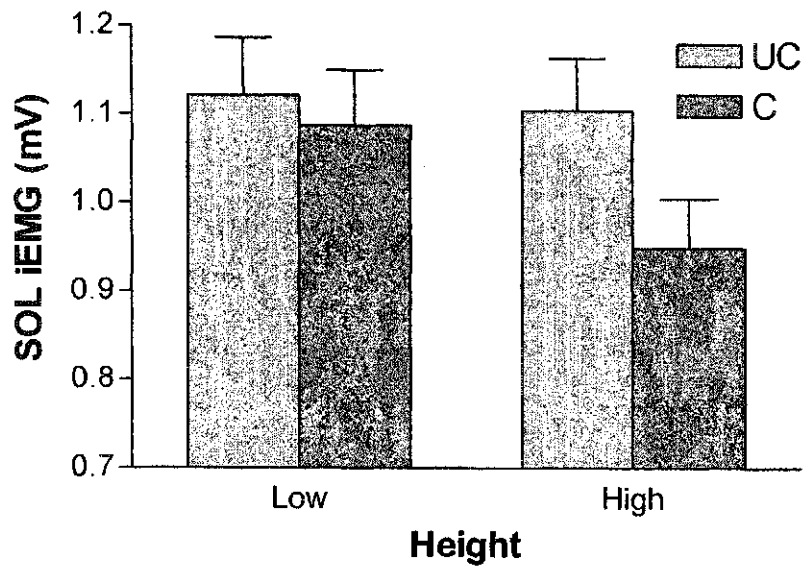
Figure 3.12: SOL iEMG muscle activity under four conditions of postural threat. Note that a significant HxC interaction revealed that SOL muscle activity decreased in the HC condition compared to the other three conditions of postural threat.

D. Discussion

The purpose of this study was to identify age-related modifications to postural control during preparatory and focal movement phases of a forward reach under conditions of increased postural threat. Our results confirmed that the environmental manipulation of postural threat successfully heightened physiological arousal and adversely affected reaching performance in all participants, particularly the older adults. Postural threat influenced the control of posture during both the preparatory and the focal movement phases. Although the magnitude and duration of COP APA were not altered under conditions of postural threat, participants, particularly older adults, adopted a backward lean strategy prior to movement initiation. We have interpreted this behaviour as a conservative adaptation to the potential consequences of instability because it allows for a greater range of AP COM displacement before the limits of stability are exceeded, should a balance disturbance occur. Our findings also indicated similar conservative modifications to postural threat in measures of COP and COM kinematics during the focal movement. More posterior peak positions, smaller ranges of displacements, and slower velocities during the reach action were observed for the COP and COM among all participants in the most threatening condition. Of interest, older adults demonstrated significantly larger posterior shifts in COM position in the threatening conditions compared to younger adults. We propose that the conservative accommodations in COP and COM kinematics observed in older adults during the focal movement are due to a preparatory extension of the hip joint prior to the reach action. No age-related effects of postural threat were found in the patterns of muscle activation that serve to regulate postural control. Interestingly, however, all participants did show earlier activations of the postural muscles and smaller amplitudes of SOL muscle activity as threat

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increased. We suggest that the observed threat-induced modifications to postural control between younger and older adults were due to differences in the kinematics of reaching strategy.

1. What are the kinematic consequences of postural threat during the preparatory phase of a forward reach?

Postural threat altered the starting position of the COP and COM in all participants. Participants demonstrated posterior shifts in COP and COM positions, concurrent with a backward lean strategy, under conditions of postural threat. This preparatory backward shift in start position has been well documented in previous studies that incorporate the postural threat paradigms during quiet standing (Carpenter et al., 2001; Adkin et al., 2000; Carpenter et al., 1999; Polych & Brown, Study 1), prior to external perturbations (Brown & Frank, 1997), and prior to voluntary movements (Adkin et al., 2002). This backward lean strategy is considered to be a conservative modification for the regulation of postural control because the COM is positioned further from the anterior edge of the platform. Consequently, this adjustment serves well to reduce the likelihood of a forward fall. Further support for a backward lean strategy was illustrated by the posterior shifts in the horizontal starting position of the shoulders and hips observed in the HC conditions.

Interestingly, older adults were more affected by the imposed postural threat than younger adults. During high conditions, older adults adopted more posterior COP and COM start positions than during low conditions while younger adults maintained similar start positions across height conditions. This finding suggests that the effects of postural threat were more pervasive for older adults compared to younger adults. It is possible that older adults are leaning further back in high conditions in an effort to further minimize the probability of a forward fall. This finding was particularly intriguing because prior research

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from our laboratory did not reveal any age-related effects of postural threat on the proactive accommodations for the maintenance of upright standing (Polych & Brown, Study 1). We suggest that the observed age differences in the start position of COP and COM under conditions of increased postural threat reflect task-dependent changes related to anticipatory postural control. In particular, it appears that the potential for instability during execution of a reaching task is perceived by older adults as sufficient to demand greater proactive modifications to postural control than observed in younger adults.

Contrary to our hypothesis, the characteristics of the COP APA were not affected by postural threat. Neither the magnitude nor the duration of COP APA changed under the different conditions of postural threat. These results are contrary to findings by Adkin and colleagues (2002) who found reduced magnitude and rate of postural adjustments during a rise-to-toes task as postural threat increased. During a rise-to-toes task, the COP APA is critical for the initiation of movement. As forwarded by Adkin et al. (2002), a larger COP APA causes greater displacement and acceleration of the COM, which if inappropriately arrested, will cause a forward fall to occur. A COP APA of smaller magnitude and velocity will allow for a concomitant reduction in the movement and acceleration of the COM. Thus, in the rise-to-toes paradigm, the observed changes in the COP APA would serve well to reduce the risk of falling. However, current research on anticipatory postural control during a reaching movement suggests that the COP APA may serve a dual purpose: 1) its most recognized function, a preparatory adjustment made to stabilize the COM prior to the destabilizing action of the focal movement (Cordo & Nashner, 1982; Bouisset & Zattara, 1981; Belen'kii et al., 1967), and 2) to aid in the initiation of the focal movement (Stapley et al., 1999; Stapley et al., 1998). Because of these opposing functions, it may be difficult to

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modify the characteristics of the COP APA without detrimentally affecting postural stability. Further research is warranted to address the function of COP APA during forward reaching.

2. What are the kinematic consequences of postural threat during the focal movement phase of a forward reach?

Participants modified COP and COM kinematics during the focal movement phase under conditions of postural threat. In particular, participants demonstrated restricted peak COP and COM positions, smaller ranges of COP and COM displacement, and slower COP and COM velocities under threatening conditions. We have interpreted these findings to indicate that the CNS made conservative accommodations to the control of posture when the potential consequences of a fall were more severe. The restricted range and maximum forward position of the COM minimises the probability that the COM will exceed the BOS and cause a forward fall, and the decreased COM velocity means smaller amplitudes of muscle activity are required to control a slower moving COM. These findings imply that the CNS is executing a tighter control over posture in conditions of increased postural threat. These results are congruent with previous studies that have investigated the effects of postural threat on anticipatory postural control (Adkin et al., 2002) and reactive postural control (Brown & Frank, 1997) where conservative adaptations to the control of the COM were observed in the most threatening condition.

Age-related differences in the COM kinematics also emerged under conditions of postural threat. As observed in the start position data, older adults showed larger postural adjustments to increased postural threat than younger adults. In particular, older adults attained more posterior peak COM positions than younger adults in the more threatening conditions. Therefore, older adults may be allowing for larger "safety buffers" between the

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COM and the limits of stability because they have more negative perceptions of their ability to perform the task without falling than younger adults.

3. What are the effects of postural threat on reaching strategy?

Participants altered their postural strategy under conditions of postural threat. When the consequences of a fall were more severe, participants constrained their forward movement as evidenced by restricted peak hip positions and maximum points and ranges of hip flexion. We interpreted this finding to indicate that participants did not show as much decoupling at the hip joint in more tenuous conditions. Because hip flexion generates large moments about the hip joint, the potential exists for larger displacements and accelerations of the COM to occur if not effectively controlled, these changes may result in a loss of balance. Thus, the reduction in hip flexion would serve well to minimise the rate and amplitude of COM movement. Our findings suggest that the decreases in hip flexion were in fact successful in reducing displacement and velocity of the COM. Therefore, we surmise that this conservative accommodation to postural control may serve to minimise the likelihood of a fall under conditions of increased postural threat.

Inspection of the data demonstrated that adults under 60 years of age and those much older in age (>70 years of age) tended to reduce the range of hip flexion. Those between 60 and 70 years of age tended to maintain range yet still attain the same posterior shift in their point of maximum hip flexion, indicating a more extended hip angle prior to the reach. Indeed, a Chi-square analysis revealed a significant difference for age for range of hip flexion ($X^2(1$, N=25)=4.57, p=0.033) but not for peak hip angle ($X^2(1$, N=25)=0.17, p=0.678). Hip extension may be a compensatory mechanism that allows older adults to perform a backward lean similar to younger adults who achieve this position through

backward leaning at the ankle joint. In this case, the hip extension and subsequent backward lean may enable older adults to shift the COM posteriorly, minimising the probability of a potentially injurious forward fall. Adults over 70 years old frequently suffer from reduced axial flexibility (Schenkman, Shipp, Chandler, Studenski, & Kuchibhatla, 1996) and therefore may not be able to adopt this compensatory mechanism.

4. What are the neuromuscular consequences of postural threat on arresting forward COM movement during the focal movement phase of a reach?

The onset latency and magnitude of activation in the postural muscles were altered under conditions of postural threat. All posterior postural leg muscles, i.e. SOL, BF, and ES were activated earlier relative to peak COM velocity as postural threat increased. Activation of these muscles produces a counter clockwise torque that serves to control and counteract forward displacement of the COM. We speculate that earlier muscle onset latencies restrict the rate of forward COM displacement and ensure that the COM movement is arrested in time to preserve balance.

Contrary to expectation, we observed a reduction in SOL muscle activity. This decrease may reflect the changes in muscle onset during increased postural threat. Because the muscles are active earlier, smaller amplitudes of activity may be required to arrest the forward movement of the COM. Without a concomitant reduction in SOL muscle activity, overcompensation for the destabilizing arm movement may result, creating a greater propensity for backward falls.

5. What are the consequences of conservative adaptations to postural control during forward reaching?

We have interpreted the observed modifications to postural control during a forward reaching task to indicate that participants adopted a more conservative approach to postural

control during tenuous situations where the consequence of a loss of balance is potentially injurious. However, participants did not achieve the same reach distance in the most threatening conditions. Therefore, we propose that reaching ability was significantly impaired by these same conservative postural strategies. It appears that there may be a trade-off between safety and performance. When presented with circumstances under which the consequences of a fall were more severe, participants conformed to a 'posture-first hypothesis' (Marsh & Geel, 2000; Shumway-Cook, Woollacott, Kerns, & Baldwin, 1997) where balance was prioritised over performance on a secondary task, i.e. a forward reach.

As previously discussed, older adults were more affected by postural threat than younger adults. For all kinematic measures, older adults demonstrated more conservative adaptations to postural control under conditions of greatest postural threat; however, they also demonstrated a greater decrement in reach ability between the LUC and HC conditions than younger adults. As previously discussed, older adults may perceive the propensity for injury to be greater in tenuous conditions than younger adults. Therefore, older adults compromise reach performance to increase safety when presented with conditions that may have potentially debilitating consequences in the event of a fall occurrence. These findings are consistent with previous work from our laboratory that has shown support for a posture first hypothesis in older adults under conditions of increased postural threat (Brown et al., 2002).

6. Conclusions

Younger and older adults demonstrate modifications to the control of posture during forward reaching under conditions of increased postural threat. Postural threat conditions increased physiological arousal and caused kinematic and neuromuscular modifications to

reaching strategy that may reduce the likelihood of a potentially injurious fall. In the most tenuous conditions, participants demonstrated reduced hip flexion and earlier activation of postural muscles that arrest forward COM movement. These strategies coincided with decreased displacement and velocity of COM during the focal movement. Interestingly, although these conservative strategies to balance control may increase safety, reaching performance was adversely affected, suggesting a posture-first strategy for postural control under conditions of increased postural threat.

Furthermore, this study provided evidence that older adults are more affected by postural threat than younger adults. Older adults accommodate for the upcoming disturbance through a pre-emptive extension of the hip. Older adults may employ hip-dominant strategies as a mechanism for controlling trunk movement and thereby, decreasing the rate and magnitude of COM movement in more tenuous conditions. However, this strategy may increase the propensity for falls, particularly in the backward direction following a balance disturbance. Therefore, further research is warranted to determine the value of these age-related modifications to postural threat under reactive balance tasks.

IV. GENERAL DISCUSSION

This thesis investigated the age-related modifications to postural strategy under conditions where the consequences of a loss of balance are potentially injurious. Two studies were conducted to assess age-dependent effects of a manipulation of environmental context, or postural threat, on postural strategy. Study 1 examined the effect of postural threat on the regulation of upright stance among younger and older adults. Study 2 explored the age-related effects of postural threat on postural control during the preparatory and focal movement phases of a forward reach. In both studies, participants were exposed to four conditions of postural threat (see Figure 2.1)

A. The Effects of Postural Threat on Arousal

To address the research questions presented in this thesis, it was first necessary to substantiate the claim that postural threat heightened physiological arousal. Galvanic skin conductance (GSC) was used to measure arousal levels. GSC is a sensitive psychophysiological index of changes in the sympathetic autonomic nervous system that measures changes in the electrical conductance of the skin in response to increased sweat secretion (Critchley, 2002; Critchley et al., 2000). As expected, GSC increased as postural threat increased (see Figure 2.3 and Figure 3.2). Thus, the lowest GSC values were recorded in the LUC condition and the highest GSC values were recorded in the HC condition. From previous studies using GSC (Critchley, 2002; Critchley et al., 2000; Ashcroft et al., 1991), we have interpreted these increased arousal levels to indicate increased anxiety about balance.

Furthermore, our results indicate that older adults experienced greater balance anxiety, denoted by higher GSC levels, in response to increased postural threat than younger adults during the forward reaching task (see Figure 3.3). This finding is not surprising

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because a significant proportion of falls in the elderly are due to challenging tasks such as reaching (Nevitt et al., 1991). In fact, older adults report reaching as one of the most difficult activities of daily living (ADLs) to perform without falling (Lachman et al., 1998; Manning et al., 1997; Powell & Myers, 1995; Tinetti et al., 1990). Thus, it follows that our manipulation of postural threat would foster more anxiety in older compared to younger adults.

Self-reported measures of anxiety were also collected. Participants were asked to rate their level of balance anxiety on a scale of 1-10. Although very few participants reported feeling anxious, behavioural observations suggested otherwise. In the HC condition, participants, particularly older adults, stepped back from the edge of the platform and grasped the guardrails between trials, an observation not frequently seen in the other three conditions. These findings indicate that participants, especially older adults, did not feel as comfortable standing in conditions where the potential consequences of a fall were more severe. Overall, we concluded that the imposed postural threat was sufficient to heighten arousal and induce an anxiety about balance.

B. Effects of Postural Threat on the Maintenance of an Upright Stance

Postural threat influenced the regulation of postural control during quiet standing. When the potential consequences of a fall were injurious, participants demonstrated three main mechanical and neuromuscular modifications to balance: 1) a backward lean strategy, 2) a tighter control over posture, and 3) increased agonist/antagonist muscle cocontraction. We propose that these changes reflect conservative adaptations to posture that reduce the likelihood of a loss of balance. The backward lean strategy places a larger "safety buffer" between the position of the COM and the anterior edge of the platform. Thus, in the event

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of a perturbation, a greater range of COM displacement can occur before the stability limits are exceeded. The tighter control of posture indicates that the CNS responds to smaller displacements of the COM by generating smaller, more frequent displacements of the COP to restrict the range and rate of COM movement. The above changes in the variability and frequency of postural sway support increased ankle muscle stiffness during conditions of increased postural threat. As forwarded by Winter (2001; 1998; 1996), the CNS imposes stiffness by means of increased antagonist muscle coactivity at the ankle joint; therefore, we propose that the observed increases in ankle muscle cocontraction reflect the mechanism by which the CNS modifies ankle stiffness in response to postural threat.

C. Age-related Effects of Postural Threat on the Maintenance of Upright Stance

The results from this thesis indicate that younger and older adults make the same mechanical modifications to balance under conditions of postural threat. This finding suggests that older adults have maintained the ability to adapt to different environmental contexts. Interestingly, however, we discovered age-dependent differences in the neuromuscular mechanisms underlying the control of posture. Although all participants demonstrated increased antagonist muscle coactivity at the ankle joint, older adults showed additional antagonist muscle cocontraction at the knee joint as well. These findings suggest that the ankle stiffness strategy in older adults may not be sufficient to produce the same mechanical adaptations to threat observed in younger adults. Rather, older adults may require an additional, more proximal mechanism to control the movement of the COM. The adoption of a more proximal control strategy may be a compensatory mechanism for the age-related decline in the ability to control trunk movement (Gill et al., 2001) and to

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attenuate perturbations (Jensen, Brown, & Woollacott, 2001) when the consequences of a fall are more severe.

D. Effects of Postural Threat on the Preparatory and Focal Movement Phases of a Forward Reaching Task

The preparatory and focal movement phases of forward reaching were altered by postural threat. Although the characteristics of COP APA remained constant across conditions of postural threat, participants did make preparatory adjustments to their body position by adopting a backward lean prior to the reach in more threatening conditions. During the focal movement, the range and rate of COM movement were also reduced in the most tenuous conditions. Further analysis of individual joint and muscle activity suggests that these adaptations may be the result of kinematic and neuromuscular changes to reach strategy. Under conditions of increased postural threat, participants restricted the peak position of the hip and reduced the maximum position and range of hip flexion. Decoupling of the hip generates larger moments than those generated by an ankle strategy. Thus, the reduction in hip flexion would serve well to minimise the displacement and acceleration of the COM. Furthermore, the CNS activated the postural muscles responsible for braking the forward movement of the COM earlier in the reach. These conservative adaptations to the kinematics and neuromuscular control of balance function to exact tighter control over COM movement. Because successful reduction in displacement and velocity of the COM were observed, we surmise that this mechanism may serve to minimise the propensity for falls when the consequences of instability present a greater possibility for injury.

However, our results also indicate an interesting relationship between postural control and reach performance under conditions of postural threat. We found evidence for

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a trade-off between postural control and reach performance. In the most tenuous conditions, participants demonstrated the most conservative postural accommodations to balance while also demonstrating the most adverse effects on reach performance. We have interpreted these findings to indicate that participants prioritise posture, at the risk of adversely affecting task performance, when the potential consequences of a fall were more severe.

E. Age-related Effects of Postural Threat on the Preparatory and Focal Movement Phases of a Forward Reaching Task

Contrary to quiet stance findings, older adults did demonstrate more conservative adaptations to balance than younger adults in response to postural threat. During the preparatory movement phase, older adults leaned further back than younger adults in the condition of greatest postural threat. During the focal movement phase, older adults demonstrated a greater restriction of peak COM position in threatening conditions compared to younger adults. Further analysis suggested that the differences observed in COM regulation during focal movement may be due to preparatory changes in the kinematics of reach strategy. Although all participants reduced the maximum point of hip flexion, younger and older adults differed in the changes in hip flexion range in response to postural threat. In contrast to the majority of younger adults, few older adults reduced their range of hip flexion between LUC and HC conditions. These results imply that older adults extended at the hip joint in preparation for a forward reach. This hip extension strategy may explain why prior to the reach, older adults achieved larger backward leans than younger adults in more threatening conditions even though younger adults initially had more posterior hip and shoulder positions. Thus, we have interpreted these findings to mean that similar to upright stance (Polych and Brown, Study 1), older adults make use of a more

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proximal, hip strategy for preparatory postural adjustments, perhaps in combination with the more distal, ankle strategy observed in younger adults. These findings are congruent with previous research on reactive balance control that has shown that when instructed to use a "feet-in-place" balance responses, older adults are more likely to respond to external perturbations with a "hip" strategy regardless of perturbation characteristics (Manchester et al., 1989; Horak et al., 1989).

F. Implications for Fear of Falling

All participants who participated in this thesis were healthy and free from any contraindications, i.e. neurological and orthopaedic conditions that could affect their ability to maintain postural control. Furthermore, none of the participants reported any aversions to heights or any fear of falling during activities of daily living. Given these demographics, we cannot make conclusions about the effects of fear of falling in the elderly on the regulation of postural control. Although this thesis cannot be generalised to fearful populations, these findings do enable us to make predictions that may direct future research in this area.

Results from Study 1 indicated that the CNS modifies the regulation of upright stance under conditions of increased postural threat. In particular, we suggest that the CNS adopts an ankle stiffness strategy through the cocontraction of antagonist muscles in an effort to gain tighter control over posture. Recent research has found that fearful older adults possess greater levels of antagonist muscle cocontraction and demonstrate increased stiffness compared to their nonfearful counterparts (Okada, Hirakawa, Takada, & Kinoshita, 2001). Although stiffening appears to reduce the risk of falling, researchers are still unclear on the effects of a stiffer system on the ability to recover balance from perturbations. In

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fact, increased stiffness may make individuals more susceptible to falls following a balance disturbance. Okada and colleagues (2001) proposed that the increased stiffness observed in fearful older adults may reduce the ability to attenuate abrupt decelerations, resulting in increased displacement and consequently, an increased risk for a fall. Further work is necessary to elucidate the consequences of a stiffer system on reactive balance, particularly among fearful older adults.

Results from Study 2 indicated that older adults may adopt different kinematics of reach strategy that younger adults. Our findings suggest that during voluntary movement in tenuous situations, older adults adopt hip-dominant strategies, rather than the ankle strategies employed by younger adults, to achieve conservative adaptations that reduce the propensity of a forward fall. From a biomechanical perspective, a hip strategy may be beneficial to older adults because smaller moments are needed to control the movement of the trunk. However, in the event of an unexpected perturbation, this strategy may actually increase the risk of falling, particularly backwards, due to the misalignment of the body segments. We would predict that a hip-dominant strategy associated with anticipatory postural control may place fearful older adults at a greater risk of falling because they already suffer from poorer balance ability than their nonfearful counterparts (McAuley et al., 1997; Baloh et al., 1995; Baloh et al., 1994; Franzoni et al., 1994; Tinetti et al., 1990).

Study 2 also demonstrated support for a posture-first hypothesis in older adults under conditions of increased postural threat. Older adults prioritised postural control at the expense of reach performance. In particular, conservative adaptations that may reduce fall risk were performed regardless of the adverse effects on reach performance. These findings have implications for the quality of life of fearful older adults. If older adults with a fear of falling are unable to successfully perform activities of daily living, these individuals may

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suffer from depression, further activity restriction, and ultimately, a loss of independence (Howland et al., 1998; Cumming & Nevitt, 1994; Timiras, 1994; Arfken et al., 1994; Howland et al., 1993; Black et al., 1993; Grimley Evans, 1992; Tinetti et al., 1988; Murphy & Isaacs, 1982). In a recent study by Salkeld and colleagues (2000), older adults reported that independence was of primary importance to their quality of life. In fact, many of these individuals stated that they would rather face death than institutionalization (Salkeld et al., 2000). Therefore, it is pertinent that older adults with or without a fear of falling maintain the ability to perform everyday tasks such as reaching.

G. Research Applications

The findings from this thesis indicate that an imposed postural threat increased anxiety about balance and influenced postural control among younger and older adults. The main application of this thesis is to provide insight into the behavioural modifications to postural control in fearful older adults. Study 2, in particular, has direct application to real world situations. The effects of increased anxiety during a functional forward reach parallel common activities of daily living that may require older adults to reach while standing on a step stool or step ladder. The results from this study indicate that fearful older adults might have a greater propensity for falls due to the potentially harmful effects of a hip-dominant strategy during voluntary movement.

H. Limitations

The older adults who participated in this thesis were healthy and free from any contraindications that may affect the ability to maintain postural control. Moreover, none of these older adults possessed a fear of falling or an aversion to heights. Therefore, although the conditions of postural threat were sufficient to heighten physiological arousal, we cannot establish whether the environmental contexts imposed an anxiety identical to that of fear of falling in these adults. Furthermore, participants were required to wear a safety harness that would prevent any serious injuries from occurring in the event of a fall. Because the safety harness reduced the potential consequences of instability, participants may not have perceived the conditions as threatening.

Results of Study 1 are limited because a direct measure of stiffness was not made. In this experiment, stiffness was inferred from changes to the kinematic and neuromuscular measures of postural control. According to the inverted pendulum model of postural

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control, decreased variability and increased frequency of postural sway reflects increased ankle stiffness (Winter et al., 1998). In addition, researchers have speculated that stiffness is achieved via increased cocontraction of agonist/antagonist muscle pairs (Maki et al., 1991; Kearney & Hunter, 1990). Although our findings provide indirect support for stiffness, we cannot confirm that a stiffness strategy was adopted in tenuous conditions. Furthermore, without an actual measure of stiffness, it is difficult to conclude whether the increased cocontraction of antagonist muscles was a mechanism for stiffness or simply, a secondary consequence of a backward lean strategy. Future research is warranted to substantiate the hypothesis that participants increase ankle stiffness, via antagonist muscle cocontraction, in response to postural threat.

Results from Study 2 are limited by the instructional constraints of our forward reach protocol. Our goal was to examine forward reaching from a functional perspective such that our task mimicked reaching during real world situations of daily living. Therefore, the reach involved the combination of two common proactive balance tasks: 1) an arm raise and 2) an arm extension. These tasks utilise different preparatory muscle synergies, and therefore, EMG analysis of the postural muscles during the APA was very difficult. As such, this thesis only investigated the postural muscles involved in arresting the forward movement of the COM during the focal movement phase. Future investigation into the age-related effects of postural threat on anticipatory postural control during arm raise and arm reach is needed. From previous work on proactive postural control, we predict that postural muscles would be activated earlier to allow for effective preparation for the impending disturbance. Contrary to findings by Adkin and colleagues (2002), we would also expect increased latencies between the onset of the postural and the onset of focal muscle under conditions of increased postural threat, particularly among older adults.

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The results of both Study 1 and Study 2 may have been compromised by a structural limitation of the experiment. Because the elevating platform used in these experiments was opaque, participants in the unconstrained conditions (located in the middle of the platform) were unable to visualise the threat imposed by the height of the platform. In the constrained conditions, however, participants were at the edge of the platform where they were able to visualise the threat. Thus, participants were given different visual information in the constrained versus the unconstrained conditions. Because visual information is so critical for postural control, particularly in the case of older adults, it would be useful to replicate this experiment using a transparent elevating platform or preferably, designing a 'false floor', i.e. an opaque object, that will breakaway if any force is applied to it, added to the edge of the platform. Both approaches are applicable to real world situations; however, the latter allows for a dissection of the effects of postural threat with and without visual feedback.

I. Conclusions

The findings from this thesis indicate that environmental contexts that increase anxiety about balance influenced the regulation of postural control among younger and older adults. When the potential consequences of a fall were more severe, adults made conservative mechanical and neuromuscular modifications to balance that would serve well to reduce the likelihood of a fall. During upright standing under conditions of increased postural threat, participants adopted a backward lean to increase the permitted range of COM displacement before a loss of balance would occur. Furthermore, the CNS employed a tighter rein of control over posture, possibly through the coactivation of antagonist muscles consistent with a stiffness strategy (Winter et al., 2001; Winter et al., 1998; Winter et al., 1996). Although these adaptations appear to be conservative accommodations to

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postural threat, the potential consequences of stiffening on recovery following a balance disturbance are not well known. However, recent research has indicated that a stiffer system may be detrimental to reactive balance control (Okada et al., 2001).

Postural threat altered the control of posture during preparatory and focal movement phases of forward reaching among younger and older adults. Under conditions of increased postural threat, participants made preparatory adjustments to their body position by adopting a backward lean prior to the reach. The magnitude and duration of the COP APA, however, remained constant across all conditions of postural threat. During the focal movement, COM displacement and velocity were restricted, perhaps to minimise the likelihood of a fall event under more tenuous conditions. We surmise that these changes may be due to kinematic and neuromuscular modifications to reaching strategy. In particular, the rate and amplitude of COM movement may have been reduced through the combination of restricted maximum positions and ranges of hip flexion and earlier onsets of postural muscle that arrest forward COM movement.

Our results revealed an interesting relationship between postural control and reaching performance. The adoption of conservative modifications corresponded to decrements in reach distance. In more tenuous conditions, increased the use of postural accommodations that may increase safety at the expense of reaching performance. Thus, participants conformed to a posture-first strategy for postural control under conditions of increased postural threat.

Interestingly, this thesis revealed that older adults were more affected by postural threat than younger adults. As such, older adults appeared to use different strategies for achieving conservative accommodations to postural control than younger adults under threatening conditions. Our findings indicate that older adults demonstrated increased use

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of proximal balance control strategies under conditions of postural threat. During upright stance, older adults demonstrated increased cocontraction of antagonist muscles at the knee and ankle joints compared to younger adults who increased antagonist muscle coactivity at the ankle joint only. These results suggest that older adults may increase stiffness at the knee joint, as well as the ankle, to provide more proximal control over the movement of the COM. Although increased stiffness may enable a tighter control over posture, it remains unclear whether stiffness is helpful or harmful when exposed to unexpected perturbations. During forward reaching, older adults employed hip-dominant strategies to restrict the rate and magnitude of COM movement in the most tenuous conditions. A hip strategy may allow older adults to control the movement of the trunk segment by generating moments of smaller magnitude. This strategy may be beneficial to older adults because they suffer from age-related declines in the ability to control trunk movement. However, hip extension may cause a misalignment of body segments that may actually increase the risk of falling, particularly in the backward direction. Further research is necessary to elucidate the consequences of these age-related modifications to balance recovery under conditions of increased postural threat.

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APPENDIX 1: INFORMED CONSENT FORM

You are invited to participate in a research study performed by Ryan Sleik, B.Sc. and Dr. Lesley A. Brown in the Department of Kinesiology at the University of Lethbridge. The purpose of this study is to determine the effects of fear of falling on the cognitive demands associated with postural control. The experiment requires that you perform two different cognitive tasks on an elevated platform. The results from this research may help determine why people who have a history of falls are more likely to fall again and could lead to information that can help reduce the fall rate in older populations.

Should you consent to participate in this study, you will be asked to stand on the edge of an elevated platform (about 1.5m off the ground) and perform trials of simple cognitive tasks. Rest assured that we will not let you fall! An assistant will be there to guard you and you will be wearing a safety harness.

In order to understand you body movement during the testing procedure, we will place small reflective balls on the surface of your skin. In addition, an electrode will be placed on your chest to record your heart rate. A computer will be used to collect information from these sensors. We will also use video to record your movement during testing.

Potential Risks: There is a risk that you may lose your balance during the testing. However, to reduce this risk, an attendant will guard you and you will be wearing a safety harness. There is also a slight risk that skin irritation could arise from tape applications. This risk is minimised by using hypo-allergenic products when adhering the reflective markers and electrodes. There is also a risk of losing confidentiality or information. This risk will be minimised by assigning each subject to a code of letters and numerals. All personal subject information will be locked in a file cabinet that can only be accessed by researchers involved in this study.

To ensure visibility of joint markers and to prevent interference with the muscle sensor wires, we ask that you wear a short sleeved shirt, blouse, or t-shirt. If you do not have these items, one will be provided for you. You will be provided with a clean pair of specialised shorts (lycra cotton) to wear during testing. A private change facility is available on-site, and testing will be conducted in a private laboratory with a maximum of three research personnel present.

Any information that is collected during this study will be held confidential and will not be disclosed without your permission. We may, however, wish to use the video tape recording or graphical illustrations of your movements for research and educational purposes in the future. Your identity would be kept confidential; only your code would be utilised in said cases. If you would like to give your permission at this time for used of this tape record or graphical presentation for research and educational purposes, please place your initials by "yes". If you do not wish to give permission at this time, please place you initials by "no".

Yes _____   No _____

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Your participation is entirely voluntary. Should you decide not to participate in this study, you relationship with the Balance Research Laboratory or any other department of the University of Lethbridge will not be affected in any way. If you have any further questions about this research, please feel free to contact Ryan Sleik at 382-7181. If you have any further questions regarding your rights as a research subject, please contact the Office of Research Services at the University of Lethbridge at 329-2747. No payment can be provided in the unlikely event of injury or a medical problem as a result of your participation in this study. However, basic first aid will be provided at the time of injury and you will be encouraged to consult your physician.

Your signature on the attached page indicates that you have read and understood the information provided above, that you willingly agree to participate, and that you understand that if you withdraw your consent at any time, you are free to discontinue participation without penalty.

I have read the attached Informed Consent form and I consent to participate in the "Cognitive Demands and Fear of Falling" research study.

_____   _____
             Printed Name   Date

_____
             Signature

APPENDIX 2: FALLS HISTORY QUESTIONNAIRE

1) Are you afraid of falling during your daily activities?

Y   Sometimes   N

(I am scared to do many things)   (I never fear falling)

10   9   8   7   6   5   4   3   2   1

2) Are you afraid of heights?   Y   N

3) Are there any circumstances that make you feel nervous about losing your balance or may be a cause for a fear of heights? If so, what are they?

4) When was the last time you lost your balance or fell?

5) If you did fall, what was the cause of the fall?

6) Are there any conditions or medications that you believe may affect your balance?

APPENDIX 3: THE SELF PERCEPTIONS OF BALANCE QUESTIONNAIRE

The Self Perceptions of Balance (SPB) questionnaire is composed of the Gait Efficacy Scale (GES; items 1 through 5) and the Activities-specific Balance Confidence Scale (ABC; items 6 through 26)

The Gait Efficacy Scale (McAuley, Mihalko, & Rosengren, 1997)

1) How would you rate your balance?
5. excellent   4. very good   3. good   2. fair   1. poor

2) How much does your balance interfere with your physical activities or general movement?
5. never   4. rarely   3. sometimes   2. usually   1. always

3) How often do you engage in exercise?
5. daily   4. 4-6x/week   3. 1-3x/week   2. monthly   1. never

4) How often are you afraid of falling?
5. never   4. rarely   3. sometimes   2. usually   1. always

5) How confident are you that you can walk about your house without losing your balance or falling?

10	9	8	7	6	5	4	3	2	1

very confident   confident if I am careful   not at all confident

The Activities-Specific Balance Confidence Scale Items (Powell & Myers, 1995)
(Note: the same 10 point confidence scale was applied to all subsequent questions.)

10	9	8	7	6	5	4	3	2	1

very confident   confident if I am careful   not at all confident

6) How confident are you that you can prepare your meals without losing your balance or falling?

7) How confident are you that you can get on and off the toilet without losing your balance or falling?

8) How confident are you that you can get dressed without losing your balance or falling?

9) How confident are you that you can get in and out of a chair without losing your balance or falling?

10) How confident are you that you can answer the door or phone without losing your balance or falling?

11) How confident are you that you can get in and out of bed without losing your balance or falling?

12) How confident are you that you can take a bath (or shower) without losing your balance or falling?

13) How confident are you that you can climb a step stool without losing your balance or falling?

14) How confident are you that you can go to the bathroom at night without losing your balance or falling?

15) How confident are you that you can walk outside at night without losing your balance or falling?

16) How confident are you that you can walk outside when it is rainy or icy without losing your balance or falling?

17) How confident are you that you can go grocery shopping without losing your balance or falling?

18) How confident are you that you can go outside and garden without losing your balance or falling?

19) How confident are you that you can go up stairs with a handrail without losing your balance or falling?

20) How confident are you that you can go down stairs with a handrail without losing your balance or falling?

21) How confident are you that you can go up stairs without a handrail without losing your balance or falling?

22) How confident are you that you can go down stairs without a handrail without losing your balance or falling?

23) How confident are you that you can use an escalator to go up?

24) How confident are you that you can use an escalator to go down?

25) How confident are you that you can get off an escalator easily?

26) How confident are you that you can get on an escalator easily?