

BIOMECHANICS OF POSTURAL STABILITY WHEN ACCEPTING A WEIGHT IN THE OUSTRETCHED HANDS

by

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ABSTRACT

In order to maintain postural stability while standing, humans must control and maintain the centre-of-gravity (COG) within the boundaries of the base of support (BOS). This would be especially challenging when accepting a weight in the hands, which tends to abruptly displace the COG. Yet we perform this task amazingly well. The current thesis examines the biomechanics of maintaining postural stability when accepting a weight in the outstretched hands. I used a weight acceptance paradigm that varied the location of the object to be accepted and the control the subject had over the perturbation rate and onset. I found that subject control over the rate and onset of weight acceptance increased the ability to minimize COG displacement. Furthermore, the results suggested that ability to control postural stability was, in part, related to strength, anthropometrics and neuromuscular variables that govern performance on other postural stability tasks.

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CHAPTER 1: BACKGROUND

1.1 Introduction

Research, statistics and personal experiences have shown us that with age comes an increased propensity for health problems and injury risks. Studies have shown that approximately a third of all individuals over 65 years of age living in the community will fall annually (Campbell, Borrie, & Spears, 1989; S. R. Cummings & Nevitt, 1994; Nevitt, Cummings, Kidd, & Black, 1989). This number increases to 50% after the age of 75 years and for individuals in a nursing home setting (Greubel, Stokesberry, & Jelley, 2002; Tinetti, Speechley, & Ginter, 1988). Although only 1% of these falls will result in a hip fracture more than \$1 million in health care is spent each year as a result of these injuries (Wiktorowicz, Goeree, Papaioannou, Adachi, & Papadimitropoulos, 2001), at an approximate cost of \$26,527 a year to the individual (Wiktorowicz et al., 2001). A reported 89% of all hip fractures result from falling, indicating that decreasing in the occurrence of such accidents in elderly persons could make a substantial contribution to a reduction in the prevalence of this type of injury (S. R. Cummings & Nevitt, 1994). Identifying the various movements that place elderly adults in precarious positions could help us to decrease the probability of falls. Research laboratories have just begun to understand the difficulties that elderly individuals have in completing various destabilizing tasks associated with daily life. Activities which require turning or move the body's mass, such as the acceptance of a weight in

the outstretched hands, are particularly destabilizing (S. R. Cummings, Klineberg, R.J., 1994; Tinetti et al., 1988). During quiet stance, stability is maintained by controlling the whole-body centre-of-gravity (COG) within the boundaries of the base of support (BOS). This control comes from the manipulation of the centre-of-pressure (COP) under the feet. The addition of a weight to the outstretched hands causes flexor torques about the joints, moving the COG closer to the boundaries of the BOS and into a position of greater instability. Therefore it is the aim of this thesis to gain a better understanding of how postural stability is maintained while accepting a weight to the outstretched hands.

1.2 Falls and Aging

Gerontology and biomechanical laboratories have studied the fall and injury risk in elderly persons extensively, developing various lists of health problems and other age-related declines that are key risk factors (Campbell et al., 1989; S. R. Cummings & Nevitt, 1994; Nevitt et al., 1989). These risk factors can be classified in four categories; environmental, physical health/medical, cognitive and biomechanical. Environmental risk factors include the physical space surrounding the individual (i.e. the home) and the various obstacles and tripping hazards present there (Campbell et al., 1989; Nevitt, Cummings, & Hudes, 1991). A person's medical condition could also enhance the chance of a fall. Diseases such as diabetes, osteoarthritis, osteoporosis, Parkinson's disease and hypo- and hypertension have been associated with an increased risk of

falling (Campbell et al., 1989). As well, several of the medications taken to treat these illnesses have side effects that include episodes of vertigo, fatigue, and mental confusion that put the individual at further risk. Biomechanical risks are numerous and are often associated with the type of activities the individual engages in. Past research has shown that falling is most strongly associated with turning and activities that change the position of the whole-body centre-of-gravity (COG) with respect to its base of support (BOS, area enclosed by the contact between the foot and the floor) (S. R. Cummings, Klineberg, R.J., 1994; Tinetti et al., 1988). Cognitive risk factors include dementia (Buchner & Larson, 1987) and a condition often referred to as Fear of falling. Fear of falling, generally described as a constant fear of or lack of confidence in an ability to avoid falls (Legters, 2002), has been associated with a risk of falling in several studies (Howland et al., 1998; Howland et al., 1993; Lachman et al., 1998; Tinetti, Mendes de Leon, Doucette, & Baker, 1994). As of yet, no studies have shown whether this relationship is causal or only resultant. However, Fear of falling has also been reported in elderly that have no history of falling (Tinetti et al., 1994; Tinetti et al., 1988). Furthermore, elderly persons have been seen to restrict their activities after falling or due to a Fear of falling (Friedman, Munoz, West, Rubin, & Fried, 2002; Lachman et al., 1998; Murphy, Williams, & Gill, 2002; Tinetti et al., 1994). While this may decrease the individual's environmental risks, restriction in activity may lead to decreases in strength and

mobility which are also fall risk factors (Fayet, Rouche, Hogrel, Tome, & Fardeau, 2001; Lachman et al., 1998).

Physical limitations due to activity restriction are particularly dangerous for the elderly because aging alone has detrimental effects on the properties of muscles and bones. In the muscles, sarcopenia, typically related to aging, results in loss of muscle fiber mass. Biopsies of the vastus lateralis muscle of elderly females show a decrease in the number of type II muscle fibers compared to younger adult subjects (Larsson, Sjodin, & Karlsson, 1978). Type II muscle fibers are responsible for rapid powerful movements. Therefore, decreases in strength generation abilities experienced by the elderly are due, in part, to this depletion of type II fibers. Furthermore, toe flexor muscles, critical to maximizing the stability limits of functional BOS, see a 28.9% decrease in strength with age (Endo, Ashton-Miller, & Alexander, 2002). Although studies have shown that there is a greater change in lower limb muscles due to age (Reed, Pearlmutter, Yochum, Meredith, & Mooradian, 1991), there also exist similar changes in upper limb muscles. Fayet et al. (Fayet et al., 2001) reported a 36.7% decrease in type II fibers in the deltoid muscles of women aged 70 to 79 compared to women aged 50 to 59. Another study showed that elderly individuals (aged 77 years and older) have about 22% less of their elbow strength compared to young adults (Klein, Allman, Marsh, & Rice, 2002). Decreases in the number of muscle fibers results in reductions in the rate of muscle force production in elderly individuals. Finally, Izquierdo et al. (Izquierdo, Aguado, Gonzalez, Lopez, & Hakkinen, 1999)

further demonstrated that individuals 70 years and older also have decreased explosive force production, which suggests that not only is there a reduction in the fast twitch type II fibers, but there is also a decreased rate of muscle fiber recruitment with age.

The strength of bone is a function of the amount of stress and strain the bone can withstand before breaking (or the area under a stress-strain graph, (Hamill & Knutzen, 1995)). Stress and strain are dependant on several factors including bone density, load application, bone architecture, cross-sectional area and elasticity. With age these factors change, causing the strength of the bone to decrease. Age results in a decrease of the production of osteoblasts which are responsible for bone deposition. Therefore with age comes an increase in the ratio of bone resorption to bone deposition, causing a decrease in bone formation and concurrently a reduction in bone density. This reduction in bone density means that there are fewer trabeculae to create the required bone architecture to ensure flexible but strong bones. Furthermore, the cross-sectional area of the bone is influenced by the bone density and trabecular architecture and women have a 20-30% reduction in their bone cross-sectional area compared to men (Mosekilde, 2000). This material difference is reflected in dissimilarities in bone strength between men and women (Mosekilde, 2000; Seeman, 2001). Aging has also been associated with a loss of elastic components of bone (Rogers, 1982), resulting in brittle bones that are more prone to fracture under smaller loads (Mosekilde, 2000).

These neuromuscular limitations in elderly subjects cause alterations in the body's biomechanical behavior and the individual's mobility during even the most basic activities of daily living.

1.3 Postural Stability Biomechanics

1.3.1 Overview

The prevalence of falls is a consequence of two separate factors, first the number and nature of destabilizing events a person experiences and second the person's ability to recover their balance once it has been perturbed. Accordingly, mobility ultimately depends on the ability to control stability while carrying out various movements regardless of voluntary or unexpected perturbations. Several Biomechanists have focused their research on understanding the neuromuscular strategies used by young adults to maintain postural stability, either in quiet stance or during varying activities.

Postural stability is the control of the body's COG within its BOS (D. A. Winter, Patla, Rietdyk, & Ishac, 2001). The weight of the body on the ground is reflected within the BOS as the vertical ground reaction force at a distance from the ankle joint referred to as the center-of-pressure (COP). During ideal quiet standing the COG and the COP are aligned. However, in reality, quiet standing is characterized by small sways in the COG (Murray, Seireg, & Sepic, 1975; D. A. Winter et al., 2001). In order to counter any forward, backward or lateral angular velocities and whole-body torques that may develop, the COP is constantly moving to catch up with and surpass the COG, creating a torque

about the ankle joint to oppose and force the torque of the whole-body back to zero. Using these principles, characteristics of the COG and the COP are used to monitor the postural control of an individual.

At present, several different parameters of the COP and its relationship with the COG are used by researchers to quantify postural control. Studies have looked at COP and COG excursion (the total distance traveled by the COP during an activity), the COP velocity, and the COP displacement (anterior, posterior or lateral). More complicated measures have included variables described by Patton and colleagues (Patton, Lee, & Pai, 2000) which measure an individual's State Boundaries (the COP displacement and velocity combinations beyond which a fall would be most likely to occur) and Torque Boundaries (range of ankle torques beyond which a fall would be most likely to occur (Patton et al., 2000)). Ultimately, it has been demonstrated that an individual's postural control can best be described through a combination of velocity and displacement variables of COP (Riley, Benda, Gill-Body, & Krebs, 1995). Specifically, velocity vectors of COP provide a better indication of the limitations that the individual has during dynamic exercises. Using these various measures of postural stability, studies have demonstrated that postural control has three different primary roles: preparation for upcoming perturbations (with respect to voluntary movement and movement planning), maintenance of stability during the primary movement and recovery of equilibrium after.

1.3.2 The role of anticipation in postural control

The maintenance of stability during voluntary movements is controlled through the use of two different types of motor control responses, anticipatory (or feedforward) and corrective (also known as compensatory, or feedback-based responses, and are discussed further in Section 1.3.3). The anticipatory responses precede voluntary movement and include particular muscle activation patterns that either help to maintain postural stability by preparing the body for the upcoming perturbation or help to develop the appropriate joint torques required to complete the movement (Crenna, Frigo, Massion, & Pedotti, 1987). When subjects were asked to lean as far forward as possible, anterior muscles (tibialis anterior, the vastus medialis and the rectus abdominus) involved in postural control were activated prior to extensor muscles (soleus, the medial gastrocnemius, the biceps femoris and the erector spinae) used to initiate the forward lean. It was therefore hypothesized that the early activation of the anterior muscles prior to the movement-specific extensor muscles assisted in preparation for the forthcoming forward perturbation to the COG. Activation of anterior postural muscles would cause a forward movement of the COP, in anticipation of the forward movement of the COG. Furthermore, with the COP anterior to the COG an internal forward torque would result assisting in the initiation of the desired forward body lean. Similar early muscle activation patterns are seen in a variety of other activities (Aruin, Forrest, & Latash, 1998; Bouisset, Richardson, & Zattara, 2000; Bouisset & Zattara, 1987; Crenna et al., 1987; W. Liu, Kim, Long, Pohl, & Duncan, 2003; Shiratori & Latash, 2001; P. J.

Stapley, Pozzo, Cheron, & Grishin, 1999; Stelmach, Phillips, DiFabio, & Teasdale, 1989; Toussaint, Commissaris, & Beek, 1997; Toussaint, Commissaris, Hoozemans, Ober, & Beek, 1997). Often referred to as anticipatory postural adjustments (APA), these early muscle activation patterns can be further characterized by anticipatory COP movements. APAs are typically seen 100-150 milliseconds prior to the primary movement (Aruin et al., 1998) which indicates that these muscle activation patterns are timed as a discrete functional component (concerned with postural control) of the overall voluntary focal movement (Aruin et al., 1998). These different functional segments of the movement are most likely mitigated by a common central command (Crenna et al., 1987; Inglis & Woollacott, 1988).

Findings of Aruin, Forrest and Latash (Aruin et al., 1998) that examined the production of APAs during periods of instability indicate that the production of APAs depends not only on the expected perturbation but also the initial stability of the individual. Weight release experiments were carried out with subjects in positions of decreased stability (reduced BOS). With increased instability there was a decrease in the magnitude of the APA (or smaller spikes in the activation of postural muscles). Aruin, Forrest and Latash hypothesized that because the APA itself can be destabilizing, the body may choose to react to the upcoming destabilization rather than prepare for it and cause a potentially greater imbalance to the COG (Aruin et al., 1998). In further experiments, when perturbed while in an inclined position, subjects again produced significantly

smaller APAs compared to those seen when standing upright with a stable BOS (Aruin et al., 1998). The authors concluded that the initial posterior position of the COP, due to the inclined position, was enough preparation for the upcoming destabilization. In other words, the posterior position of the COP would make the APA a redundant safety measure (Aruin et al., 1998).

1.3.3 Corrective postural control

If anticipatory postural responses are not initiated correctly, or if they are insufficient to prevent disruption to the whole-body COG, then the postural control system must rely on corrective postural responses in order to return the body to a position of postural stability. These corrective postural responses are initiated in response to feedback provided to the motor control centres in the central nervous system (CNS), and in particular the brainstem (Kandel, Schwartz, & Jessell, 1991). Feedback information comes primarily from cutaneous proprioceptors, mechanical receptors in joints and muscle receptors (such as Golgi Tendon Organs and muscle spindles). The higher order control centres adjust commands to core postural muscles in order to adjust joint torque magnitudes and the stiffness of muscle. Because these responses are based on feedback, new commands come after the perturbation.

1.3.4 Activity-specific APA roles

APAs and COP movements have been examined with respect to different tasks associated with activities of daily living; including sway, balance recovery (W. Liu et al., 2003), gait, arm raising (Bouisset et al., 2000; Bouisset & Zattara,

1987; Inglin & Woollacott, 1988), jumping (Le Pellec & Maton, 1999), reaching (Kaminski & Simpkins, 2001; P. Stapley, Pozzo, Grishin, & Papaxanthis, 2000; P. J. Stapley et al., 1999; Toussaint, Commissaris, Hoozemans et al., 1997), pulling (W. A. Lee & Patton, 1997; Patton et al., 2000) and load lifting (Holbein & Redfern, 1997; T. H. Lee & Lee, 2003; Shiratori & Latash, 2001; Toussaint, Commissaris, & Beek, 1997; Toussaint et al., 1995). Generally, the characteristics of the APA and the movement pattern of the COP and the COG are dependant on the magnitude and timing of the upcoming perturbation, the type of voluntary movement to be performed and the stability of the individual prior to the perturbation (Bouisset & Zattara, 1987; Horak, Esselman, Anderson, & Lynch, 1984). More particularly, experimental results and analysis suggest that while the amplitude of the APA is dependant on the perturbation itself, the duration of the APA is related to both the perturbation and the initial postural stability. Therefore, there is a necessity to consider both the task to be performed (including associated body position constraints) and the effort demands associated with the related change in postural stability (Bouisset et al., 2000).

On the other hand, the apparent need or purpose of the APA can depend specifically on the forthcoming task. For example, with gait initiation, the APA appears to play a role in the production of the disequilibrium necessary for toe lift off (Bouisset et al., 2000; Breniere, Do, & Bouisset, 1987; Le Pellec & Maton, 1999; Lepers & Breniere, 1995). The APA prior to gait initiation has a much

greater duration than the APA created prior to other voluntary perturbations, such as forward leaning, suggesting that the early muscle activation generates a larger torque than needed to oppose the perturbation to the COG. This increase in torque would help generate the toe lift off required to initiate gait. These longer APA durations are also seen in vertical jumps, which again require initial body disequilibrium at the start of the focal movement (Le Pellec & Maton, 1999).

1.4 Biomechanical Considerations for Reaching

In 1987, Crenna et al. published a paper that examined the movement patterns of body segments during forward and backward voluntary leans (Crenna et al., 1987). The paper described what was referred to as 'axial synergy', where movement of the upper and lower body segments occurs in opposition to each other to effectively stabilize the COG against postural disturbance during the leaning exercises. EMG recordings taken during the trials further suggest that these movement patterns were controlled by a central command and not simply caused by action and reaction forces of the different body segments. Furthermore, studies completed by Pedotti et al. suggest that this synergistic activation of muscles involved in postural control may be a learned strategy picked up in childhood (Pedotti, Crenna, Deat, Frigo, & Massion, 1989). Hyperextension of the trunk and differences in muscle activation patterns between two groups of young adults (trained verses untrained gymnasts) were examined and demonstrated that only the individuals with training had similar anticipatory muscle activation in trunk extension to that seen in all individuals

prior to trunk flexion. It was inferred that the anticipatory muscle activation pattern seen in trunk flexion must be a strategy learned in childhood because trunk flexion is so much a part of everyday activities. On the other hand, the anticipatory activation of muscles in trunk extension is only developed in individuals who complete this motion on a regular basis, such as trained gymnasts.

In his PhD thesis research (Oddsson, 1990), Lars Oddsson examined the control of voluntary hip flexion. Oddsson noted that the centre-of-mass (COM, location of the centre of all body segment masses, commonly used interchangeably with COG) moved posterior prior to initiation of the primary movement to increase the anterior space available for the forward movement of the COM during hip flexion. Moreover, subjects were seen to flex their knees prior to hip flexion. Oddsson postulated that the required backward excursion of the COM could be accomplished in two ways, through the knee flexion (which could alternatively be produced through ankle extension). Following this preparatory movement, hip flexion is accompanied by hyperextension of the knee joint and rotation about the ankle joint. Oddsson further explains that these coordinated movement patterns of different limbs are most likely controlled through commands from the CNS.

Body segment movements are achieved through accurate activation (typically, reciprocal activation and co-contraction) and inhibition of muscles about a joint and are known to be mitigated by commands from higher order

centres in the CNS (Kandel et al., 1991). Reciprocal activation is the temporal propagation of muscle activations in association with the speed, duration and movement phase, and co-contraction is the simultaneous activation of antagonistic muscle groups. Both of these methods of joint movement control have their benefits and pitfalls, for example, reciprocal activation is known to be more efficient with the use of energy, but is dependant on precise knowledge of the load to be experienced by the joint, while, on the other hand, co-contraction is not as efficient but it does not require the same degree of load knowledge (Kandel et al., 1991). Oddsson demonstrated that during hip flexion experiments, both methods of control are used by muscles about the ankle joints suggesting that these muscles were both primary movers and postural stabilizers.

In two related articles completed by Stapley and colleagues (P. Stapley et al., 2000; P. J. Stapley et al., 1999) experimental results were reported describing the movement pattern of the body segments during whole-body reaching. Whole-body reaching involves forward flexion and arm extension to pick up a dowel when starting from an initial standing position with arms at the sides. Stapley and colleagues investigated whether it was possible to complete whole-body reaches without large movements of the COM, and if so, how this was accomplished. It was proposed that perhaps body segment movement patterns may demonstrate a similarity to the 'axial synergies' seen during forward leans (Crenna et al., 1987). For example, when reaching forward, the upper

body segment would move forward while the pelvis and proximal lower extremity segments would move backwards resulting in little displacement of the COG (Crenna et al., 1987). Contrary to their predictions, although there was some axial synergy for whole-body reaching, the trunk segment and the COG were not controlled within some optimum position to be used as a postural reference point (P. J. Stapley et al., 1999). Although the hips were found to move in opposition to the head and trunk, this movement decreased with greater reach distances. The authors suggested several possible reasons for these unexpected results. First, increasing the movement of the trunk segment position would allow the COG to move into a position that maximized the amount of anterior movement available therefore, decreasing the chances of falling backwards if suddenly perturbed during the reach. A second hypothesis (and that favored by Stapley and colleagues) stated that by decreasing hip movement and moving the COG forward, a flexor torque is generated which moves the body under the influence of passive gravitational forces assisting in forward rotation and completion of the task. This second hypothesis was supported by later work completed by Stapley et al. (P. Stapley et al., 2000); computer simulations of whole-body reaching demonstrated that the movement could be completed without substantial COM displacement. Therefore, because whole-body reaches are completed with COM displacement, body segment movements must be organized in preference of generating the focal movement over postural stability.

The results of the work completed by Stapley and colleagues have been furthered by experiments conducted by Kaminski and Simpkins (Kaminski & Simpkins, 2001). Trials which involve manipulation of the BOS and the distance of the dowel lift during a whole-body reach were conducted to test the hypothesis that APAs generated during whole-body reaching help to accelerate the body forward into a region where passive gravitational forces can help to pull the body forward. Movement of the COM increased with increases in reach distance, yet the COP movement was much greater than needed solely to stabilize against the perturbation of a whole-body reach, therefore supporting their tested hypothesis.

1.5 Biomechanical Considerations for Weight Acceptance

In 1997, Toussaint and colleagues (Toussaint, Commissaris, Hoozemans et al., 1997) reported a study that examined the movement of the COP and COM during forward whole-body reaching with weight acceptance. Several different variables were examined including the whole-body angular momentum, the external moment generated by the ankle and also the COP and COM positions. Subjects completed the forward reach several times before being cued to pick up the load, thus making the weight acceptance part of a continuous fluid movement. Just prior to weight acceptance the COP was positioned anterior to the COM, therefore the external moment generated by the COP was greater than the flexor moment generated by the COM causing an external extensor angular momentum of the whole body about the ankle. In addition, there was a posterior movement of the COM to further increase the posterior-directed angular

momentum. At weight acceptance the COM rapidly moved forward with the addition of the extra mass. Further anterior movement of the COP and resulting increases in the external extensor whole-body moment followed. Therefore, there was a clear presence of an APA initiated by the body to control the perturbation to the COM caused by picking up a load. The results suggest that while the reaching task demands postural adjustments to control the movement of the body during descent.

Toussaint, Commissaris and Beek continued this work by examining the particular APA created for two different load lifting techniques: a leg lift and a back lift (Toussaint, Commissaris, & Beek, 1997). The investigators found that a different APA was seen for the two techniques; prior to weight acceptance in the leg lift there was posterior movement of the COP and a resultant flexor torque of the body that was not seen in the back lift. It was postulated that this short negative momentum generated by the movement of the COP was produced because the initial forward position of the COM (specific to the leg lift) and the constraints of anterior foot length limit the forward movement of the COP with weight acceptance. Accordingly, the peak anterior position of the COP is delayed until the moment when a positive momentum (towards upright stance) is required to lift the weight.

Complementary to these studies (Toussaint, Commissaris, & Beek, 1997; Toussaint, Commissaris, Hoozemans et al., 1997) research has been conducted on the effect of additional weight to an individual's postural stability (Holbein &

Redfern, 1997; T. H. Lee & Lee, 2003). It was found that loads held at a higher height caused greater instability and demanded greater whole-body moments to counteract forces resulting from the load (T. H. Lee & Lee, 2003). Furthermore, greater instability was seen when the weight was held further from the body's COG and sway-based hip strategies were adopted to maintain balance with loads held on the shoulders instead of at the subject's sides (Holbein & Redfern, 1997).

1.6 Postural Stability and Aging

In section 1.1 (Falls and Aging), factors were outlined that increase the elderly individuals risk for falling. Likewise, balance deficits and postural instability can be attributed to many of these same factors. Elderly individuals have been shown to have increased postural sway during quiet stance compared to young individuals suggesting a decrease in postural control (Tanaka, Takeda, Izumi, Ino, & Ifukube, 1997). This is even more evident in situations of decrease feedback, as in standing on a piece of foam or with the eyes closed (Choy, Brauer, & Nitz, 2003).

Decreases in postural stability are frequently tested with functional balance tests, for example, functional reach. Functional reach is the distance beyond arms-reach that an individual can reach forward without lifting the heels off the ground (Duncan, Weiner, Chandler, & Studenski, 1990). Typically, functional reach declines with age and has been shown to be an appropriate indicator of frailty (Weiner, Duncan, Chandler, & Studenski, 1992). As well,

functional reach is associated with increases in other instability measures such as tandem stance and sit-to-stand times (Duncan, Studenski, Chandler, & Prescott, 1992; Weiner et al., 1992). Concurrently, another related measure of instability, functional BOS, is also influenced by age. An individual's functional BOS is the proportion of the individual's BOS that is used in maximal leans in the forward, backward and lateral directions (King, Judge, & Wolfson, 1994). Ultimately, the functional BOS is influenced by the strength about the ankles and the ability to move the COP as far forward and back as possible. Studies have shown that with age and a related decrease in ankle strength there is a decrease in functional BOS size (King et al., 1994).

This decrease in functional BOS size significantly influences the method of postural control employed by elderly adults during voluntary movements and in response to perturbations. For example, during quiet stance, elderly individuals were seen to rely more on the posterior or heel area of their foot for support than younger adults (Tanaka et al., 1997). Alternatively, Stelmach et al. (Stelmach et al., 1989) analyzed reflexes associated with postural control in various age groups and found that although subjects aged 60 years and older had the same balance strategies as young adults, with respect to muscle activation patterns, these responses were less rapid. Furthermore, elderly subjects demonstrated greater variability between trials suggesting that their motor control was not as precise as that seen in younger subjects. Desynchronization of the reflexes seen in the two lower extremities would also suggest that limitations are not solely in

reduced strength and reaction time abilities, but also in impairments in higher-level motor control systems (Stelmach et al., 1989). These limitations have implications in the spectrum of activities that the elderly can engage in. Motions that cause greater instability could cause greater dependence on these age-impaired reflex systems and an increased likelihood of loss of balance episodes (Stelmach et al., 1989). Moreover, examination of balance in elderly subjects compared to younger subjects has demonstrated increases in COP movement and velocity and other COP-related indicators of instability (King et al., 1994; Mecagni, Smith, Roberts, & O'Sullivan, 2000; Okuzumi et al., 1995; Tanaka et al., 1997).

Proprioception also plays a key role in postural stability, particularly in elderly individuals. Typically, study results suggest that postural stability is not directly influenced by proprioceptive information. Evidence for this comes from experiments in which subjects' balance was tested when a sphygmomanometer cuff was placed around the ankle. A sphygmomanometer cuts off the passage of the necessary mechanoreceptor information to the spinal cord to prevent proprioceptive input from the feet. It was found that the reduction in proprioceptive information had little effect on the balance of the individual when standing (Diener, Dichgans, Guschlbauer, & Mau, 1984). However, proprioceptive inputs can have an indirect effect through influences on motoneuron firing and motoneuron pool excitability (Aniss, Diener, Hore, Gandevia, & Burke, 1990). Furthermore, during moments of greater instability,

proprioceptive information becomes more important (Diener et al., 1984). The information is provided to central command centres of the brain to influence the muscular control of balance via mechanoreceptors in the skin of the foot and joints, and through muscle receptors (including spindle fibers and Golgi Tendon Organs). From this information, appropriate control of stretch reflexes and muscle activation responses can assist in development of appropriate movement about the joint to maintain stability during voluntary movements (Aniss et al., 1990). Unfortunately, age-related diseases like diabetes result in degradation of proprioceptive sensors and other neuromuscular functions. This loss of proprioceptive feedback has been associated with an increased risk of postural instability in these elderly adults (Richardson, Ching, & Hurvitz, 1992).

1.7 Specific Aims

Despite the extensive research completed so far, there has been little investigation of the joint torque demands associated with whole-body postural stability when accepting a weight to the outstretched hands. Nor have the preparatory movements and efforts associated with such a task been studied in depth. Furthermore, little work has been done with elderly subjects completing these specific tasks, or of their ability to prepare for the upcoming perturbation by generating the appropriate anticipatory postural movements. The general aim of this thesis is to understand the postural control strategies required for an individual to successfully accept a weight in their outstretched hands.

The main goal of the study outlined in Chapter 2 was to extend the understanding of movement preparation effects on a person's ability to minimize COG displacement while accepting a weight in the outstretched hands. Then in Chapter 3 I attempted to gain a better understanding of the mechanics and abilities of senior women to prepare for an impending perturbation. These goals were addressed through a series of weight acceptance tasks performed by young and elderly subjects, where weight acceptance occurred by lifting or 'catching' an object, suspended from a rope, when the release is self-initiated or investigator-initiated.

CHAPTER 2: EFFECTS OF ANTICIPATION ON ABILITY TO MAINTAIN POSTURAL STABILITY WHILE ACCEPTING A WEIGHT IN THE OUTSTRETCHED HANDS.

2.1 Abstract

Two strategies are available for minimizing the displacement of the whole-body centre-of-gravity (COG) when accepting the weight of an object in the outstretched hands, anticipatory postural responses and corrective postural responses. Adequate preparation is needed to produce appropriate anticipatory responses, and if these preparations are incorrect, then the individual must rely on corrective responses. The goal of this study was to compare the efficacy of these two techniques for maintaining the COG's position during weight acceptance. In particular, we tested the following hypotheses: 1) use of anticipatory responses would cause decreases in COG displacement compared to corrective responses, 2) these decreases in COG displacement with anticipatory responses would further cause decreases in COP displacement and joint torques, 3) changes in the horizontal distance and vertical height of the object to be accepted will influence COG, COP displacement and joint torque increases. Fifteen young women [mean age: 23+/-2.73yrs] participated in the study. Trials began with subjects reaching forward to wrap their hands around the wooden dowel handlebar of an object suspended by a rope. Subjects accepted the weight of the object in one of three ways; (1) self-initiated

(voluntary) lifting (VL), (2) self-initiated (voluntary) release of tension in the suspending rope (VR) or (3) investigator-initiated (involuntary) release. COG increased between VL and VR but not between VR and IR trials. Furthermore, COP displacement increased between VL and VR trials but not between VR and IR trials. This indicated that subjects were as able to minimize COG displacement in the IR trials as they were in VR trials. However, disruption to the COG is minimized the most in the VL condition, which allows the individual to control both the rate and onset of weight acceptance and to reduce joint torques demands.

2.2 Introduction

When accepting the weight of an object into the hands, as in lifting or catching, precise postural control is required to maintain the centre-of-gravity (COG) within the base of support (BOS). This can be achieved through anticipatory postural adjustments (APAs) or corrective postural responses. While corrective responses are based primarily on proprioceptive feedback, APAs have been shown to depend on the individual's ability to obtain knowledge and control the perturbation onset, duration and direction (Paulignan, Dufosse, Hugon, & Massion, 1989; Struppler, Gerilovsky, & Jakob, 1993). For example, when using precision grasping to catch an object between the index finger and thumb, the duration of muscle activation (used to limit movement about the elbow joint due to the added object force) was based on the expected, not the actual, weight of the object (Johansson & Westling, 1988a, 1988b; Westwood, Dubrowski,

Carnahan, & Roy, 2000). If the weight of the object was increased without the subject's knowledge then muscle activation was insufficient for minimizing elbow joint movement. Similarly, whole-body moments and L5-S1 torques were too small and short to effectively lift a dowel off the ground when subjects used whole-body back or leg lifts to pick up an object that was thought to be a lighter weight (Commissaris & Toussaint, 1997b; Toussaint, Michies, Faber, Commissaris, & van Dieen, 1998). Furthermore, when subjects self-initiated the release of a weight from the left into the right hand, sufficient (and early) muscle activation was seen in the arm to minimize elbow joint rotation (Shiratori & Latash, 2001). Conversely, in trials where an unseen investigator induced the release of the weight no early activation of arm muscles occurred (Johansson & Westling, 1988a, 1988b; Paulignan et al., 1989; Struppler et al., 1993; Westwood et al., 2000). Presently, little research has been done to understand the specific contributions of subject control over perturbation rate and onset in maintaining postural stability.

Theoretically, if a weight is accepted into the outstretched hands and moves the whole-body COG forward, then the COP would have to move even farther anterior to control COG from moving outside the BOS (D. A. Winter et al., 2001). Therefore, if the COG was placed closer to the boundaries of the BOS then the possible anterior movement of the COP would be reduced. Postulations could therefore be made that the initial COG position within the BOS could affect the preparation and execution of postural control strategies.

The aim of this study was to extend our understanding of the effects of movement preparation on ability to maintain postural stability while accepting a weight in the outstretched hands. In particular I aimed to test the hypotheses that 1) use of anticipatory responses would cause decreases in COG displacement compared to corrective responses, 2) decreases in COG displacement due to anticipatory responses would cause decreases in COP displacement and joint torques, 3) changes in the horizontal distance and vertical height of the object to be accepted will influence COG, COP displacement and joint torque increases.

2.3 Methods

2.3.1 Subject Recruitment

Subjects consisted of 15 young women (mean \pm -SD: age=23 \pm -2.73yrs, height=1.67 \pm -0.044m, body mass=57.81 \pm -8.61kg) recruited primarily through postings of notices at Simon Fraser University. Subjects were interviewed initially by phone to see if they passed the exclusion criteria. Subjects were excluded if they had a Folstein Mini-Mental State Examination (FMMSE) score less than 24, had uncorrected visual problems, or had a history of severe orthopedic or neuromuscular problems. All participants signed a written consent form (Appendix A) and the experiment was approved by the Research Ethics Committee at Simon Fraser University.

2.3.2 Weight Acceptance Reaching Protocol

Each subject was required to complete a series of weight acceptance tasks (Figure 2.1). The subject began each trial in a position of impending weight

acceptance, by reaching forward to wrap her hands lightly around the handlebars of the object to be accepted. The object was suspended from an electromagnet by way of a length-adjustable rope. During trials the subject was required to accept the weight of an object into her outstretched hands in three different ways, subject-induced (voluntary) lifting (VL), subject-induced (voluntary) release of tension in the suspending rope (VR) or investigator-induced (involuntary) release of tension in the suspending rope (IR), each presented in a random order. During VL trials the subject was instructed to vertically lift the object just enough to accept its full weight and to hold that position for 2 seconds. In VR trials the subject was instructed to “catch” the object after she self-initiated release of the rope by pressing two release buttons located on the object handlebars which discharged the electromagnet (release time was approximately 50ms). During IR trials the subject was instructed to “catch” the object after the investigator discharged the electromagnet following a random time delay (between 0 and 4s in duration). In both VR and IR trials, the subject was instructed to maintain her original position, and allow as little movement as possible of the object following weight acceptance. The subject was asked to bend only at the hips and ankles (keeping the knees straight) but was permitted to raise her heels off the ground. Trials were rejected if the object moved after weight acceptance more than 2 cm from its original position (in any direction).

During the trials, I also varied the horizontal distance from the ankle to the object (near = 45% body height, far = 50% body height) and the height of the

object (low = 60% body height, medium = 80% body height, high = 100% body height). In total, the subject was asked to perform 18 different combinations of weight acceptance condition, horizontal distance and height (study design outlined in Figure 2.2) in a random order. Each combination of location and weight acceptance condition was repeated 3 times (consecutively) for a total of 54 trials.

Pilot studies demonstrated that the subject had a tendency to pull the object towards herself during weight acceptance in order to complete the task with minimal effort. Consequently, I used a program (Biofeedtrak, Motion Analysis Inc., Santa Rosa, CA, USA) to alert the subject, with a buzzing sound, if she moved the object more than 2 cm in any direction from its original position. If the buzzer sounded, the trial was discarded and repeated.

In all trials, the subject grasped the handlebars with the tip of each index finger resting on top of two release buttons which were 8cm from the centre of the object. The object to be lifted consisted of two load cells (capture rate = 960Hz, models 31 and 41, Sensotec Inc., Columbus, OH, USA) which sandwiched a wooden cylindrical (diameter: 3cm) handlebar (Total mass=1.89Kg, Figure 2.3.a). Hand force was determined by subtracting the tension in the upper load cell from that in the lower load cell using the following equations,

$$F_{hand} = F_{LCb} + F_{LCt} \quad \text{Eq.1}$$

$$F_{LCb} = m_{LCb}(g - a_z) \quad \text{Eq. 2}$$

where, F_{hand} is the force at the hand, F_{LCt} is the force registered by the upper load cell and the tension in the suspending rope, m_{LCb} is the mass of the lower load cell and a_z is the acceleration of object (see also Figure 2.3.b,c). Trials were repeated if the load cells indicated that the subject was pushing down or pulling up on the handlebar prior to the 'go' cue (in VL trials) or release of the electromagnet (in VR and IR trials).

In order to standardize the COP position at the start of the trial, a computer screen was projected in front of the subject and provided visual feedback of the horizontal position of the COP. The subject was asked to maintain her COP close to a target value (method of calculation, Appendix B) shown on the screen until weight acceptance occurred. After weight acceptance, the subject no longer had to monitor her COP position.

In each trial, I used an 8-camera motion measurement system (Eagle model cameras with EVaRT software, Motion Analysis Corp., Santa Rosa, CA) to measure the three-dimensional positions (at 120Hz) of reflective markers placed bilaterally on the acromion processes, the lateral humeral epicondyles, the lateral ulnar condyles, the right and left ASIS, the greater trochanters, the lateral femoral epicondyles, the lateral malleoli and the fifth metatarsals. Markers were also placed at the top of the head, on the sacrum, and on the ends of the handlebars.

I also collected surface electromyography, sampling at 960Hz (Myosystem 1400, Noraxon USA, Inc., Scottsdale, AZ) from 5 muscles on the right side of the body: the tibialis anterior (TA), the medial gastrocnemius, the erector spinae

(ES), the anterior deltoid and the trapezius. EMG traces were filtered to remove motion artifacts using a 4th order Butterworth high-pass filter having a cutoff frequency of 40Hz. The EMG signal was then rectified and filtered again to determine the envelope of signal intensity using a 4th order Butterworth low-pass filter with a cutoff frequency of 10Hz.

2.3.3 Data Analysis

I used a sagittal plane linked-segment model (Figure 2.4) to estimate the position of the whole-body COG displacement in the anterior-posterior direction, and to estimate joint torques at the hip, ankle and shoulder. The model consisted of six body segments (including torso/head/neck, upper arms, forearms, thighs, shanks and feet) and included the effect of inertial forces in solving the equations of motion through inverse dynamics. Link segments had centre of mass positions and lengths as described by Dempster's anthropometric data (D.A. Winter, 1990). I used customized MATLAB programs (The MathWorks, Natick, MA, USA) to calculate the dependant variables (as described below) from the motion and force plate data. By inputting motion capture data into the model, I estimated the temporal variations in the horizontal (anterior-posterior) position of each segment COG position and corresponding changes in the whole-body COG. COG calculations accounted for the force acting on the hand (Figure 2.3.b, c and Figure 2.4) using the following equation,

$$COG = \frac{[\sum (COM_i * m_i) + F_{hand} / g * (L_{arm} + L_{forearm})]}{[BW + F_{hand} / g]} \quad \text{Eq. 3}$$

where COM_i is the COM of i body segments, m_i is the mass of i body segments, F_{hand} is the force acting on the hand, BW is body weight, and L_{arm} and $L_{forearm}$ are the lengths of the upper and lower arms respectively. I considered not including the forces acting on the hand in the COG calculations, similar to those seen in Commissaris et al. (Commissaris, Toussaint, & Hirschfeld, 2001), but opted to include the weight so that I could relate the COG position more clearly with movements of the COP. Figure 2.5 indicates that without the additional weight added, the COP does not oscillate about the COG as would be expected in a stable upright system.

I determined the anterior-posterior COP position from force plate (Model 6090H, Bertec Corp., Columbus, OH) traces with the following equation: $COP = M_y / F_z$ (where M_y is the moment measured by the force plate about the medial-lateral axis and F_z is the force in the vertical direction, Figure 2.4 inset).

2.3.4 *Dependant variables*

Dependant variables included changes with weight acceptance in hip torque (T_{hip} , ΔT_{hip}), ankle torque (T_{ankle} , ΔT_{ankle}), shoulder torque ($T_{shoulder}$, $\Delta T_{shoulder}$), COP displacement and COG displacement in the horizontal plane. Changes were calculated as the difference between peak and baseline values. Baseline values of dependant variables were equal to the mean value seen prior to weight acceptance (duration: 0.5s). Peak values were equal to the maximum value that occurred within the weight acceptance interval which I defined as the

1.5s interval starting at the instant when the force acting on the hand first equals (or exceeds) the weight of the object (Figure 2.6).

For hypotheses testing, I used a three-factor repeated measures multivariate analysis of variance (MANOVA, $\alpha=0.05$) to identify if there were any significant main effects (or two-way interactions) for the weight acceptance condition (3 levels: VL, VR or IR), object horizontal distance (2 levels: near or far), and height (3 levels: high, medium or low). These were followed by post hoc pairwise comparisons using a total alpha level of 0.05 and Bonferroni's correction. All statistical tests were carried out using SPSS statistical analysis software (v.12.0 for windows, SPSS Inc., Chicago, IL, USA). Due to equipment malfunction or post experimental analysis which indicated that the subject had been applying pressure to the object prior to cued weight acceptance, some trials were eliminated. As a consequence, many data sets were incomplete and MANOVA tests were only carried out on the 8 subject sets that were complete (see Table 2.1). Unless otherwise stated, reported results are means +/- 1 SD.

2.3.5 Supplementary Analysis

I conducted a variety of supplementary analyses to gain further insight into how stability was achieved in each position. Angular changes at individual joints ($\Delta\theta_{hip}$, $\Delta\theta_{ankle}$, $\Delta\theta_{shoulder}$) and total body angle changes (sum of $\Delta\theta_{hip}$, $\Delta\theta_{ankle}$, $\Delta\theta_{shoulder}$) were calculated and compared between trials using a three-factor (weight acceptance condition, horizontal distance, and height) repeated measures ANOVA.

Additionally, I created composite traces by averaging data from each trial with all 15 subjects after synchronizing to the instant of weight acceptance, for all dependant variables. This analysis was done only for trials conducted with the object in the near-medium position.

As a measure of response time, I calculated the time interval between the onset of COG anterior movement and COP anterior movement. Using one-way repeated measures ANOVA I compared response times between the different weight acceptance conditions across all trial positions. The initiation of the anterior movement of the COG and COP were calculated as the first point in time when the trace rose 2 standard deviations above its mean initial values. In addition, I examined response times from composite traces of COG and COP horizontal movement in each condition.

I used Pearson's correlation coefficients to determine associations between response times, COP, COG and object height displacement during weight acceptance. Unless otherwise stated, reported p values are for r not r^2 .

2.4 Results

2.4.1 Hypotheses results

Anterior movement of the COP and the COG, as well as increases in joint torques and changes in angles at the ankle, hip and shoulder accompanied increases in the force experienced by the hand in all three weight acceptance conditions (Figure 2.7). Hip flexion, ankle dorsiflexion and shoulder flexion were seen to increase with weight acceptance in VL trials, and hip flexion, ankle

dorsiflexion and shoulder extension were seen to increase with weight acceptance during VR and IR trials. Also, there was a much greater rate of weight acceptance in VR trials than in VL trials. This increase in rate of weight acceptance appears to cause increases in the rates of change of COP and COG displacement, joint torques and angles.

There was a significant main effect of weight acceptance condition on COG displacement ($F_{0.05, 2, 14} = 24.120$, $p < 0.001$, Figure 2.8.a). VL trials had significantly less COG displacement from baseline than VR trials (1.74 \pm 0.51cm vs. 2.28 \pm 0.70cm, 95%CI diff= 0.35 to 0.74cm, $p = 0.001$) and IR trials (1.74 \pm 0.51cm vs. 2.27 \pm 0.48cm, 95%CI diff= 0.35 to 0.73cm, $p = 0.001$). On the other hand, no significant differences were seen in the COG displacement between VR and IR trials (2.28 \pm 0.70cm vs. 2.27 \pm 0.48cm, 95%CI diff = 0.24 to 0.26cm, $p = 1.000$).

There was a significant main effect of weight acceptance condition on COP displacement ($F_{0.05, 2, 14} = 31.336$, $p < 0.001$, Figure 2.8.a). Significantly more displacement of the COP was seen in the VR than in the VL conditions (3.13 \pm 0.98cm vs. 2.28 \pm 0.66cm, 95%CI diff=0.39 to 1.30cm, $p = 0.010$). Furthermore, significantly more COP displacement was seen in IR than in VL trials (3.67 \pm 0.62cm vs. 2.28 \pm 0.66cm, 95%CI diff=0.05 to 1.05cm, $p < 0.001$) but again no significant difference was seen between the VR and IR conditions (3.13 \pm 0.98cm vs. 3.67 \pm 0.62cm, 95%CI diff=-0.12 to 1.21cm, $p = 0.110$).

There was a significant main effect of weight acceptance condition on the increases in hip ($F_{0.05, 2, 14} = 36.691$, $p < 0.001$), ankle ($F_{0.05, 2, 14} = 30.920$, $p < 0.001$) and shoulder ($F_{0.05, 2, 14} = 8.640$, $p = 0.021$) torque (Figure 2.8.b). There was a significantly greater increase in hip torque in IR trials than in VR trials (48.03 \pm 11.92Nm vs. 32.03 \pm 7.82Nm, 95%CI diff=7.37 to 24.62Nm, $p = 0.010$) and in VR than in VL trials (32.03 \pm 7.82Nm vs. 21.58 \pm 6.35Nm, 95%CI diff=6.02 to 14.89Nm, $p = 0.003$). Moreover, there was significantly greater increase in ankle torque in VR than in VL trials (21.10 \pm 5.03Nm vs. 15.13 \pm 3.38Nm, 95%CI diff=3.59 to 8.36Nm, $p = 0.001$) and IR than in VL (25.76 \pm 4.63Nm vs. 15.13 \pm 3.38Nm, 95%CI diff=6.72 to 14.55Nm, $p < 0.001$) but not between VR and IR (21.10 \pm 5.03Nm vs. 25.76 \pm 4.63Nm, 95%CI diff=-0.70 to 10.00Nm, $p = 0.089$). Finally, increases in shoulder torque were greater in VR than in VL trials (15.94 \pm 3.81Nm vs. 12.67 \pm 3.40Nm, 95%CI diff=1.98 to 4.56Nm, $p = 0.002$) and IR than in VL trials (24.20 \pm 10.08 vs. 12.67 \pm 3.40Nm, 95%CI diff=0.11 to 22.95Nm, $p = 0.048$) but again, not between VR and IR (15.94 \pm 3.81Nm vs. 24.20 \pm 10.08Nm, 95%CI diff=-2.06 to 18.58Nm, $p = 0.122$).

There was a significant main effect of object height on COG displacement ($F_{0.05, 2, 14} = 5.986$, $p = 0.013$, Figure 2.9.a). In particular there was significantly greater COG displacement in medium than in low height trials (2.27 \pm 0.68cm vs. 1.96 \pm 0.44cm, 95%CI diff=0.09 to 0.53cm, $p = 0.040$). A significant main effect of object height was also found for COP displacement ($F_{0.05, 2, 14} = 6.166$, $p = 0.028$, Figure 2.9.a) however, pairwise comparisons with Bonferroni's corrections

indicated no significant differences between low, medium and high object heights (3.07 \pm 0.96cm vs. 3.22 \pm 0.90cm vs. 2.79 \pm 0.98cm, low medium: 95%CI diff=-0.07 to 0.35cm, p=0.222, medium high: 95%CI diff=-0.89 to 0.04cm, p=0.073).

There was a significant effect of object height on the increases in ankle torque ($F_{0.05, 2, 14} = 5.238$, $p=0.028$, Figure 2.9.b) however, again, pairwise comparisons with Bonferroni's correction indicated no significant differences between the low, medium and high object heights (21.15 \pm 6.22Nm vs. 21.74 \pm 6.02Nm vs. 19.10 \pm 6.12Nm, low medium: 95%CI diff=-1.32 to 2.49Nm, $p=1.000$, medium high: 95%CI diff=-0.74 to 6.02Nm, $p=0.134$). There was no significant effect of object height on increases in hip ($F_{0.05, 2, 14} = 1.907$, $p=0.203$) or shoulder torques ($F_{0.05, 2, 14} = 4.140$, $p=0.063$, Figure 2.9.b).

Trials which increased the horizontal distance between the ankles and the outstretched hands resulted in varying effects on COG, COP and joint torque increases (Figure 2.10.a,b). There was greater COG displacement during far than near trials (2.20 \pm 0.69cm vs. 2.00 \pm 0.54cm, 95%CI diff=0.05 to 0.35cm, $F_{0.05, 1, 7} = 9.443$, $p=0.018$), but no significant difference in COP displacement ($F_{0.05, 1, 7} = 0.290$, $p=0.607$, Figure 2.10.a). In addition, there was significantly greater changes in hip torque during far than near trials ($F_{0.05, 1, 7} = 7.096$, $p=0.032$), but no significant differences in ankle torque ($F_{0.05, 1, 7} = 0.001$, $p=0.997$) or shoulder torque ($F_{0.05, 1, 7} = 0.012$, $p=0.915$, Figure 2.10.b).

There were no significant two-way interaction effects (see Table 2.2) between the object height, horizontal distance and the weight acceptance condition.

I found that during VL trials the TA and medial gastrocnemius muscles experienced little or no change in activation (Figure 2.11). During VR trials, small spikes in activity occurred in both the TA and medial gastrocnemius at approximately 200ms after weight release synchronous with the peak in COP displacement. During IR trials, I again saw spikes in the activity of the TA and medial gastrocnemius muscles. However, these spikes were significantly greater than those experienced during VR trials. I found that prior to weight acceptance there appeared to be no change in activity of the TA or gastrocnemius muscles in any of the different weight acceptance conditions. I also found no appearance of a triphasic muscle activation pattern in the antagonist/agonist grouping of the TA and the medial gastrocnemius.

2.4.2 Supplementary results

There was a significant main effect of weight acceptance condition (Figure 2.12) on total angle change ($F_{0.05,2,14}=29.745$, $p<0.001$). Significantly greater increases in total angle occurred in IR than in VR trials ($11.80\pm 7.16\text{deg}$ vs. $5.69\pm 4.07\text{deg}$, 95%CI diff=1.82 to 10.40deg, $p=0.009$) and greater increases in total angle occurred in VR than in VL trials ($5.69\pm 4.07\text{deg}$ vs. 3.42 ± 3.93 , 95%CI diff=0.18 to 4.36deg, $p=0.034$).

The weight acceptance condition also had a significant main effect on response time ($F_{0.05, 2, 97} = 4.331$, $p = 0.016$, Figure 2.13). On average (based on composites trace behaviour) the COP began to increase in unison (i.e. with no detectable delay) with the onset of COG movement in VL trials. However, in VR trials the average time delay was 33ms, and in IR trials it was 177ms.

Finally, results indicated a significant minor association between COP displacement and object height deflection ($r^2 = -0.054$, $p > 0.001$) and also between COG displacement and object height deflection ($r^2 = -0.06$, $p < 0.001$). No significant association was found between COP and COG displacement and response times.

2.5 Discussion

I found that, for the specific weight acceptance task that I examined, corrective (or balance recovery) responses (as relied upon in VR and IR trials) were not as effective as APAs (as relied upon in VL trials) in stabilizing the whole-body COG. In particular, APA required control over both the rate and onset of the weight acceptance. This supports the notion that involuntary perturbations lead to increased risk for injury during lifting. One mechanism by which anticipatory adjustments minimize COG displacement is by reducing the time delay between the onset of COP movement with respect to COG displacement. An additional protective mechanism during voluntary lifting is the slower rate at which the hand force is increased. Together, these two

mechanisms contribute fundamentally to our ability to lift weights without losing balance, and with reduced musculoskeletal effort.

My results provide insight on the relative importance to postural control of the ability to control the rate versus the onset of weight acceptance during lifting or catching. In VL trials, subjects were able to control both the onset and the rate of weight acceptance. In VR trials, subjects were able to control the onset but not the rate of weight acceptance. Finally, in IR trials the subjects were unable to control both the onset and rate of weight acceptance. Since COG and COP displacement were not significantly different in VR and IR trials, it appears that the ability to control the onset of weight acceptance had little effect on the ability to minimize COG disruption. However, when the subject had limited ability to control the rate of weight acceptance, there was a decreased ability to minimize COG displacement. Moreover, decreases in torque demands during voluntary lifting compared to weight release trials, were primarily due to the subject's ability to control and reduce the rate of weight acceptance in the VL condition.

Previous studies indicate that minimizing the position change of the upper limbs during weight acceptance is significantly more successful when the subject initiates the onset of the perturbation (Johansson & Westling, 1988a, 1988b; Westwood et al., 2000). During subject-initiation, the subject's system is able to produce APAs, of which the magnitude and timing is appropriate for opposing the perturbation. The present study shows that this is also true for more complicated whole-body disturbances. In particular, the current research

indicates that the ability to control (and correctly predict) both the onset and the rate of weight acceptance is key to effective APA development and postural control (Horak et al., 1984; Shiratori & Latash, 2001; Toussaint, Commissaris, & Beek, 1997; Toussaint, Commissaris, Hoozemans et al., 1997).

I also found delays in the onset of COP displacement during VR (33ms) and IR (177ms) trials. Hence it appears that when control of the rate and onset of weight acceptance was lost (as in VR and IR trials, respectively) corrective responses are initiated. However, these corrective responses were initiated earlier when the subject had control over the onset of the perturbation (as in VR trials) than when they did not. Delays in the initiation of corrective responses would allow greater acceleration of the whole-body COG thus demanding increased torque development and related increases in COP control.

Analysis of the composite traces indicated that there was a large increase in the rate of weight acceptance from VL to VR trials. This increase in rate meant that the force (and subsequently the torque) at the hand increased more quickly, therefore allowing the perturbation to have a greater effect on the body before postural responses were initiated. Consequently, greater torques were required at the joints during the postural response.

My results also show that the ability to control the COG within the BOS during weight acceptance depends on the height and horizontal distance of the object from the subject's ankles. Increasing the horizontal distance of the object caused the baseline position of the COG to move closer to the boundaries of the

BOS. Furthermore, positions that required increased hip flexion (as when the object was placed in low and high heights) also moved the baseline position of the COG closer to the boundaries of the BOS.

In particular, I found that with increases in the horizontal distance of the object there were increases in the COG displacement and change in hip torques, but no increases in the COP displacement or change in ankle torque. During far object placement the subject had to flex at her hips increasing the moment arm of the torque generated by the weight of the object and therefore increasing the joint torques demands to oppose this larger perturbation. Conversely, the baseline COG position was closer to the boundaries of the BOS, leaving little room for COP corrections, so subjects may have employed an alternative strategy (similar to those seen in older adults in previous studies (Benjuya, Melzer, & Kaplanski, 2004; Ho & Bendrup, 2002)) which caused a reduction in the torque generated at the ankle. It could also be that there was a greater preference to minimize hip flexion rather than ankle dorsiflexion and therefore increases were seen in the hip extensor torques much more than the ankle plantarflexor torques. Further testing on muscle activation and joint torque propagation patterns would be needed to understand the strategy employed during the different object positions.

Finally, I found that subjects experienced the greatest amount of COG displacement when the object was at a medium height. During low and high height trials, the subject was required to flex at the waist to reach the object,

bringing the baseline COG position closer to the boundaries of the BOS. Thus, variations in the postural strategy would be needed to minimize ankle torque. However, increases in peak hip and shoulder torques were not seen suggesting that a modified postural strategy was used. This modified or alternative strategy may have been used when the COG was under a threshold distance from the boundaries of the BOS. As I used a task that allowed for multiple degrees of freedom, there is opportunity for various combinations of body segment rotations that could dictate the movement strategy used by different subjects and across different conditions and object positions.

There are several important limitations to this research. First, results of this study with healthy young women may have limited applicability to males or elderly populations.

Second, I could not create a situation where subjects were completely unaware of the impending weight acceptance. However, past studies show that even if the subject is expecting a perturbation to the upper extremity, they cannot prepare for it properly if the perturbation onset is not initiated by the subject (Johansson & Westling, 1988a, 1988b; Westwood et al., 2000). Precise timing of the postural responses is required in order to minimize the effect of the additional weight. Yet, this precise timing of postural responses can only be produced if the onset and duration of the perturbation is known. In this study, random time delays between the start of the trial and the moment when the investigator initiated weight release prevented subject knowledge of perturbation onset.

Third, I focused on only the weight acceptance and not the transport portion of lifting or catching. That is, subjects began trials with their hands outstretched in a position of impending weight acceptance. Thus my analysis may not apply for highly ballistic lifts or catches, where the body segments have high velocity prior to weight acceptance. However, my design helped ensure consistent positioning of COP, COG, and body segments between trials. Future studies may help expand our analysis by including the transport portion of lifting and catching.

Fourth, all subjects were instructed to monitor their COP and match it with a target COP at the start of the trial. As a consequence, the initial torques may have been greater in some subjects than others due to anthropometric differences. However, I felt the comparisons were more relevant if subjects began trials in the same body configuration despite the weight acceptance condition.

Finally, during all trials I instructed the subject to minimize body movement during weight acceptance, and clearly this is often not the priority during daily weight acceptance. However, this provided the subject with a well-defined goal, and a means to evaluate task performance across different conditions. For example, it would not have been meaningful to compare VL trials where the subject lifted the object 4 cm to VR trials subjects dropped the object 8 cm.

Despite these limitations, we were still able to show that young women were better able to minimize the disturbance to the whole-body COG with

reduced torque demands when accepting a weight in the outstretched hands using pre-planned preparatory movements in VL trials compared to corrective or feedback-driven response movements. Preparatory strategies were most successful and efficient (involving lower joint torques) when the subjects were allowed control of the rate as well as the onset of weight acceptance. Additionally, when the COG was initially farther from the boundaries of the BOS, the preparatory or postural control responses were able to more effectively minimize movement of the whole-body COG during weight acceptance.

In the next chapter of this thesis, this work is expanded by comparing postural control during weight acceptance in young and elderly women. However, in the future more work will be needed to understand the effects of limited preparation on the timing of muscle activation and also the effects of varying rates of weight acceptance.

2.6 Tables

Table 2.1 Subjects' completed trials summary

<u>Distance</u>	<u>Height</u>	<u>Condition</u>	<u>1*</u>	<u>2</u>	<u>3</u>	<u>4</u>	<u>5</u>	<u>6</u>	<u>7</u>	<u>8</u>	<u>9</u>	<u>10</u>	<u>11</u>	<u>12</u>	<u>13</u>	<u>14</u>	<u>15</u>	<u>Total**</u>	
Near	Low	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15	
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	14
	Medium	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	14
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	13
	High	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	14
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	13
Far	Low	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	14
	Medium	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	14
	High	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	12
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	13
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	12

* Subject numbers

** Total number of subjects that completed that combination

Table 2.2 Multivariate ANOVA results for two-way interactions

<u>Dependant Variable</u>	<u>Interaction</u>	<u>F</u>	<u>p</u>	<u>DOF</u>
COG	Distance * Height	0.319	0.732	2, 14
	Distance * Condition	0.703	0.512	2, 14
	Height * Condition	1.171	0.345	4, 28
COP	Distance * Height	1.460	0.266	2, 14
	Distance * Condition	0.334	0.722	2, 14
	Height * Condition	1.574	0.232	4, 28
Δ Thip	Distance * Height	0.194	0.826	2, 14
	Distance * Condition	0.588	0.507	2, 14
	Height * Condition	0.535	0.601	4, 28
Δ Tankle	Distance * Height	0.864	0.443	2, 14
	Distance * Condition	0.388	0.685	2, 14
	Height * Condition	0.860	0.459	4, 28
Δ Tshoulder	Distance * Height	0.141	0.849	2, 14
	Distance * Condition	0.860	0.445	2, 14
	Height * Condition	0.696	0.487	4, 28

2.7 Figures

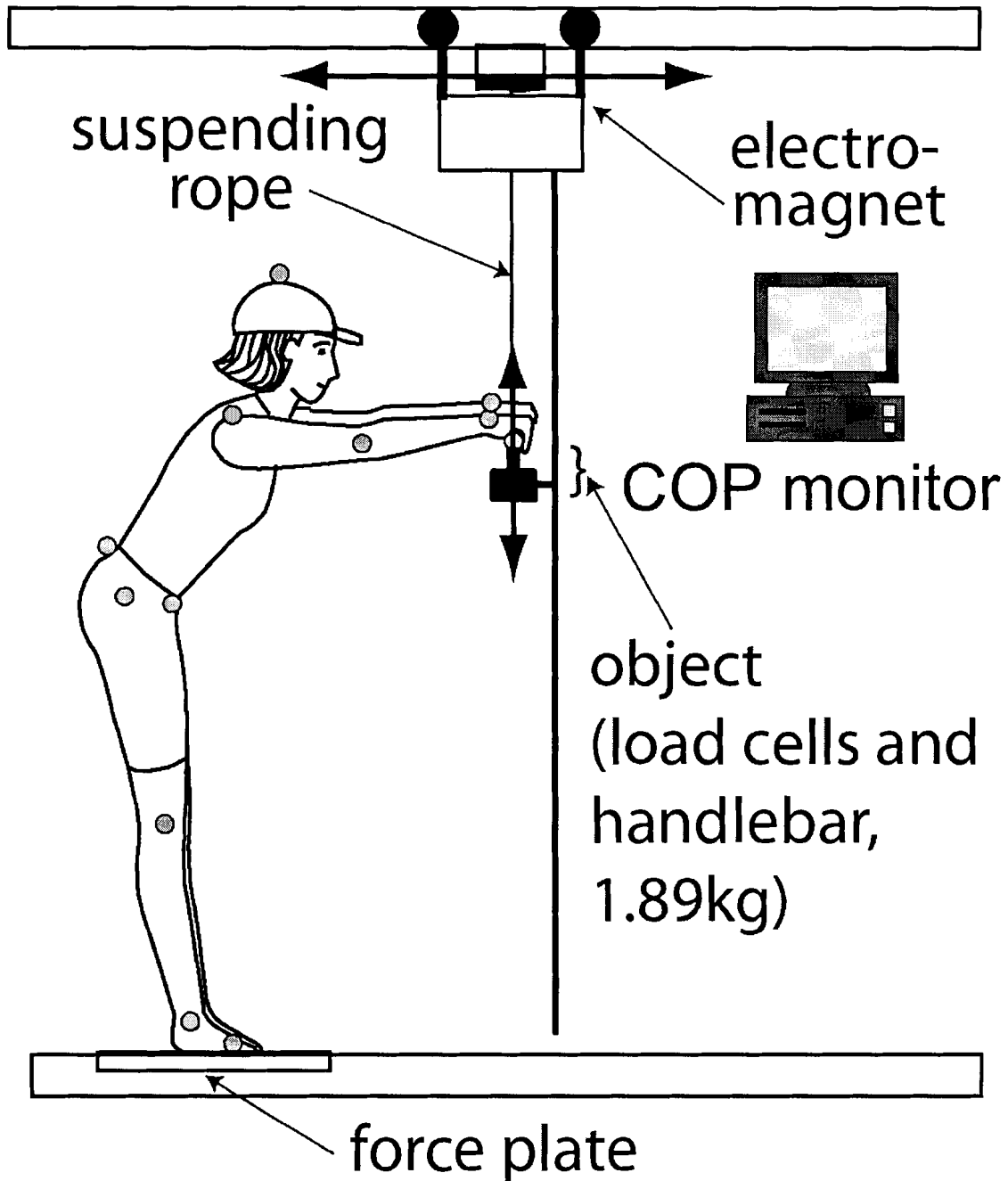


Figure 2.1 Experimental Schematic. The subject began the trials in a position of impending weight acceptance, with her hands lightly wrapped around the wooden cylindrical handlebar of the object. The COP was monitored prior to weight acceptance to ensure consistent initial body configurations between subjects and conditions.

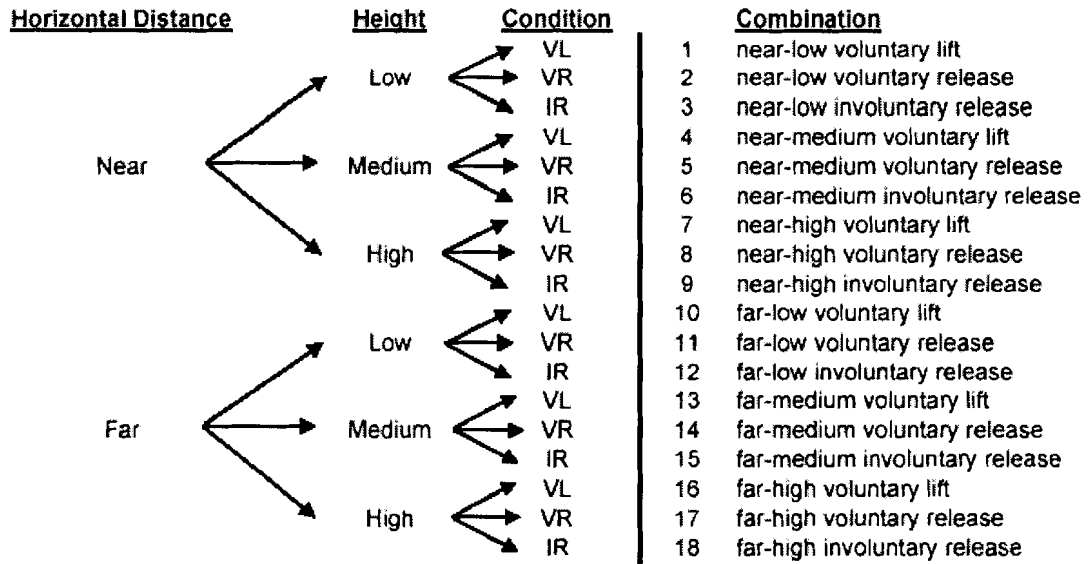


Figure 2.2 Experimental Design. The subject completed trial combinations in a random order. Each trial combination was repeated 3 times, consecutively, for a total of 54 trials..

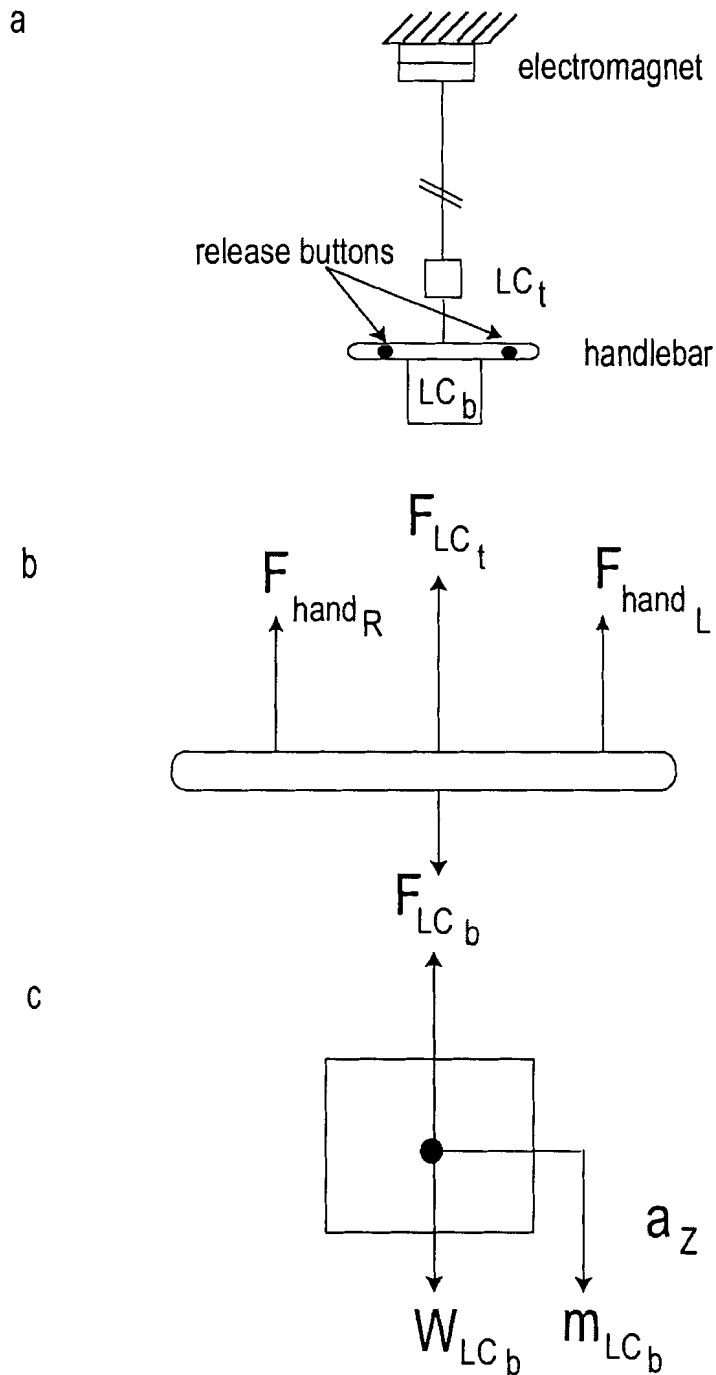


Figure 2.3.a. Object Schematic. The subject wrapped her hands around the handlebar so that her fingertips rested on top of the two release buttons. The handlebar was sandwiched between the two load cells which measured forces on the hand and object. **b.** Free body diagram of object handlebar. The force at the hands was calculated as the difference between the force of the object and the tension in the suspending rope. **c.** Free body diagram of bottom load cell. The force measured on the bottom load cell is due to the difference between gravitational and inertial forces.

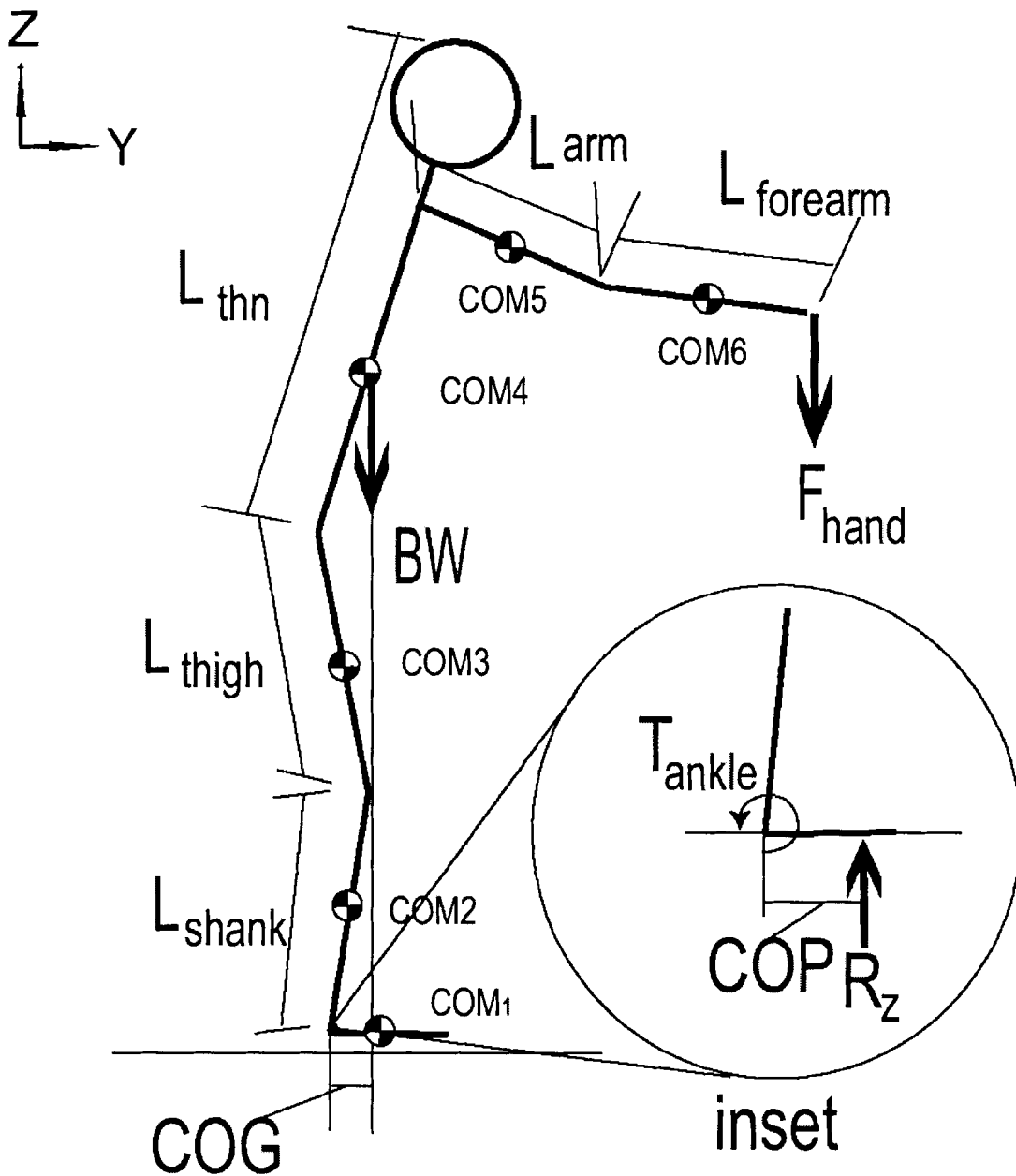


Figure 2.4 Sagittal plane 2D model of the body during weight acceptance. The model is comprised of six segments (i): the feet, the shanks, the thighs, the torso/head/neck (thn), upper arms and forearms with lengths (L_i) and centre-of-mass positions (COM_i) approximated by Dempster's data (D.A. Winter, 1990). COG calculations accounted for body segment COGs ($COG_i = m_i \cdot COM_i$) and the force at the hands. BW= body weight. **Inset.** COP was calculated from the ground reaction forces ($COP = T_{ankle} / R_z$).

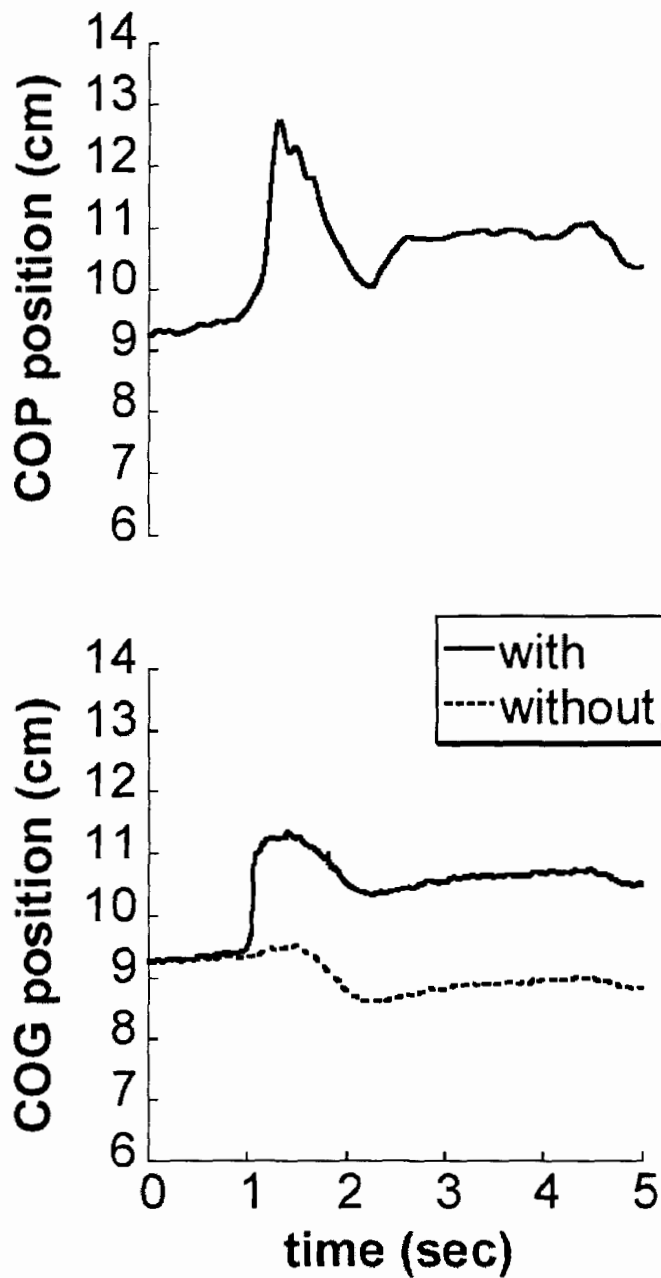


Figure 2.5 COG calculations with and without consideration for forces acting on the hand. If the force at the hand is not included in the calculation (dotted line) then the COG displacement is not very great compared to the COP displacement indicating that the subject is unstable, even when they have accepted the weight and are known to have maintained an upright position. Graphs are created from one subject (#13) during the VR condition. Position of the COP and COG are with respect to the subject's ankle.

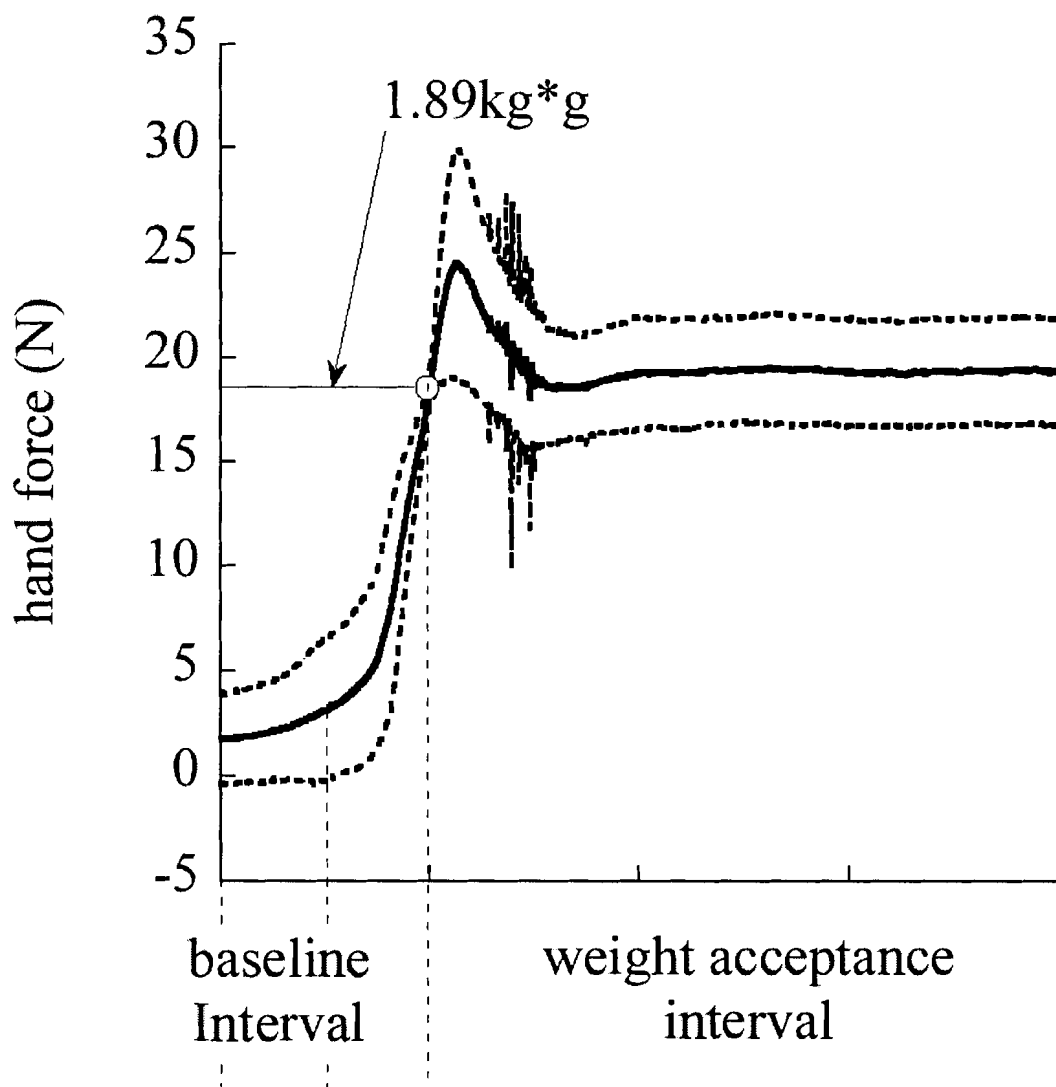


Figure 2.6 I determined the instant of weight acceptance to be the point in time when the hand force was equal to or greater than the weight of the object. The weight acceptance interval was defined as the 1.5s time interval starting at weight acceptance. Baseline values were measured from the 0.5s prior to object release.

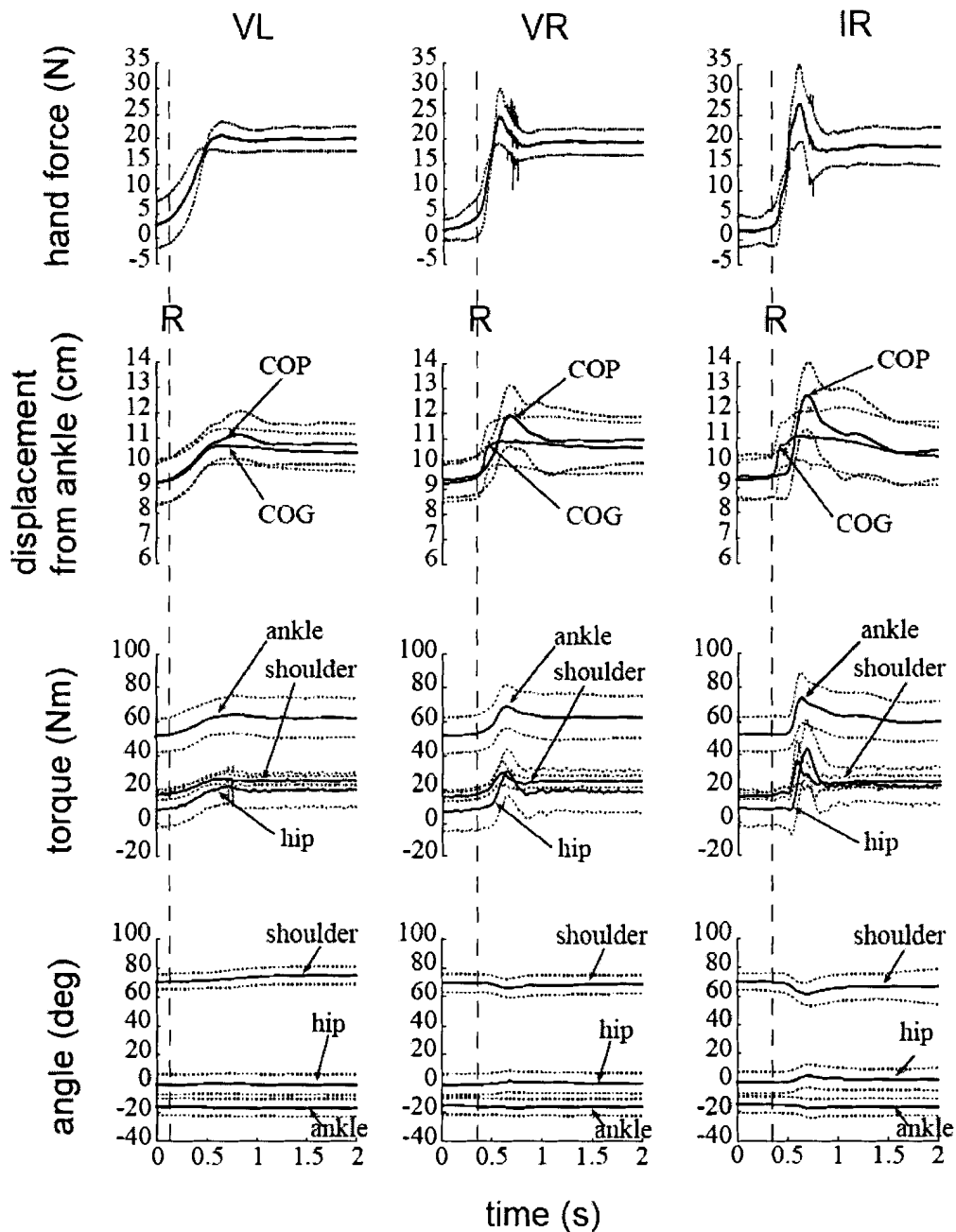


Figure 2.7 Composite traces were created by averaging data from each trial from all 15 subjects after synchronizing to the instant of weight acceptance. These composites were done only for trials conducted with the object in the near-medium position. Increases in hand force were accompanied by anterior movement of the COG and the COP, as well as, increases in plantarflexor, hip extensor and shoulder extensor torques. Furthermore, there were increases in ankle dorsiflexion, hip flexion and shoulder extension in VL trials and increases in ankle dorsiflexion, hip flexion and shoulder flexion in VR and IR trials. Rate of COG, COP, joint torque changes and angular rotations increased from VL to VR trials. R indicates moment of tension release. ± 1 SD shown as dotted traces

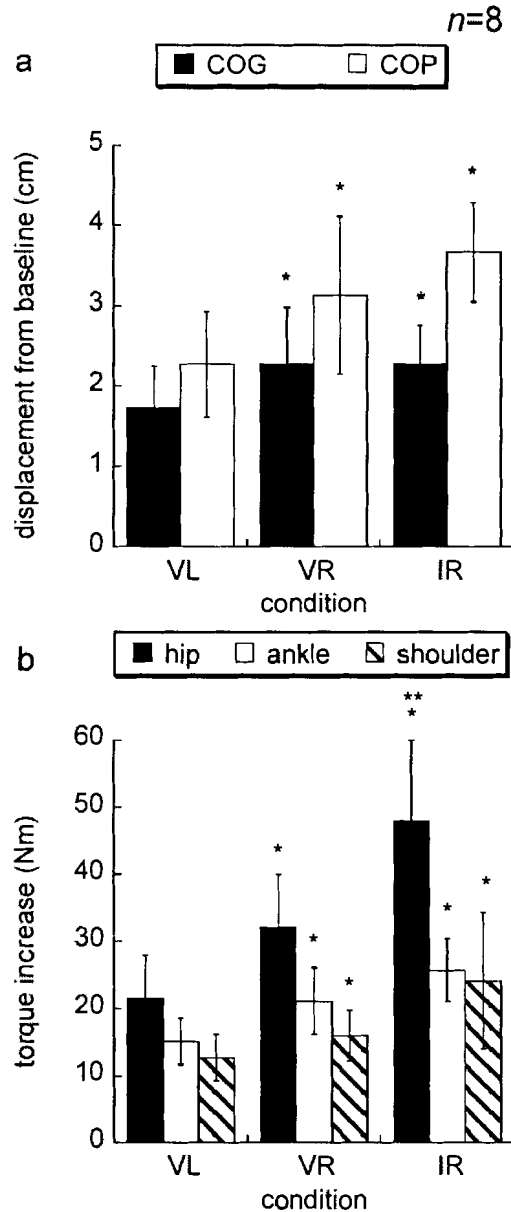


Figure 2.8 Effects of weight acceptance condition. **a.** There was 23.8% less COG displacement in VL than in VR trials but no change in COG displacement between VR and IR trials. Similarly, there was 28.8% decrease in COP displacement in VL than in VR trials but no change in COP displacement between VR and IR trials. **b.** There were 48.4% greater increases in hip torque, 59.4% greater increases in ankle torque and 25.8% greater increases in shoulder torque from VL to VR trials. There was a further 49.9% greater increase in hip torque from VR to IR trials, but no increase in ankle or shoulder torques. Bars represent means and error bars represent ± 1 SD. * indicates significant difference from VL, ** indicates significant difference from VR.

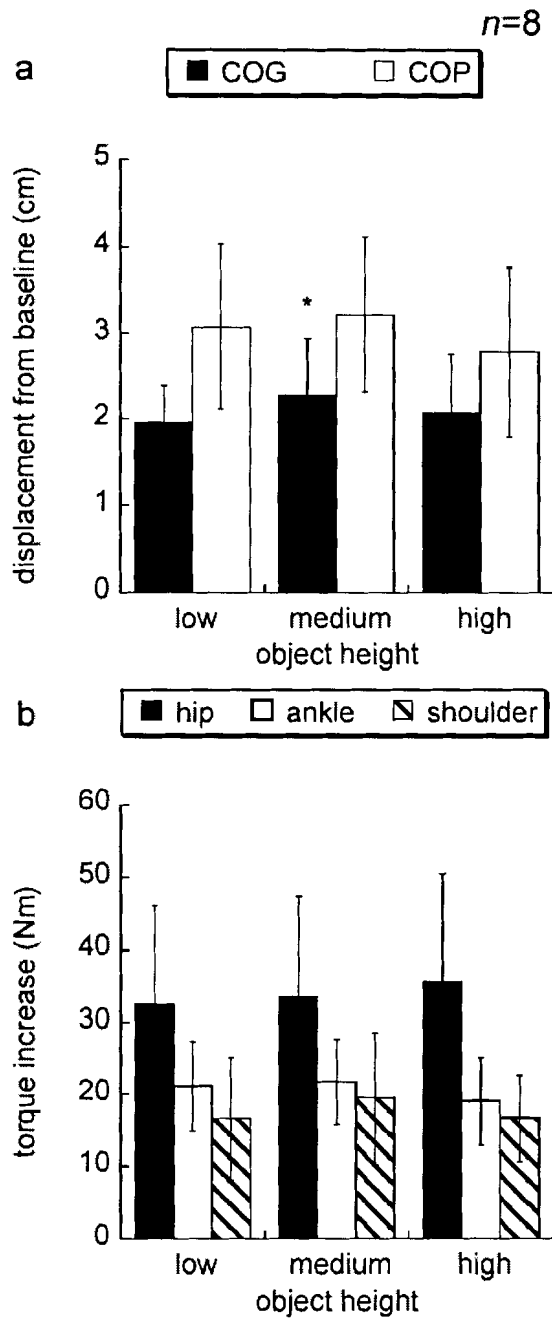


Figure 2.9 Effects of object height. **a.** There was 15.7% increase in COG displacement in the medium than in the low object height but no change in COG displacement between the medium and high object height, nor between the low and high object height. There was no significant difference in COP displacement between low, medium and high object heights. **b.** Object height had no effect on joint torque increases. Bars represent means and error bars represent ± 1 SD. *indicates significant difference from VL.

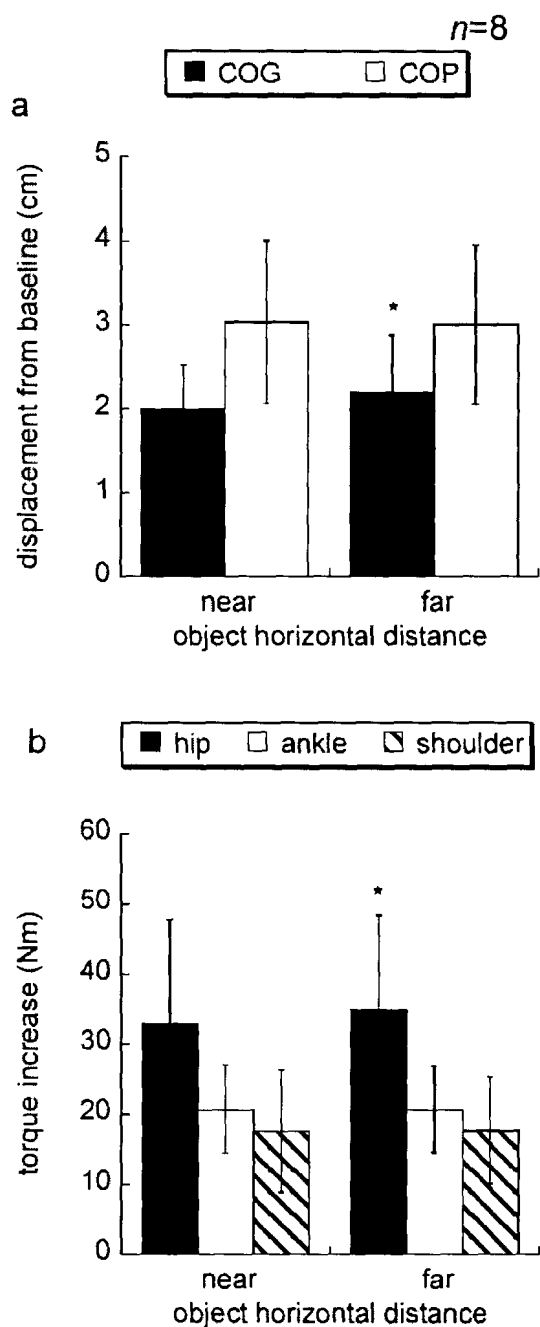


Figure 2.10 Effects of object horizontal distance. **a.** There was 9.9% increase in COG displacement in the far versus near horizontal distance of the object. However, there was no significant change in COP displacement between the near and far object distance. **b.** The increase in hip torque experienced during weight acceptance was 5.7% greater in the far than in the near object location. Bars represent means and error bars represent ± 1 SD. * indicates significant difference from VL.

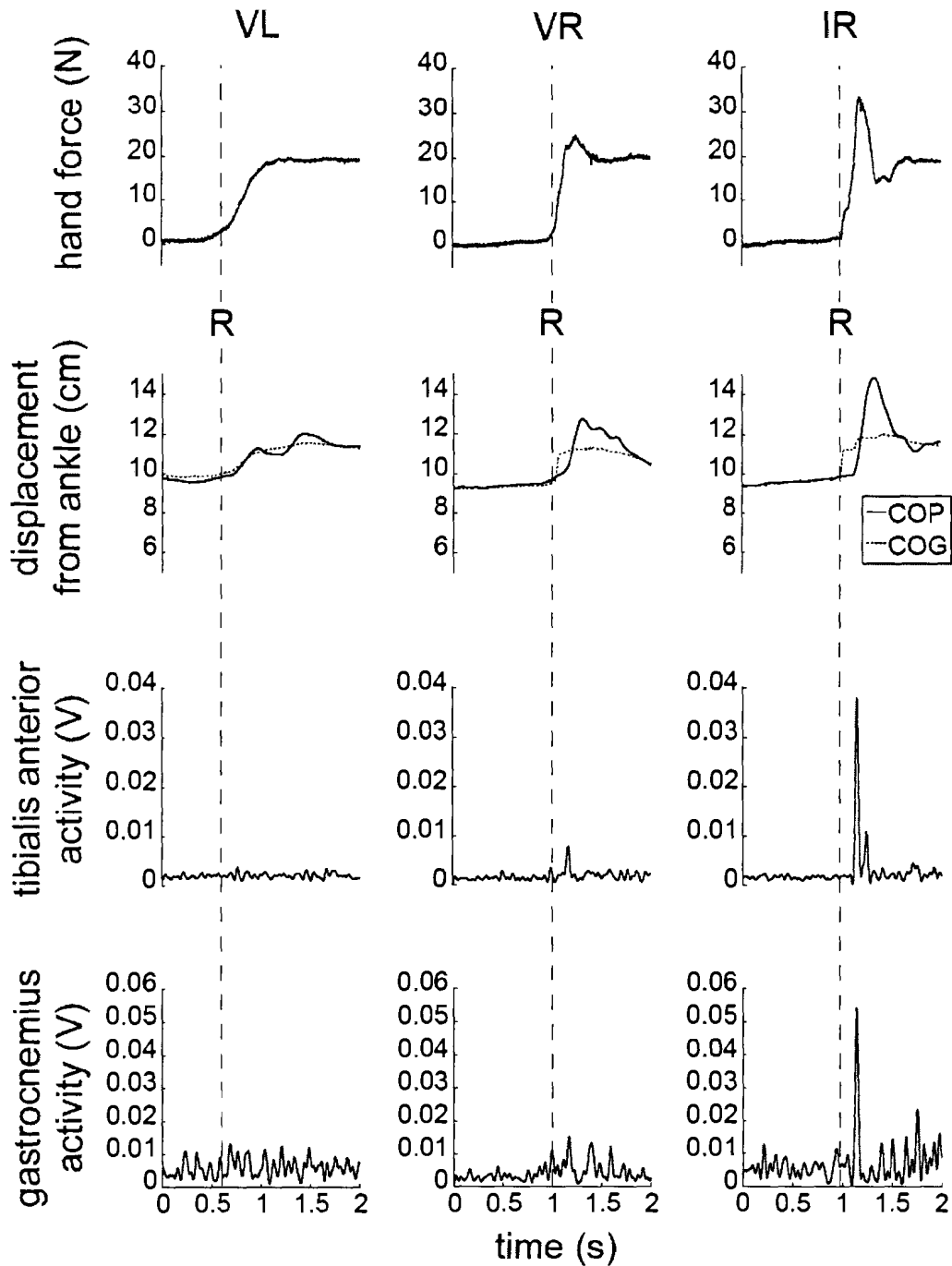


Figure 2.11 Typical EMG traces for the TA and medial gastrocnemius from one subject (#13) across weight acceptance conditions. During VL trials, there was no apparent change in the activity in either muscle. During VR trials, there was a small spike in both the TA and the medial gastrocnemius muscle approximately 200ms after weight release and synchronous with the peak in COP displacement. During IR trials, there was a significantly greater spike in both the TA and the medial gastrocnemius muscle, again at approximately 200ms after weight release. The spike in the medial gastrocnemius was much more diffuse than the spike in the TA.

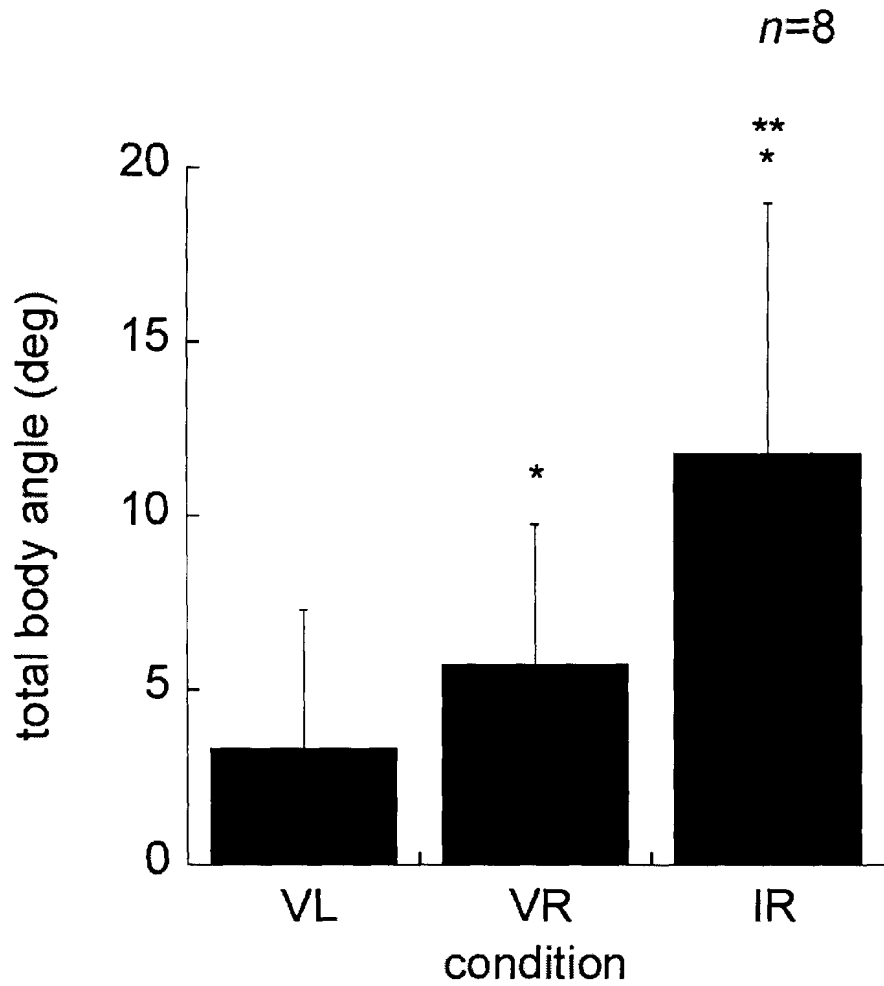


Figure 2.12 Corrective responses allowed greater changes in body joint angles than anticipatory responses. Changes in the total angle were 41.3% smaller in VL than in VR trials and 51.8% smaller in VR than in IR trials. Bars represent means and error bars represent +/-1SD. * indicates significant difference from VL.

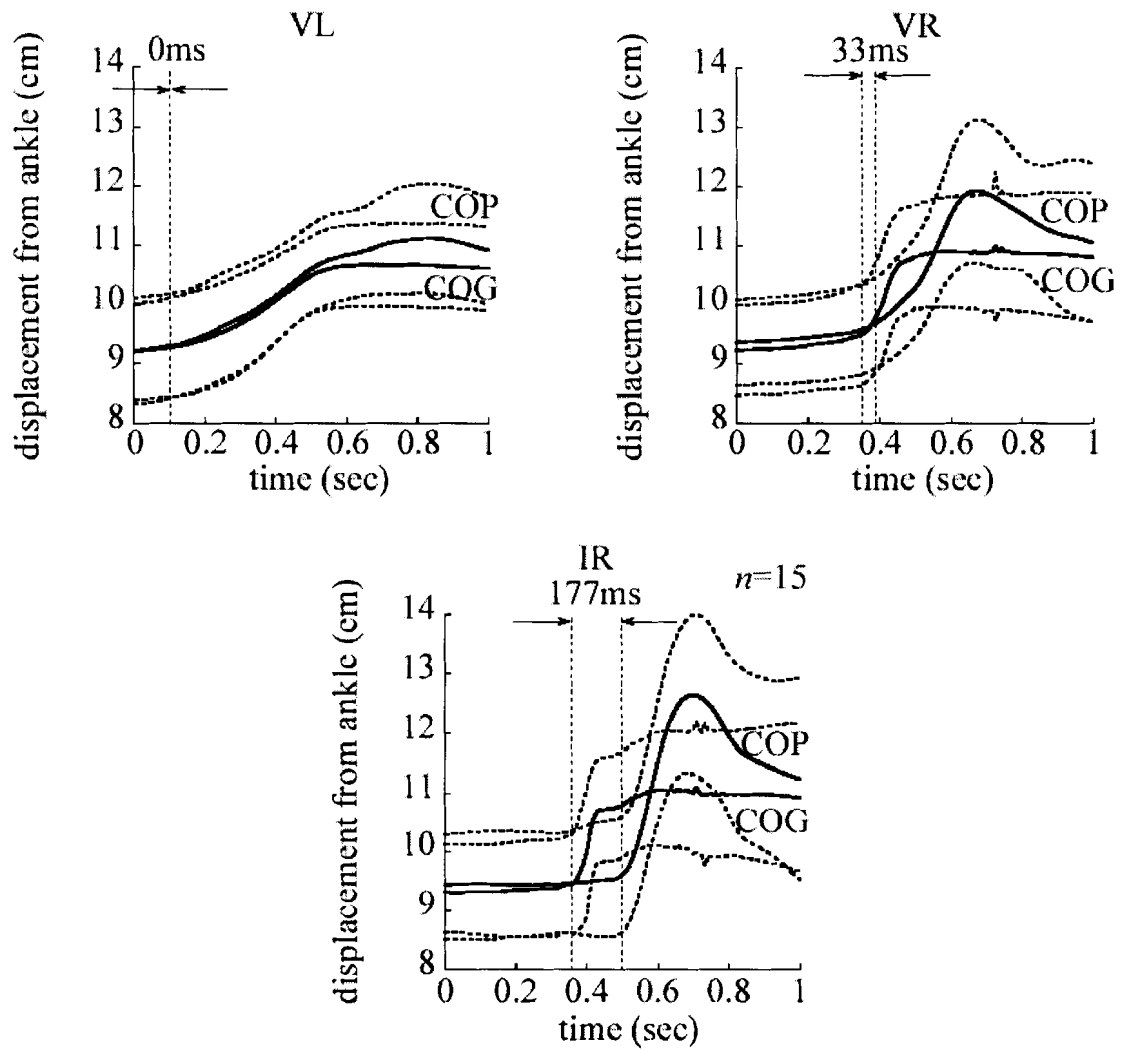


Figure 2.13 Composite traces indicate that on average COP rose in unison (no delay) with COG movement during VL trials. However, on average COP was delay 33ms after COG in VR trials and 177ms in IR trials. Dotted lines indicate +/- 1SD.

CHAPTER 3: INFLUENCES OF STRENGTH AND MOBILITY ON MAINTAINANCE OF STABILITY WHILE ACCEPTING A WEIGHT IN THE OUTSTRETCHED HANDS.

3.1 Abstract

Elderly individuals have been shown to have reductions in mobility and postural control. The goal of this study was to examine the ability of elderly women to minimize the disruption to their whole-body COG caused by weight acceptance to the outstretched hands. In particular I tested the following hypotheses: 1) elderly women experience greater whole-body centre-of-gravity (COG) displacement than young during weight acceptance, 2) elderly women generate greater centre-of-pressure (COP) displacement and joint torques than young in response to the greater COG displacements, 3) elderly women experience greater joint torque increases, COG and COP displacements than young when corrective instead of anticipatory responses are used and 4) elderly women experience greater joint torque increases, COG and COP displacements than young when the object position is increased in horizontal distance from the subject's ankles and vertical height. Young and elderly women ($n=15/\text{group}$) participated in the study. Subjects accepted the weight of an object, suspended by a rope, in one of three ways: (1) self-initiated (voluntary) lifting (VL), (2) self-initiated (voluntary) release of tension in the suspending rope (VR) or (3) investigator-initiated (involuntary) release. There were no differences in COG

and COP displacement between young and elderly women across all weight acceptance conditions. However, 10 of the elderly subjects could not complete the task when the object was placed in the far-high position and 5 of these women were unable to complete the task when the object was placed in the near-high position. The peak ankle torque generated during weight acceptance increased for those with greater functional reach and for those with smaller root mean squared medial-lateral excursion of the COP during quiet stance. Therefore, the results indicate the ability of elderly subjects to minimize the disruption to whole-body COG, due to weight acceptance, is similar to younger subjects. However, this ability to generate effective postural responses depends in part on joint strength and variables that also govern performance on functional stability tests.

3.2 Introduction

It is important to gain improved understand of the biomechanical factors that contribute to a human's ability to maintain balance following postural perturbations. Aging causes slowing of response times and increases in postural sway due to changes in both peripheral and higher level motor control systems (Stelmach et al., 1989). Further studies are still needed to understand how elderly individuals compensate for these deficits when faced with a perturbation.

Humans prepare for voluntary excursions of the centre-of-gravity (COG) during daily activities by producing anticipatory postural adjustments (APA), in the form of body segment movements and muscle activations which move the

centre-of-pressure (COP) simultaneous with the COG. However, in past studies elderly subjects demonstrate delayed APA responses (Williams, McClenaghan, & Dickerson, 1997) and differences in muscle activation patterns from those that characterize the APAs seen in younger adults (Inglin & Woollacott, 1988). Given that elderly have these deficits in APA production, it can be expected that older women would have a decreased ability to properly anticipate and prepare for a perturbation to the whole-body COG. Furthermore, it would be expected that decreases in the ability to anticipate a perturbation would cause increases in the disruption to the whole-body COG. Previous studies have shown that elderly adults experience greater displacement of body segments and their whole-body COG during perturbations to the whole body, compared to young adults (Garland, Stevenson, & Ivanova, 1997; Pyykko, Jantti, & Aalto, 1990) suggesting that they are less able to anticipate than younger subjects. The present research proposes to examine the abilities of elderly individuals to anticipate and prepare to minimize disruption to the whole-body COG when faced with a perturbation to the upper extremity.

Previous research indicates that increases in instability are related to declines in strength, response times and torque production (S. R. Lord & Ward, 1994; Mackey, 2004; Robinovitch, Heller, Lui, & Cortez, 2002). Standard tests are used to measure stability. In particular these tests provide information on the magnitude and rate of the COG disturbance that can be experienced without losing balance. For example, during functional reach, the COG is displaced

within a fixed BOS through forward reaching, and the distance beyond arms-length is measured (Duncan et al., 1990). This distance has been found to be associated with the individual's ankle strength (Q. Liu, Graham, Hall, & Robinovitch, 2001). Furthermore, the peak ankle torque used to perform the test was limited by the available ankle torque. It could therefore be speculated that these measures of stability and strength would also be related to the disruption experienced by the COG, COP and also the joint torque increases developed during other perturbing movements.

The goal of this study was to examine the ability of elderly women to maintain stability during weight acceptance to the outstretched hands. In particular I tested the following hypotheses: 1) elderly women experience greater whole-body centre-of-gravity (COG) displacement than young during weight acceptance, 2) elderly women generating greater centre-of-pressure (COP) displacement and joint torques than young in response to the greater COG displacements, 3) elderly women experience greater joint torque increases, COG and COP displacements than young when corrective instead of anticipatory responses are used and 4) elderly women experience greater joint torque increases, COG and COP displacements than young when the object position is increased in horizontal distance from the subject's ankles and vertical height.

3.3 Methods

3.3.1 Subject Recruitment

Data was collected for 15 elderly women (mean +/-SD: age=74.73+/-7.36 years, height=1.57+/-0.069m, body mass=62.31+/-16.62kg) and compared to data collected for the 15 young women in the previous study (See section 2.3.1). Elderly subjects were recruited from postings of notices in local community and recreational centers. Subjects were interviewed initially to see if they passed the exclusion criteria. Subjects were excluded if they had a Folstein Mini-Mental State Examination (FMMSE) score of less than 24, had uncorrected vision problem, had a history of severe orthopedic or neuromuscular problems or were taking medications known to cause balance deficits. All participants signed a written consent form (Appendix A) and the experiment was approved by the Research Ethics Committee at Simon Fraser University.

3.3.2 Ancillary Measures

I conducted an ancillary measures session with the subject prior to her participation in the weight acceptance reaching protocol. During this session, I took ancillary measures of the subject's anthropometrics (height, weight, and segment lengths). I also had each subject perform standard tests of mobility and postural stability. These tests included: functional reach (Duncan et al., 1990), Get-up-and-go (Mathias, Nayak, & Isaacs, 1986), the Chair rise test (Guralnik et al., 1994), and COP sway during quiet stance with the eyes either open or closed while the subject stood on either a rigid surface or a piece of foam. Furthermore, I tested the subject's simple and choice (2 and 4 choices) reaction times in the

upper extremity and her simple reaction time in the lower extremity. Finally, I conducted standardized tests of isometric strength for dorsiflexion (elderly only), hip extension, hip flexion and shoulder extension (protocols outline in Appendix C). Unfortunately, ancillary measures were not collected for 2 young subjects and no quiet stance testing was conducted on an additional 3 young subjects and 1 elderly subject due to equipment malfunction.

3.3.3 Weight Acceptance Reaching Protocol

Subjects of both age groups completed the weight acceptance reaching protocol outlined in Chapter 2, section 2.3.2.

3.3.4 Data Analysis

Data analysis was carried out as described in Chapter 2 (section 2.3.3) and dependant variables were the same as those described in Chapter 2 (section 2.3.4). However, dependant variables were normalized for comparison between young and elderly subjects. COG and COP displacement were normalized to body height, while joint torques were normalized to the product of body height times body weight. For hypotheses testing, I used a four-factor repeated measures multivariate analysis of variance (MANOVA, $\alpha=0.05$) to identify any significant main effects or two-way interactions for age (2 levels: young or elderly), weight acceptance condition (3 levels: VL, VR or IR), horizontal distance of object (2 levels: near or far) and object height (2 levels: medium or low). Pairwise comparisons with Bonferroni's correction and a total alpha level of 0.05 were used to further understand significant main effects of height and weight

acceptance conditions. All statistical tests were carried out using SPSS statistical analysis software (v.12.0 for windows, SPSS Inc., Chicago, IL, USA). Unless otherwise stated, reported results are means +/- 1 SD.

3.3.5 Supplementary Analysis

I had 5 elderly women that could not perform the task (in any condition) when the object was placed at the near-far position. Therefore, I further divided elderly subjects into two functional groups, those that could perform the near-high position (Group 2A) and those that could not (Group 2B). I then compared the mean ancillary values of mobility, strength and reaction time between the three different groups (Young vs Group 2A vs Group 2B) using paired T-tests and a total alpha level of 0.05.

Angular changes at individual joints ($\Delta\theta_{hip}$, $\Delta\theta_{ankle}$, $\Delta\theta_{shoulder}$) and total body angle changes (sum of $\Delta\theta_{hip}$, $\Delta\theta_{ankle}$, $\Delta\theta_{shoulder}$) were calculated and compared between different weight acceptance conditions using a one-way repeated measures ANOVA.

Additionally, I created composite traces of dependant measures for both the young and elderly groups by averaging data across all subjects in the group after synchronizing to the instant of weight acceptance. This analysis was done only for trials in the near-medium position. As well, I used paired T-tests to test the differences in standard deviations of the composite traces between young and elderly subjects.

As a measure of response time, I calculated the time interval between the onset of COG anterior movement and COP anterior movement. Using one-way repeated measures ANOVAs, I compared response times between the different weight acceptance conditions for each age group. The initiation of anterior movement of the COG and COP were calculated as the first point in time when the trace rose 2 standard deviations above mean initial values.

Furthermore, I used Pearson's correlation coefficients ($\alpha=0.05$) to identify associations between COG displacement, COP displacement, peak joint torques (hip and ankle) and strength measurements. I also calculated associations between COG displacement, COP displacement, reaction time, tests of mobility and tests of stability. Unless otherwise stated, reported p values are for r not r^2 . Unfortunately, due to limitations with programming and the nature of the COP and COG displacement traces, no correlations could be made with quantitative measures of response time.

3.4 Results

Of the 15 elderly women that participated in the study, only 5 were able to perform the weight acceptance task in all positions. Subjects were considered to be unable to perform the task if after more than 10 attempts, they still 1) took a step upon weight acceptance or 2) could not maintain the initial position long enough to start the trial. Ten women were unable to complete the trials (in all weight acceptance conditions) where the object was placed in a far-high position and of those, 5 women were also unable to complete trials where the object was

placed in a near-high position. Two young subjects were also unable to complete the task when the object was placed at the far-high position. I collected and analyzed only trials that were successfully completed (including only two object heights, low and medium). Furthermore, analysis of hand forces after trial collection indicated that there were some trials where subjects were applying a force to the object prior to weight acceptance, and these were also eliminated from analysis (total acceptable trials per subject are outlined in Table 3.1.a-b).

3.4.1 Hypotheses results

In general the patterns of COG movement, COP movement and joint torque change with weight acceptance were similar between young and elderly women (Figure 3.1.a, b, c). Although, the variability (standard deviation) of the hip ($T_{0.05,722}=-97.784$, $p<0.001$), ankle ($T_{0.05,722}=-75.359$, $p<0.001$) and shoulder ($T_{0.05,722}=-65.865$, $p<0.001$) torque composite traces were greater in elderly subjects compared to young. Alternatively, elderly had less variability in COG displacement ($T_{0.05,722}=13.598$, $p<0.001$).

There was no significant main effect of age in COG or COP displacement (COG: $F_{0.05, 1,24}=1.289$, $p=0.267$, COP: $F_{0.05, 1,24}=0.066$, $p=0.799$, Figure 3.2.a). Furthermore, there was no significant main effect of age on the increase in hip ($F_{0.05, 1,24}=0.053$, $p=0.820$), ankle ($F_{0.05, 1,24}=0.350$, $p=0.560$) or shoulder ($F_{0.05, 1,24}=0.000$, $p=0.984$) torque during weight acceptance (Figure 3.2.b).

A significant main effect of weight acceptance condition was seen for COG ($F_{0.05, 2, 48}=36.419$, $p<0.001$) and COP ($F_{0.05, 2, 48}=97.850$, $p<0.001$)

displacement, as well as increases in hip ($F_{0.05, 2, 48}=139.037$, $p<0.001$), ankle ($F_{0.05, 2, 48}=89.523$, $p<0.001$) and shoulder ($F_{0.05, 2, 48}=45.057$, $p<0.001$) torques (Figure 3.3.a,b). Specifically, there was greater COG, COP and torque increases in the IR compared to the VR trials, and in the VR trials compared to the VL trials (see Table 3.2).

With both young and elderly subject's combined, there was significantly greater COG displacement when the object was at the medium compared to low height ($0.0118\pm 0.0042\text{cm/cm}$ vs. $0.0108\pm 0.0033\text{cm/cm}$, 95%CI diff=0.000 to 0.002cm/cm , $F_{0.05, 1, 24}=9.012$, $p=0.006$, Figure 3.4.a). On the other hand, unlike results from young subjects only (section 2.4.1, which tested 3 object heights), there was no significant main effect of object height on COP displacement ($F_{0.05, 1, 24}=3.602$, $p=0.070$, Figure 3.4.a), and the increases in ankle torque ($F_{0.05, 1, 24}=0.106$, $p=0.748$) with weight acceptance (Figure 3.4.b). Moreover, while young subjects alone (with three object heights) showed no significant main effect of object height on hip and shoulder torque increases, young and elderly subjects combined did (hip: $F_{0.05, 1, 24}=9.007$, $p=0.006$, shoulder: $F_{0.05, 1, 24}=25.491$, $p<0.001$, Figure 3.4.b). Specifically, there were greater increases in hip and shoulder torques in the medium compared to the lower heights (hip: $0.314\pm 0.137\text{Nm/kg}\cdot\text{m}$ vs. $0.299\pm 0.134\text{Nm/kg}\cdot\text{m}$, 95%CI diff=0.005 to $0.025\text{Nm/kg}\cdot\text{m}$, shoulder: $0.173\pm 0.074\text{Nm/kg}\cdot\text{m}$ vs. $0.156\pm 0.069\text{Nm/kg}\cdot\text{m}$, 95%CI diff=0.011 to $0.026\text{Nm/kg}\cdot\text{m}$).

Similar to results for young only (section 2.4.1), for both young and elderly subject's combined increasing the object's horizontal distance from the subject's ankles resulted in an increase in COG ($F_{0.05, 1, 24}=13.349$, $p=0.001$) displacement and greater increases in hip torque ($F_{0.05, 1, 24}=21.964$, $p<0.001$) with weight acceptance (Figure 3.4.c,d).

For both young and elderly subjects combined, there was a significant two-way interaction between horizontal distance and height on COG ($F_{0.05, 1, 24}=10.449$, $p=0.004$) and COP displacement ($F_{0.05, 1, 24}=13.295$, $p=0.001$) as well as the increase in ankle torque ($F_{0.05, 1, 24}=8.289$, $p=0.008$). Furthermore, there was a significant two-way interaction between object height and weight acceptance condition for young and elderly subjects combined on the increase in ankle torque ($F_{0.05, 2, 48}=3.270$, $p=0.047$). However, there were no two-way interactions between the age factor and the other trials conditions (including object height, horizontal distance and weight acceptance condition, see Table 3.3).

3.4.2 Supplementary results

Qualitative analysis of composite traces indicated greater response times in elderly than young subjects (Figure 3.5). On average, COP began to rise in unison with onset of COG movement in VL trials for young subjects; however, COP onset was, on average, 89ms prior to COG movement in VL trials for elderly subjects. During VR trials young subjects experienced anterior movement of their COP, on average, 33ms after COG movement while COP

onset was, on average, 130ms prior to COG movement in elderly. Then during IR trials, there was an average 177ms delay in COP onset after COG onset in young subjects, but an average 194ms delay for elderly subjects.

I found no significant differences in hip extensor ($T_{0.05, 25}=0.170$, $p=0.866$), hip flexor ($T_{0.05, 25}=0.829$, $p=0.415$) or shoulder extensor ($T_{0.05, 25}=0.657$, $p=0.517$) strength in young compared to elderly subjects. However, body height ($T_{0.05, 28}=5.223$, $p<0.001$) and functional reach ($T_{0.05, 25}=4.185$, $p<0.001$) were significantly smaller in elderly compared to young; there were also increases in Get-up-and-go time ($T_{0.05, 26}=-3.987$, $p<0.001$), Sit-to-stand time ($T_{0.05, 26}=-4.109$, $p<0.001$), and reaction times (simple: $T_{0.05, 19}=-3.443$, $p=0.003$, choice: $T_{0.05, 20}=-2.710$, $p=0.013$, lower extremity: $T_{0.05, 23}=-2.888$, $p=0.008$) for elderly compared to young (Table 3.4). Furthermore, I found no significant differences in tests of mobility and significant differences in only one measure of stability between Group 2A and Group 2B elderly subjects. There was significantly greater root mean squared (RMS) medial-lateral COP excursion during quiet stance (on a rigid surface with eyes open) in the Group 2B compared to the Group 2A ($T_{0.05, 12}=-3.694$, $p=0.003$). The only significant anthropometric difference that I found between the Group 2A and Group 2B elderly was much greater anterior foot length (normalized to body height) in the Group 2A compared to the Group 2B (11.74 ± 0.37 vs. 11.12 ± 0.26 cm, $T_{0.05, 13}=3.399$, $p=0.005$) and greater anterior foot length in Group 2A subjects compared to young subjects (11.74 ± 0.37 vs. 11.11 ± 0.63 cm, $T_{0.05, 23}=-2.891$, $p=0.008$, Table 3.4).

I found that there was a significant positive association between the average (across all trials) COG displacement for a subject and her average COP displacement ($r^2=0.884$, $p<0.001$). Furthermore, there was a significant positive association between COG displacement and normalized measures of hip flexor strength ($r^2=0.299$, $p=0.003$, Figure 3.6.a), and shoulder extensor strength ($r^2=0.296$, $p=0.003$). There was also a positive significant correlation between the average COG displacement and hip extensor strength ($r^2=0.321$, $p=0.027$, Figure 3.6.b) and between the average COG displacement and dorsiflexor strength ($r^2=0.314$, $p=0.030$, Figure 3.7.) in elderly only. Furthermore, I found that there was a positive correlation between the average COG displacement and the peak hip ($r^2=0.206$, $p=0.012$, Figure 3.8.a) and ankle ($r^2=0.605$, $p<0.001$, Figure 3.8.b) torques.

My results also indicate that there were significant positive associations between the peak hip torque, the peak ankle torque and the peak shoulder torque (hip vs. ankle: $r^2=0.162$, $p=0.027$, hip vs. shoulder: $r^2=0.211$, $p=0.011$, ankle vs. shoulder: $r^2=0.674$, $p<0.001$). More interestingly, I found that peak ankle torque was related to the subject's Get-up-and-go time ($r^2=0.164$, $p=0.032$) and the root mean squared medial-lateral COP excursion during quiet stance on a rigid floor with eyes open ($r^2=0.194$, $p=0.032$, Figure 3.9.a). As well, I found a significant positive association between the peak ankle torque and functional reach ($r^2=0.260$, $p=0.007$, Figure 3.9.b).

I found no significant differences in the total body angle changes in elderly compare to younger subjects ($F_{0.05, 1, 10}=0.227$, $p=0.644$) and no significant differences were seen between young and elderly subjects in the object's deflection in vertical height during weight acceptance ($F_{0.05,1, 23}=0.003$, $p=0.956$).

3.5 Discussion

I observed no differences in COP and COG displacements between the two age groups suggesting that my older subjects are similar to my young subjects in their ability to minimize COG displacement during weight acceptance to the outstretched hand. On the other hand, several of the elderly subjects were unable to accept the weight of the object without taking a step or losing balance when the object was placed in a high position. Therefore, these results suggest that beyond control of COG displacement, there were other components to this movement that limited the individual's ability to successfully perform the weight acceptance task.

Contrary to the results seen in Chapter 2 (Section 2.4.1), when both young and elderly subjects were included in the analysis, lose of control over both the rate and the onset of weight acceptance (as in IR trials) required greater COP displacement and joint torque increases compared to when only control over the rate of weight acceptance was lost (as in VR trials). These differences in results may be in part due to the absence of the high object height position in the statistical analysis, but are most likely due to the size of the population studied in the first part of this thesis. The effects of the VR and IR conditions most likely did

not reach significance with the smaller population, particularly with Bonferroni's correction. Regardless, the results from the present study, which examined the combined results from both young and elderly subjects, suggest that although subjects are as able to minimize COG displacement during weight acceptance in the IR as the VR trials, less COP displacement and joint torque increases are required to achieve this task in the VR trials. Specifically, when subjects lost control over the onset of the perturbation, they were required to use greater COP movements and joint torques to minimize the COG to the same degree as when they had control over the perturbation onset.

These results complement and expand upon previous studies which examined the effects of voluntary versus involuntary perturbation preparation. More specifically, these results suggest that, similar to the production of APAs, more appropriate corrective responses depend in part on subject knowledge of perturbation timing and magnitude. Previously, it has been shown that when lifting an unexpectedly heavy load, the activation of the postural stabilizing muscles was insufficient to produce the necessary whole-body torques to lift the load (Commissaris & Toussaint, 1997b; Toussaint et al., 1998). Muscle activity was increased in response to the increased perturbation to develop the necessary torque to lift the load and regain stability. Similarly, I found that when subjects no longer had control over the onset of the weight acceptance, subjects had to increase joint torques to complete the task and achieve the same minimal

movement of the COG that they were able to obtain when the perturbation was fully voluntary.

The correlations achieved in my study were inconsistent with past research which indicated that increases in strength lead to increases in postural stability (Choy et al., 2003; Izquierdo et al., 1999; Wu, Zhao, Zhou, & Wei, 2002). Specifically, improvements in the knee extensor strength through Tai Chi training has been seen to also decrease COP excursion during quiet sway (Wu et al., 2002). However, many of these previous studies have not examined the association of strength on more dynamic movements or those that include multiple degrees of freedom. My unexpected results suggest that perhaps those with impairments in joint strength experienced less COG displacement because they adopted alternative strategies (provided to them because of their degree of freedom) to maintain COG positioning that decreased the reliance on their depleted strength and increased their response time to the perturbation. Decreases in response time would cause an earlier onset of COP movement to reduce the disruption to the whole-body COG and decrease the required joint torques needed.

Previous research has already found that decreases in strength and increases in age-related impairments are associated with changes in postural control strategies (Accornero, Capozza, Rinalduzzi, & Manfredi, 1997), as well as changes in strategies used to perform everyday tasks such as lifting (Puniello, McGibbon, & Krebs, 2001). In particular, elderly subjects have been seen to

employ a more rigid stance during quiet stance tests than younger subjects (Accornero et al., 1997), and decreases in knee strength have been associated with use of a back dominant instead of a leg dominant strategy during a free standing box lift (Puniello et al., 2001). Moreover, other studies have found that elderly subjects used more co-contraction during quiet stance with a perturbation than younger subjects (Benjuya et al., 2004; Ho & Bendrups, 2002). By employing an alternative strategy, my weaker subjects may have also limited the feasible range of objects positions in which they could successful accept the weight, without taking a step, because of the smaller torques used. More specifically, had my weaker subjects adopted a more rigid stance in expectation of the forthcoming perturbation, they may have been able to increase their response time while decreasing the torque demands at their joints (or the effective strength required) in order to minimize their COG displacement to the same degree as stronger subjects. However, a more rigid stance would have also limited the flexibility of their strategy when the initial position increased the initial torque requirements. Further analysis is required to test this hypothesis and other possible alternative strategy that may have been used by weaker subjects.

In further support of this alternative strategy use, I found significant correlations between peak ankle torque and different tests of mobility which suggest that those with increases in impaired mobility used less peak ankle torque on average than those with full mobility. These results again contradict

the current knowledge that poor performances on mobility tests are associated with increases in instability (Duncan et al., 1990; Okuzumi et al., 1995; Tanaka et al., 1997; Weiner et al., 1992). Specifically, Get-up-and-go times have been found to increase in those with age-related deficits in mobility (Samson et al., 2000). Moreover, functional reach, which was produced as a test of stability, has been shown to be reduced in frail elderly males (Duncan et al., 1990; Weiner et al., 1992). Furthermore, previous research in our lab has also suggested that those with decreases in strength in the ankle will rely more heavily on hip flexion to achieve their functional reach, while those with hip weakness will rely more heavily on ankle dorsiflexion (Q. Liu et al., 2001). Those with weakness in both joints and/or an overall mobility impairment (i.e. other neuromuscular deficits) were not able to compensate by increasing reliance on the other joint, and would therefore demonstrate decreases in their functional reach. The current study suggests that impaired mobility also influences the strategy used to perform the task of accepting a weight in the outstretched hand. It appears that subjects that had decreases in mobility may have used an alternative strategy which reduced the use of joint torques (and ankle torque specifically). Therefore, my results further support the idea that not only are mobility impairments associated with declines in postural stability, but that impairments in mobility have an effect on the reliance on joint torques used to maintain postural control.

There are several important limitations to this research. First, results of this study with healthy women may have limited applicability to males.

Second, past experiments within the Injury Prevention and Mobility Laboratory have shown that elderly women have a smaller whole-body lean angle and initial COP position during quiet stance compared to younger women (Mackey, 2004). The present protocol required that all subjects maintain a target COP position based on values seen for young subjects holding their arms in each of the object positions (Appendix B). Consequently, elderly subjects may have been required to begin trials with their COP positioned further forward than was typically natural for them, placing them in greater initial instability. Furthermore, any variations in anthropometrics (specifically, height or anterior foot length) may have caused differences in initial joint torque demands between subjects and groups which could make the task more difficult for those with greater initial torques and/or weaker muscles. However, I felt that using target COP positions based on the performances of young subjects would help control body configurations between groups and trials, as well as test the effects of the age-related differences in stability.

Last, and perhaps most importantly, the elderly subject population chosen to participate in this study was very healthy and independent. The lack of differences between young and elderly subjects may in part be due to the recruitment of this relatively healthy group of senior women. No significant differences were found between the strength capabilities of the young subject population and that of the senior women (Table 3.4). However, the correlations that we found between the average COP and COG displacement versus

dorsiflexor, hip flexor and shoulder flexor strength suggest that had we used elderly with obvious deficits in strength, there may have been a significant change in the abilities of my elderly subjects to minimize COG displacement. Future studies could extend the present results by completing the protocol with a more frail elderly population.

Future analysis and studies including alternative elderly subject populations (such as those with greater impairments like fallers or nursing home residents) are necessary to understand the unexpected lack of difference between young and elderly subjects. As well, further analysis of response times and EMG traces could provide greater understand of the unexpected correlations. This analysis could include quantitative analysis of the differences in response times between young and elderly subjects, as well as between Group 2A and Group 2B subjects.

3.6 Tables

Table 3.1.a Young subjects' completed trials summary

<u>Distance</u>	<u>Height</u>	<u>Condition</u>	<u>1*</u>	<u>2</u>	<u>3</u>	<u>4</u>	<u>5</u>	<u>6</u>	<u>7</u>	<u>8</u>	<u>9</u>	<u>10</u>	<u>11</u>	<u>12</u>	<u>13</u>	<u>14</u>	<u>15</u>	<u>Total**</u>	
Near	Low	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15	
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	14
	Medium	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	14
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	13
	High	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	14
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	13
Far	Low	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	14
	Medium	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	14
	High	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	12
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	13
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	12

Table 3.1.b Elderly subjects' completed trials summary

<u>Distance</u>	<u>Height</u>	<u>Condition</u>	<u>1*</u>	<u>2</u>	<u>3</u>	<u>4</u>	<u>5</u>	<u>6</u>	<u>7</u>	<u>8</u>	<u>9</u>	<u>10</u>	<u>11</u>	<u>12</u>	<u>13</u>	<u>14</u>	<u>15</u>	<u>Total</u>	
Near	Low	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
	Medium	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
	High	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	11
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	10
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	10
Far	Low	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
	Medium	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	15
	High	VL	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	4
		VR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	5
		IR	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	✓	4

* Subject numbers

** Total number of subjects that completed that combination

Table 3.2 Pairwise comparisons for main condition effect

<u>Dependant Variable</u>	<u>Comparison</u>	<u>Trial type 1*</u>	<u>Trial type 2</u>	<u>95%CI diff</u>	<u>p</u>
COG (cm/cm)	VL vs VR	0.0093 0.0028	0.0122 0.0038	0.002 to 0.004	<0.001
	VR vs IR	0.0122 0.0038	0.0125 0.0038	-0.001 to 0.001	1.000
COP (cm/cm)	VL vs IR	0.0093 0.0028	0.0125 0.0038	0.002 to 0.004	<0.001
	VL vs VR	0.0129 0.0034	0.0181 0.0048	0.004 to 0.007	<0.001
Δ thip (Nm/kg*m)	VR vs IR	0.0181 0.0048	0.0215 0.0045	0.002 to 0.005	<0.001
	VL vs VR	0.190 0.053	0.295 +/- 0.090	0.080 to 0.131	<0.001
Δ tankle (Nm/kg*m)	VR vs IR	0.295 +/- 0.090	0.435 +/- 0.117	0.099 to 0.181	<0.001
	VL vs VR	0.147 +/- 0.036	0.205 +/- 0.056	0.041 to 0.075	<0.001
Δ tshoulder (Nm/kg*m)	VR vs IR	0.205 +/- 0.056	0.248 +/- 0.053	0.021 to 0.064	<0.001
	VL vs VR	0.118 +/- 0.035	0.150 +/- 0.042	0.024 to 0.040	<0.001
	VR vs IR	0.150 +/- 0.042	0.226 +/- 0.080	0.041 to 0.110	<0.001

* Cell entries show mean +/- 1 SD

Table 3.3 Multivariate ANOVA results for two-way interactions with age

<u>Dependant Variable</u>	<u>Interaction</u>	<u>F</u>	<u>p</u>	<u>DOF</u>
COG	Age * Distance	1.339	0.259	1, 24
	Age * Height	1.993	0.171	1, 24
	Age * Condition	1.174	0.318	2, 48
COP	Age * Distance	0.033	0.858	1, 24
	Age * Height	0.005	0.942	1, 24
	Age * Condition	0.365	0.696	2, 48
Δ Thip	Age * Distance	0.096	0.759	1, 24
	Age * Height	2.054	0.165	1, 24
	Age * Condition	0.328	0.722	2, 48
Δ Tankle	Age * Distance	0.035	0.854	1, 24
	Age * Height	0.062	0.805	1, 24
	Age * Condition	0.021	0.979	2, 48
Δ Tshoulder	Age * Distance	0.039	0.845	1, 24
	Age * Height	2.662	0.116	1, 24
	Age * Condition	0.303	0.74	2, 48

TABLE 3.4 Mean ancillary values* seperated by age and group

	Young		Elderly		
	n=13	Group 2A n=10	Group 2B n=5	Average n=15	
Body height (m)	1.676 +/- 0.044	1.581 +/- 0.068	1.536 +/- 0.067	1.566 +/- 0.069**	
Body weight (kg)	57.81 +/- 8.61	59.72 +/- 6.72	67.49 +/- 28.55	62.31 +/- 16.62	
Anterior foot length (cm/BH)	0.111 +/- 0.006	0.117 +/- 0.004	0.111 +/- 0.003†	0.117 +/- 0.004**	
Get up and Go (sec)	9.15 +/- 1.11	11.12 +/- 1.72	13.88 +/- 2.65†	12.04 +/- 2.39**	
Simple upper extremity reaction time (msec)	267.7 +/- 63.6	354.0 +/- 53.0	350.8 +/- 43.2	352.9 +/- 48.0**	
Choice upper extremity reaction time (msec)	395.1 +/- 86.0	484.5 +/- 69.6	515.0 +/- 107.0	495.4 +/- 82.1**	
Simple Lower extremity reaction time (msec)	453.4 +/- 73.0	582.5 +/- 133.6	637.0 +/- 187.4	600.7 +/- 149.1	
Functional reach (cm)	36.40 +/- 3.56	30.60 +/- 3.89	25.20 +/- 6.64	28.80 +/- 5.41**	
RMS ML COP excursion (cm)	4.19 +/- 2.07°	4.85 +/- 2.80°°	5.46 +/- 1.56†	5.06 +/- 2.38°°°	
Dorsiflexor strength (Nm/kg*m)	-	17.45 +/- 3.71	18.59 +/- 4.10	17.83 +/- 3.73	
Hip extensor strength (Nm/kg*m)	75.05 +/- 18.27	73.61 +/- 16.13	74.54 +/- 18.25	73.92 +/- 16.21	
Hip flexor strength (Nm/kg*m)	85.01 +/- 21.43	83.66 +/- 11.21	71.18 +/- 12.89	79.50 +/- 12.86	
Shoulder flexor strength (Nm/kg*m)	22.42 +/- 6.51	21.68 +/- 5.22	19.72 +/- 2.64	21.03 +/- 4.52	

*Cell entries show mean +/- 1SD

**Significantly different from young p<0.05

†Significantly different from Group 2A p<0.05

° n=12 for this variable

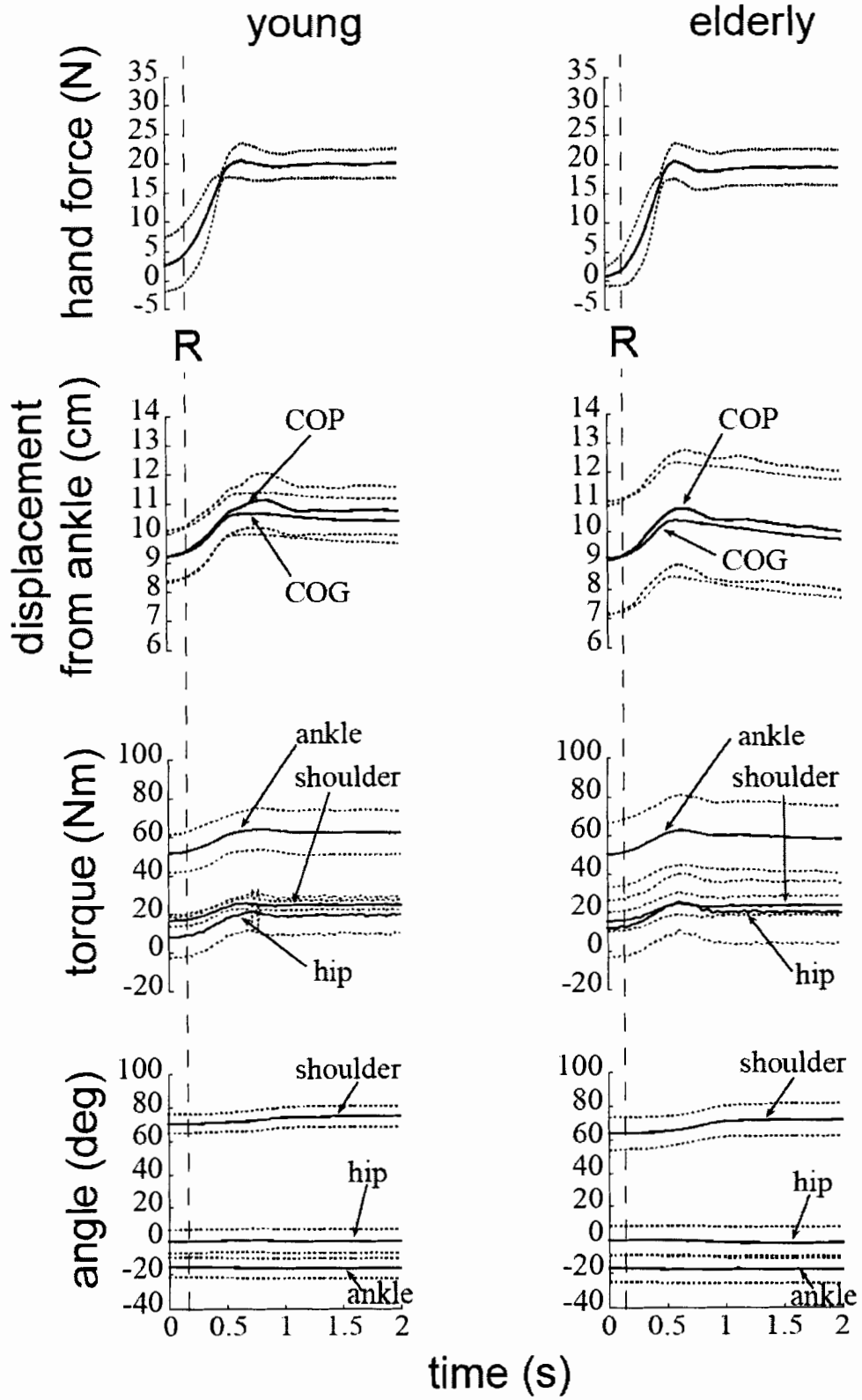
°° n=9 for this variable

°°° n=14 for this variable

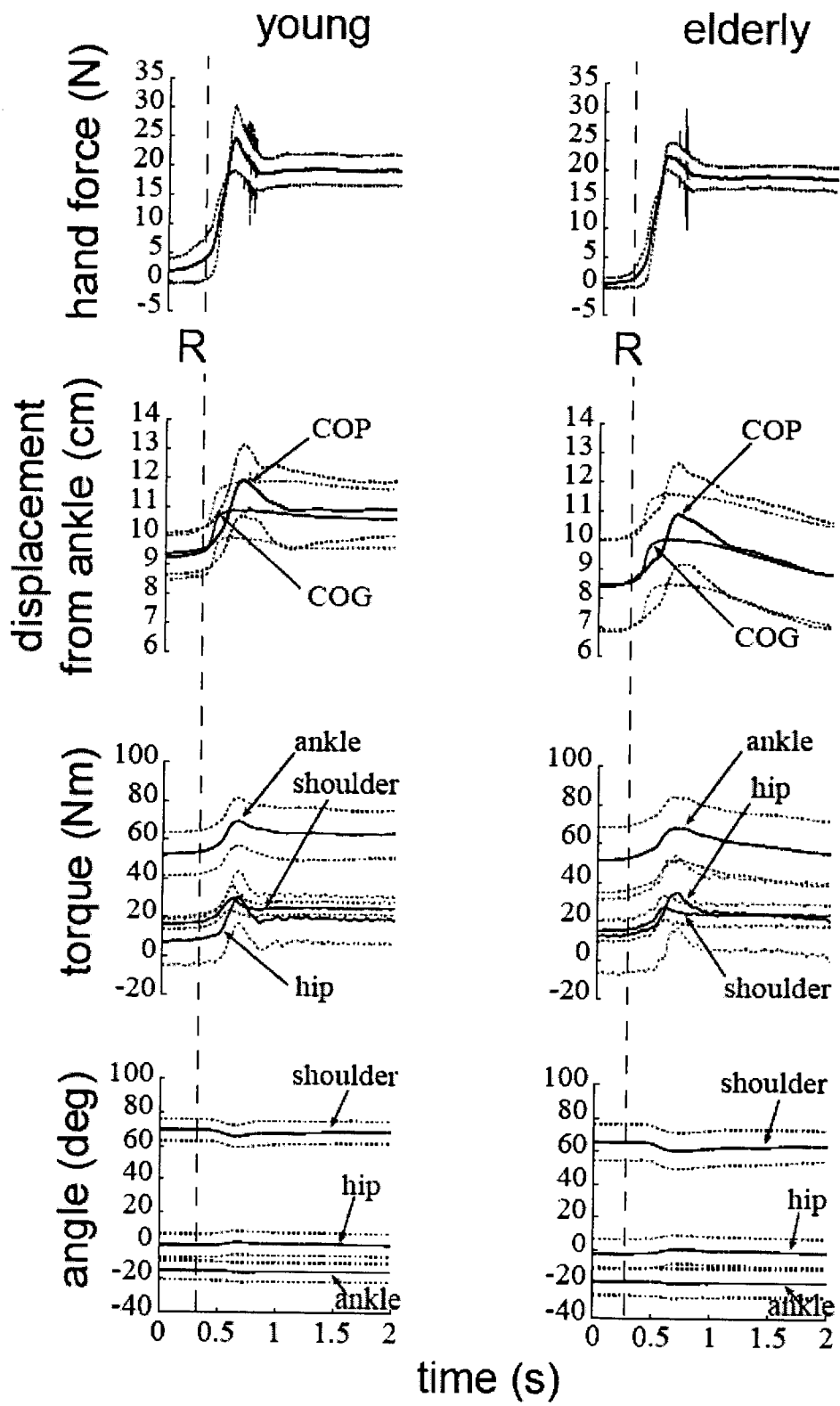
3.7 Figures

Figure 3.1 Composite traces of young versus elderly subjects in **a.** VL trials, **b.** VR trials and **c.** IR trials (see pages 84-86). Composite graphs were generated by synchronizing traces to full weight acceptance and averaging them across all 15 subjects (± 1 SD shown as dotted traces) for each group. Composites are based only on traces generated from trials in the near-medium object position. Few differences are seen between young and elderly. Increases in hand force were accompanied by anterior movement of the COP and the COG, as well as increases in plantarflexor, hip extension and shoulder extension torques. Furthermore, there were ankle dorsiflexion, hip flexion and shoulder extension increases with weight acceptance in VL (**a**) trials and ankle dorsiflexion, hip flexion and shoulder flexion increases with weight acceptance in VR (**b**) and IR (**c**) trials. Variability in traces was greater in elderly compared to young. R indicates moment of tension release.

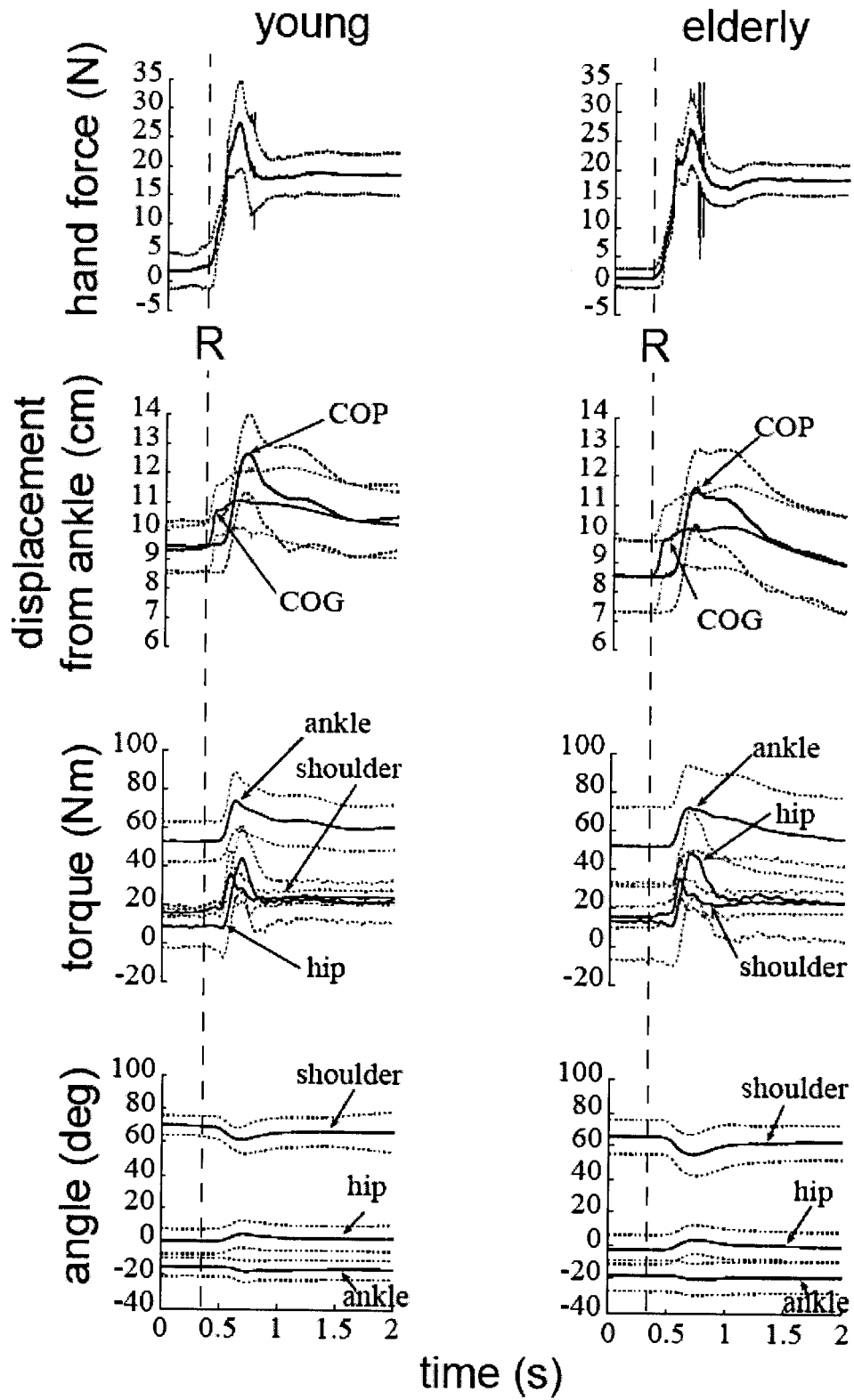
a



b



c



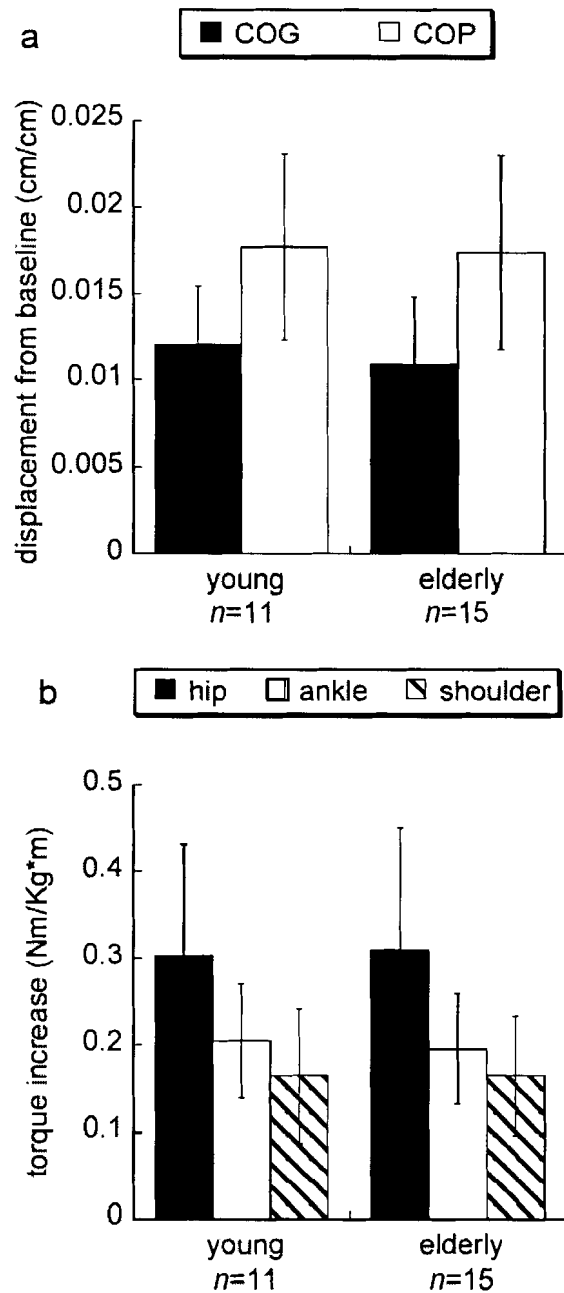


Figure 3.2.a. Effects of age. No significant differences were found in the mean displacement of the COG or COP in young versus elderly individuals. **b.** Furthermore, no significant differences were seen between the joint torque increases experienced by the elderly compared to the younger subjects. Bars represent means and error bars represent +/-1SD.

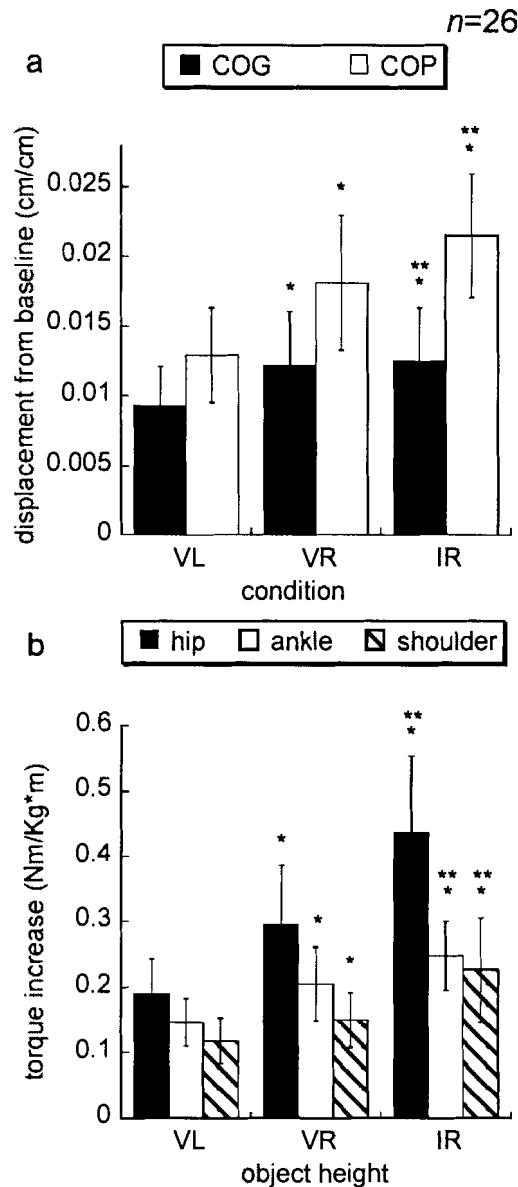


Figure 3.3.a. Effects of weight acceptance condition. **a.** There was 25% less COG displacement in VL than in VR trials but no change in COG displacement between VR and IR trials. Similarly, there was 27.8% decrease in COP displacement in VL than in VR trials and unlike results seen for young only, a further 14.3% decrease in COP displacement in VR than in IR trials. **b.** There were 55.3% greater increases in hip torque, 39.5% greater increases in ankle torque and 27.1% greater increases in shoulder torque from VL to VR trials. There was a further 47.59% greater increase in hip torque, 21.0% greater increase in ankle torque, and 50.7% greater increase in shoulder torque from VR to IR trials. Bars represent means and error bars represent +/-1SD. * indicates significant difference from VL, ** indicates significant difference from VR.

n=26

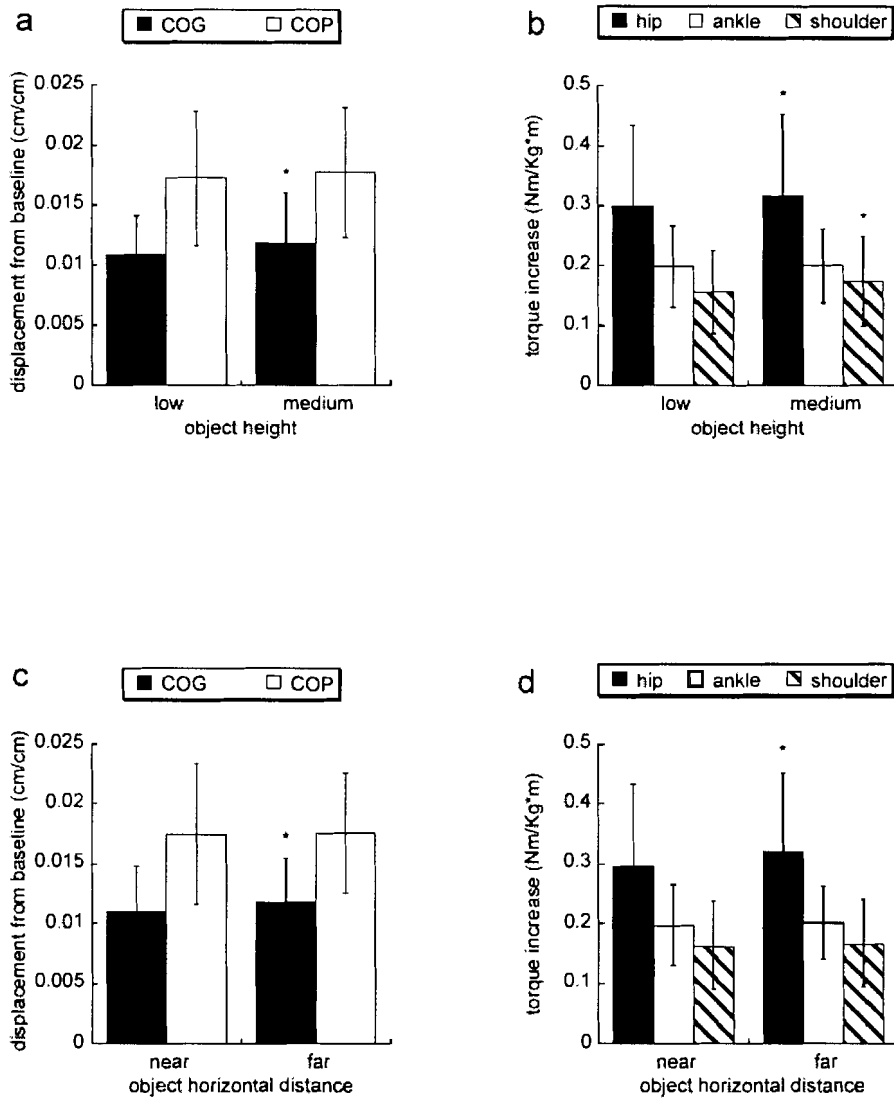


Figure 3.4 Effects of object height and horizontal distance. **a.** There was 8.3% less COG displacement in the low than in the medium object height. However, there was no significant change in COP displacement between the two object heights. **b.** Furthermore, there was a 4.8% decrease in the increases in hip torque and a 10.9% decrease in the increase in shoulder torque in the low compared to the medium object height. There was no significant change in the increase in ankle torque between the two object heights. **c.** Increasing the object's horizontal distance caused a 9.1% increase in COG displacement and **d.** 8.1% greater increases in hip torque. However, a change in the horizontal distance of the object had no significant effect on COP displacement, and the increase in ankle or shoulder torques. Bars represent means and error bars represent ± 1 SD. * indicates significant difference from low (a,b) or near (c,d).

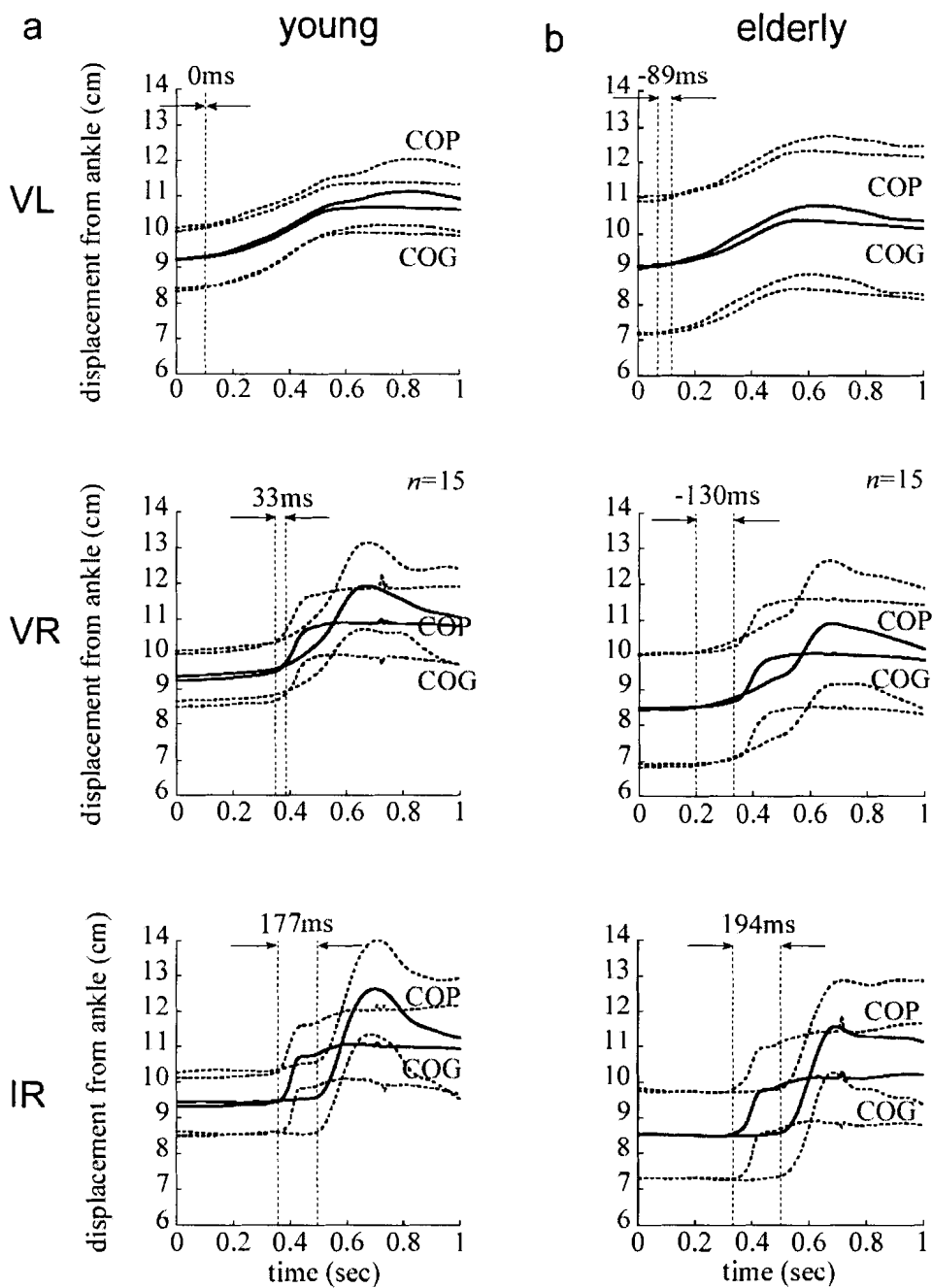


Figure 3.5 On average, elderly subjects experienced different delays in COP onset after COG displacement compared to younger women. **a.** Young subjects experienced synchronous movement of the COP and COG during VL trials, a 33ms delay in COP movement compared to COG movement in VR trials and a 177ms delay in IR trials. **b.** For elderly subjects, the COP was seen to rise 89ms prior to the COG in VL trials, 130ms prior to COG displacement in VR trials and 194ms after COG displacement in IR trials. Traces are produced from composite data. Dotted lines represent ± 1 SD.

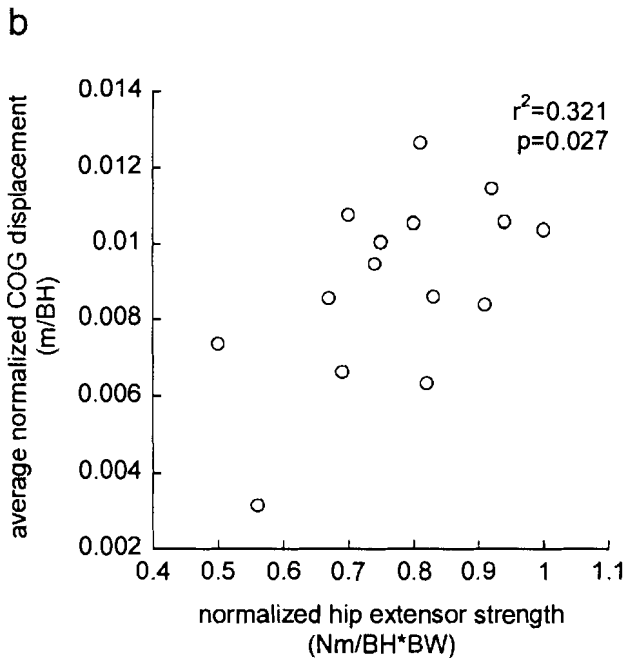
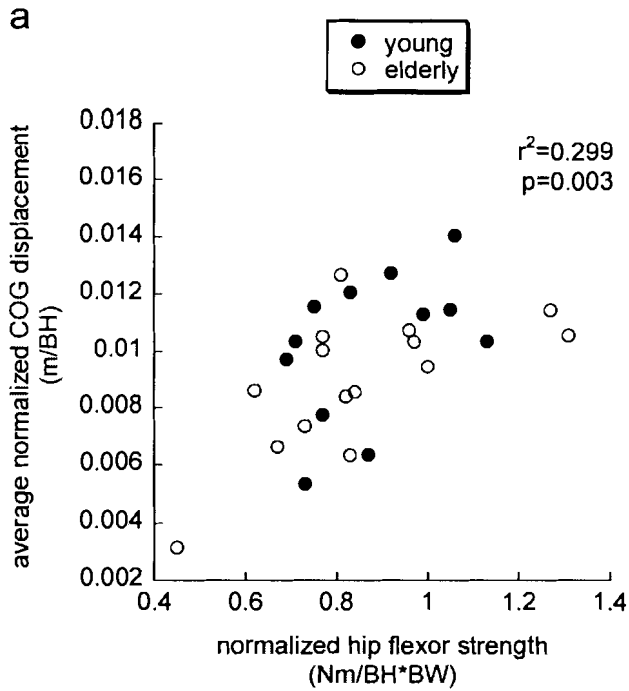


Figure 3.6.a Decreases in hip flexor strength were associated with increases in the average displacement of the COG [normalized to body height (BH)] with weight acceptance across both young and elderly subjects. Furthermore, **b.** decreases in hip extensor strength were correlated with increases in the average displacement of the COG in elderly subjects. Filled circles= young, empty circles=elderly.

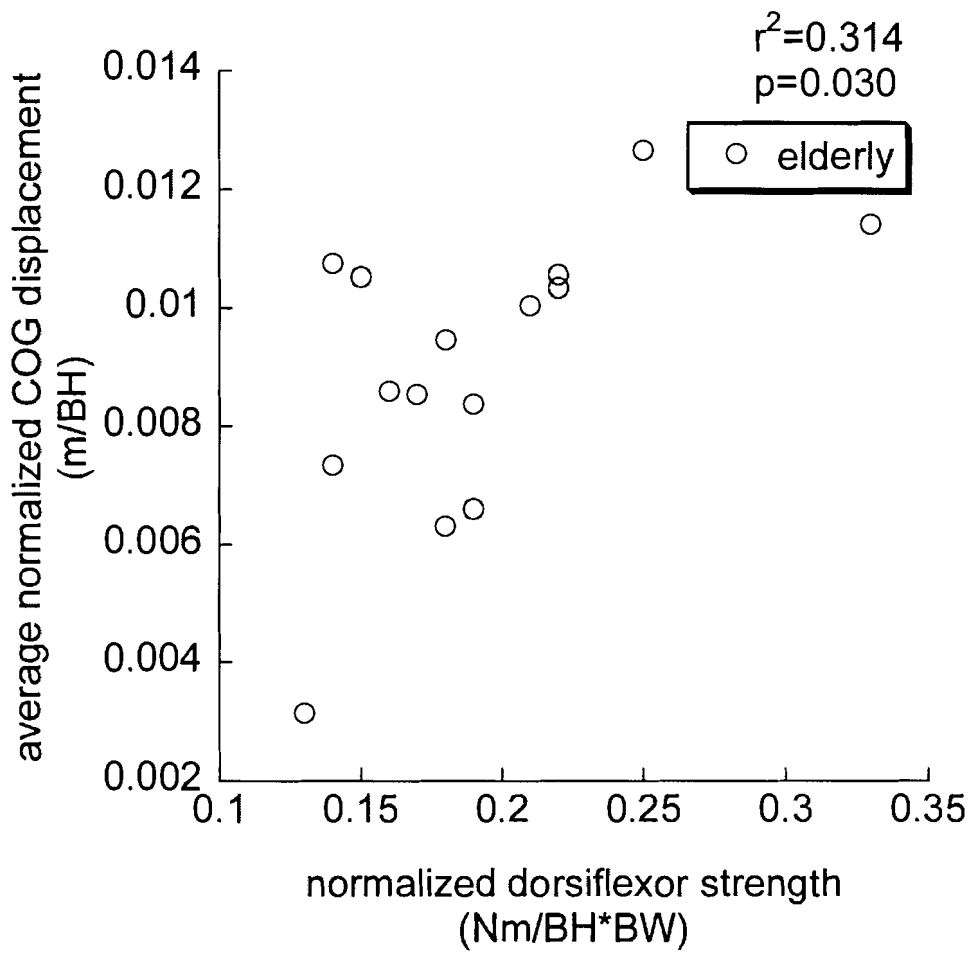


Figure 3.7 Elderly also had a significant correlation between the average displacement of the COG with weight acceptance and dorsiflexor strength. Specifically, increases in dorsiflexor strength were associated with increases in the COG displacement.

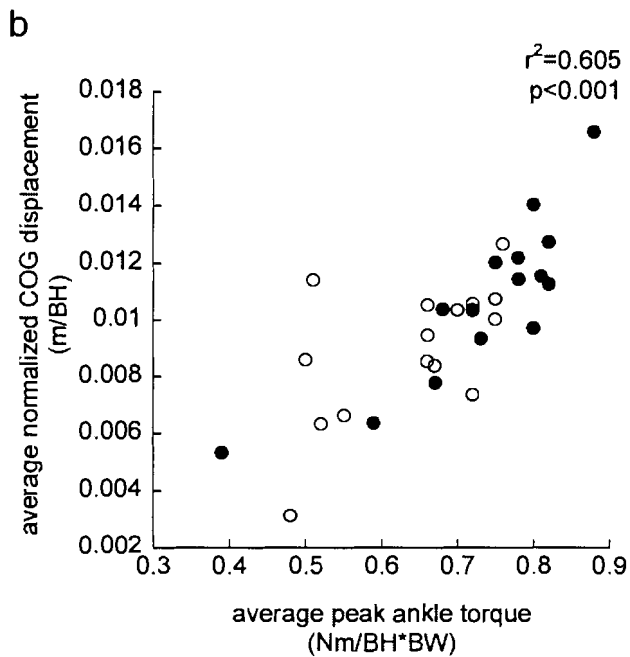
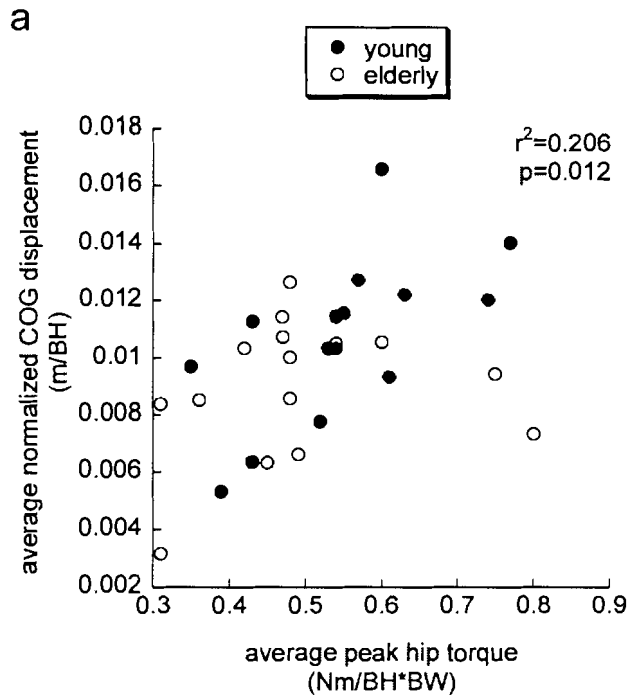


Figure 3.8 The average COG displacement seen during weight acceptance was also influenced by the average peak torque that was generated about the **a.** hip and **b.** ankle. Specifically, greater COG displacement was associated with greater peak hip and ankle torque. Filled circles= young, empty circles=elderly.

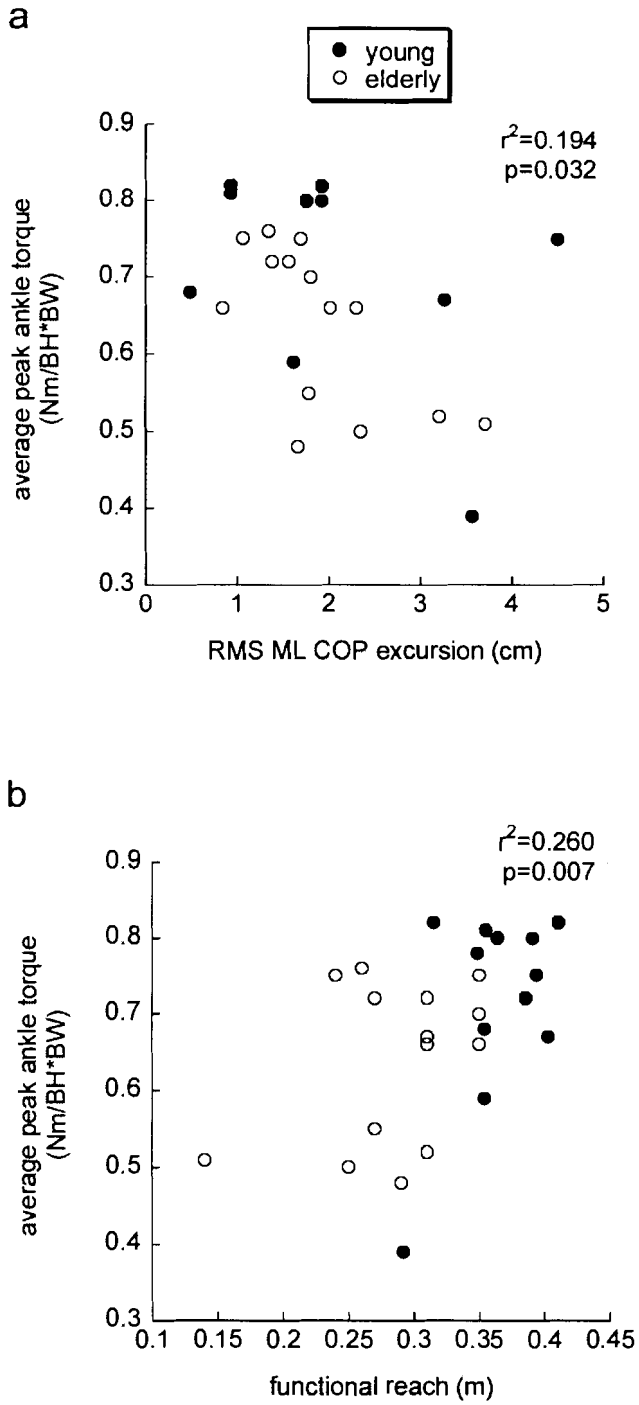


Figure 3.9 The peak torque generated about the ankle was significantly associated with standardized tests of mobility. Specifically, **a.** increases in the mean root squared medial-lateral COP excursion seen during quiet stance were associated with decreases in peak ankle torque. As well, **b.** increases in functional reach were correlated with increases in peak ankle torque. Filled circles= young, empty circles=elderly.

CHAPTER 4: GENERAL DISCUSSION

4.1 Overview: Mechanisms of Postural Control

The research outlined in this thesis suggests that the postural control required to minimize the COG displacement caused by accepting a weight to the outstretched hands depends in part on: the subject's control of the rate and onset of weight acceptance and on the initial position of the weight with respect to the individual.

Postural control during weight acceptance appears to follow similar strategies and depends on analogous factors to those seen in the production of anticipatory postural adjustments (APAs) (Paulignan et al., 1989; Struppler et al., 1993). Support of the body to accept an added weight in the outstretched hands is achieved by controlling the COG within the BOS by the generation of joint torques (and consequently a resultant COP position) to resist the flexor torque caused by the weight acceptance. Similar to previous studies on stability during voluntary movement (Kaminski & Simpkins, 2001; P. J. Stapley et al., 1999; Toussaint et al., 1995), anterior movement of the COP prevents the whole-body COG from moving outside the boundaries of the BOS by balancing whole-body torques about the ankle. Specifically, the extensor torque generated to position the COP is larger than the combined flexor torques generated by the position of the COG and the force at the hand. Positioning of the COP is controlled by the

torque generated about the ankle. Ankle plantarflexor torques, hip extensor torques and shoulder extensor torques minimize flexion of the joint angles and decrease whole-body COG displacement during weight acceptance. On the other hand, as I used a task that allowed for multiple degrees of freedom, there is opportunity for various combinations and magnitudes of these joint rotations. Thus suggesting that the overall movement strategy used may vary slightly between subjects and across different conditions and object positions. For example, some subjects could have minimized object and whole-body displacement by rotating mostly about the shoulder, while others could have rotated more about the ankles and hips in a manner similar to the “axial synergies” seen by Crenna et al (Crenna et al., 1987).

When considering a larger population in Chapter 3, I found that COG displacement was greater in VR than VL trials, but no greater in IR than VR trials. In VR trials subjects did not have control of the rate of weight acceptance and in IR trials subjects did not have control of both rate and onset of weight acceptance. It would therefore appear that the ability to control the onset of the weight acceptance did not influence the subject’s ability to minimized COG movement. However, because movement of the COG did increase between VL and VR trials, the individual’s first means to maintain stability during daily tasks of weight acceptance is most likely due to controlled decreases in the rate of weight acceptance during lifting. Alternatively, when looking at both young and elderly subjects, I found increases in COP displacement across the three different

conditions of weight acceptance as well as greater increases in the joint torques experienced during VR compared to VL trials and IR compared to VR trials. These results indicated that although the subject's ability to minimize the disruption to the COG does not depend as much on control over perturbation onset, the effective use of corrective responses that are initiated once the subjects lost control over weight acceptance rate does depend largely on control over perturbation onset. Once subjects lose control over the onset of weight acceptance, the initiation of corrective responses are delayed and so greater joint torques and COP displacement are required to successfully oppose the now increasing effects of the perturbation.

I found no differences in young versus elderly COP and COG displacements suggesting that my elderly individuals with age-related deficits in mobility were as able as my young to minimize COG displacement during weight acceptance. However, where only two of my young subjects could not complete the task when the object was placed in a far-high position, 10 of my elderly subjects could not. Moreover, a further 5 of these elderly subjects could not complete the task when the object was placed in near-high position, suggesting that some factor not tested in my study limited the functional range of my elderly subjects to accept the weight with little movement of the COG. Perhaps it was that my elderly were employing an alternative strategy from young to complete the weight acceptance task with the same degree of COG displacement. However, this alternative strategy limited the feasible range of body

configurations in which the subjects could successfully minimize COG displacement.

It could be that my elderly subject's used an alternative strategy to minimize COG because their initial stability was reduced compared to my younger subjects. This explanation is substantiated by past research which indicates that elderly women have smaller functional BOS sizes compared to younger subjects (King et al., 1994). Thus, for some of elderly subjects, their COG may have been placed closer to the boundaries of their BOS than for my younger subjects. Perhaps once the COG was within a certain proximity to the boundaries of the BOS an alternative strategy would help ensure that joint torques were not so great as to be outside the strength limits of the individual.

My preliminary examination of the onset of COG and COP displacements suggested that the elderly women may experience greater delays in response time compared to young during involuntary conditions of weight acceptance. This is in agreement with previous research that indicates that elderly have delayed response times compared to young (Williams et al., 1997). Young subjects had a delay in the anterior movement of the COP with respect to COG displacement and weight release in IR trials. Unfortunately, this increase in delay would suggest that elderly would initiate corrective responses later than younger subjects, and would therefore need to adopt their strategy to appropriately respond to the relative increases in the effects of the perturbation.

Previous research has found that during whole-body reaches and lifts, there are increases in the size of the COP and the COG displacements when the reach is farther and when the object to be lifted is heavier (Kaminski & Simpkins, 2001). I found that increases in height and horizontal distance of the object caused some increases in the joint torques and COP position demands associated with adequate postural control. This was expected as increasing the horizontal distance or placing the object in a position that required hip flexion would cause an increase in the moment arm of the torque generated by the force at the hand. This increase in moment arm would demand associated increases in joint torques to oppose the flexor torque generated by the weight at the hand.

Ancillary measures indicated that there were significant decreases in my elderly subjects' performance on standard tests of mobility, such as functional reach and Get-up-and-go time, compared to my younger subjects. However, the impairments did not result in decrements in the subject's ability in minimize COG displacement in the trials that were complete. On the other hand, there were no reductions in the strength of my elderly subjects compared to my young. The lack of differences between my young and elderly subjects could be due to one of two things, either the population of elderly that was selected to participate was too health or the protocol was too limiting. Because there were no differences in the strength measures of young and elderly subjects, I believe that our elderly subject population was too healthy to properly study the effects of age-related deficits on postural stability during weight acceptance. Previous research has

shown that there are significant differences in community dwelling versus nursing home seniors in mobility and strength (Wolinsky, Callahan, Fitzgerald, & Johnson, 1993). Perhaps those that are more frail would have shown signs of reduced control over COG displacement during weight acceptance. Furthermore, there were strong correlations between measures of strength (at the ankle, hip and shoulder) and the displacement of the COG with the perturbation, which suggested that some of my elderly may have adopted alternative strategies to maintain stability to those used by the young subjects.

Despite the fitness of my elderly population, observations made during collection suggested to me that the elderly had more difficulty than the young women performing the task. For example, elderly subjects took longer than young to get into, and get comfortable with, the initial position. Furthermore, elderly subjects had a greater tendency to use the object for support prior to weight acceptance, causing many trials to be discarded and repeated before successful attempts were captured. It would probably be beneficial to do further tests on elderly subjects to see if differences exist when the protocol did not demand a precise movement. Moreover, it would be good to take a closer look at the differences in strategies that are seen between young and elderly women, including muscle activation times and torque propagation patterns.

4.2 Implications of Discoveries

My supplementary analysis of correlations between ancillary measures and performance variables from the weight acceptance protocol (namely COG

displacement and peak joint torques) expanded understanding of previous research on standardized tests of mobility. Specifically, standard tests have been extensively studied to examine their relationship with postural stability. Functional reach, Get-up-and-go and COP excursion during quiet stance are most frequently used to test elderly individual's mobility and postural control because they have been associated with increases in age and frailty (Duncan et al., 1990; Samson et al., 2000; Williams et al., 1997). However, they do not test for the subject's ability to produce adequate torque at specific joints. The present research indicates that there was a significant correlation between the individual's performance on functional stability tests and the peak ankle torque generated during weight acceptance to the outstretched hands. Therefore, as previously suggested testing of an individual's postural stability should also include considerations for strength demands and the individual's ability to generate appropriate joint torques (Bouisset et al., 2000).

These results also have implications for future work completed in the biomechanical field with regards to elderly postural stability. Although some measures of stability may indicate that elderly individuals have similar abilities to minimize body movement during a perturbation as young individuals, they do not identify differences and limitations in the postural strategy used by elderly versus young individuals. It is important that when testing for deficits in postural stability in elderly with age-related deficits in mobility, consideration also be made for the use of alternative, less effective, strategies of postural control due to increased

initial instability or the need to decrease the associated effort at the joints. Further, these alternative strategies may mask impairments in mobility.

The unexpected associations between the displacement of the COG and the strength of the individual, as well as the association between the individuals performance on stability tests compared to the peak ankle torque used during the weight acceptance tasks, both suggest that increases in COG displacement do not necessarily indicate a reduction in stability. Rather, increased stability could result in greater freedom of the COG displacement or greater instability may result in the implementation of specific alternative postural strategies.

The present research also suggests that impairments in strength are not the only deficits that can influence the biomechanical behavior of elderly women. Deficits in response times and other factors, such as range-of-motion, coordination, and endurance, which are also associated with mobility, may also play a role in postural stability control. During the present study the displacement of the COG seen during weight acceptance was associated with performance on standard mobility tests that rely on more than just joint strength. Unfortunately, these ancillary factors were not examined in this study, but would be an important part of future work.

Currently, scientists and industry alike refer to standards outlined by the NIOSH equation to limit the risk of back injury during lifting. The NIOSH equation was first developed by the National Institute for Occupational Safety and Health in 1981 to provide a standardized Recommended Weight Limit (RWL) to a

particular lifting situation. The NIOSH equation has since been revised, taking several different situational factors into consideration (Waters, Putz-Anderson, Garg, & Fine, 1993). Each of these factors is included in the equation with a weighted ratio. These factors which are grouped into Biomechanical, Physiological, and Psychophysical criterion include (but are not limited to); relative position, relative height, weight of the object, frequency of lift, and the hand grasp on the box (Waters et al., 1993). Although, this equation has been found to be helpful in setting standards to reduce back injury during lifting, it does not consider the risk of lose of balance episodes. The current thesis supports the notion that weight acceptance is a demanding task for individuals that could increase the chances of imbalance. Therefore, I think that based on this work, it should be recommended that a similar equation limit the risk of imbalance should be produced. In particular, this equation should consider, the object's mass, position, the rate of lifting, the ability of the individual to predict and prepare for the acceptance (amount of anticipation), and the individuals strength and mobility. The present study indicates that these factors play an important role in the ability of the individual to control their COG displacement when accepting an additional weight. Many elderly individuals have been found to lose balance while lifting an object off a high shelf, causing them to fall and sometimes break a hip. Although many more elderly are becoming aware that balance is impaired at their age, they are unaware of the degree of this deficit and consequently do not know how or when they should reduce their load acceptance. If an equation to

limit risk of imbalance during weight acceptance was generated and included the factors outlined above, then more specific suggestions could be provided to the elderly community as to what lifts they could safely perform, and which should be avoided. Furthermore, appropriate standards could be set for the design of shelving in senior's complexes and other locations of which seniors are known to frequent. For example, lowering shelves would not only decrease the initial instability but would also allow visualization of the object to be lifted. This visualization would help them to control the rate and onset of the load acceptance.

4.3 Limitations

As mentioned in Sections 2.5 and 3.5, due to the nature of the study protocol, certain constraints and methodologies used to answer the hypotheses may have presented themselves as limitations that should be considered when examining the results. New and general limitations have been (re-)stated below.

First, several different studies have shown that lightly touching an object with the fingertips improves postural stability during standing (Krishnamoorthy, Slijper, & Latash, 2002; Rabin, Bortolami, DiZio, & Lackner, 1999). Therefore, by holding the object handlebars prior to weight acceptance, it is possible that the postural stability of my subjects were better than had they had no contact with the object prior to weight acceptance. However, for the purposes of this study I wished to ensure a consistent initial body configuration between trials. Preliminary trials indicated that when lifting to accept a weight, the object is first

grasped and then lifted. Therefore the individual would have the opportunity to obtain fingertip contact and increase their stability prior to weight acceptance. Finally, each trial would have had the subject start out with the same degree of assistance from the handlebar because every trial began with the subject's hands lightly touching the handlebars.

As stated before, a further limitation to the study was the fact that we could not create a situation where the subject was unaware of the pending weight acceptance. Instead, I modified the subject's control over the rate and timing of the weight acceptance. However, previous research has shown that even if the subject is expecting the perturbation to the upper extremity, they cannot limit the movement of the arm as well when the perturbation is investigator-induced versus self-induced (Johansson & Westling, 1988a, 1988b; Westwood et al., 2000). Instead, precise timing of the perturbation onset and knowledge of the perturbation magnitude is required to reduce the effects of the disturbance on the arm.

Third, subjects began trials pending weight acceptance eliminating our ability to measure the biomechanics of the transfer component. However, although I recognized that this may not duplicate daily tasks of weight acceptance, this initial position helped to control the body configuration adopted for different trials and by different subjects. Future studies could help to complete the biomechanical picture of weight acceptance to the outstretched hand by including this transfer component.

Fourth, considerations for the sudden additional weight were made during the calculation of the individual's COG. Commissaris et al. studied APAs generated for the lifting of a load and included the additional weight in their COG calculations in 1997 (Commissaris & Toussaint, 1997a, 1997b) but removed them in a later experiment (Commissaris et al., 2001) in order to examine the effects of the box as a force experienced by the hand and not as another body segment. We chose not to separate the whole-body COG from the weight of the object because we wished to understand the subject's ability to manipulate body segments, as well as the weight, in order to minimize the disruption caused by the perturbation to the whole-body COG. Further, by including the weight of the object in the COG calculations, we were able to get the full picture of how the COG related to the COP position.

Fifth, due to the nature of the protocol, the vertical direction of the object displacement during VL was opposite to that of the VR and IR trials. More specifically, during lifting trials (VL) the object was displaced upwards, whereas during release or catching trials (as in VR and IR trials) the object was displaced downwards slightly. However, we attempted to minimize this difference by instructing the subjects to minimize the displacement of the object and the body as a whole during weight acceptance. Furthermore, we focused our analysis on the horizontal displacement of the COG and COP displacement to limit the effect that the vertical object displacement had on comparisons between trials types.

Finally, and most importantly, the results of this study can only be applied to the biomechanics of healthy young and elderly women and not males. Moreover, these results can only be generalized to elderly women that reside in the community and are completely self-sufficient as the participants in this study were. These subject populations were chosen for specific reasons. First, only women were chosen as participants because previous research has shown that postural stability is different in men and women due to variations in anthropometrics and neuromuscular variables (Chow et al., 2000; Farenc, Rougier, & Berger, 2003). In addition, elderly women are at a much greater risk for falls compared to elderly men (Campbell et al., 1989; Nevitt et al., 1989; Tinetti et al., 1988), suggesting that perhaps understanding the biomechanics behind the movements employed by females are essential to decrease the incidence of falls.

I did not recruit elderly women who reside in nursing homes because nursing home elderly have been shown to have greater deficits in mobility and neuromuscular variables compared to community dwelling elderly (Wolinsky et al., 1993). Therefore, I felt it was more appropriate to complete the research on the healthier of the two groups to see if any differences existed where deficits were not that great. Furthermore, many nursing home residents have additional medical complications (i.e. Parkinson's, mental disorders, hyper/hypotension etc.) that further decrease their postural stability in comparison to that of the general aged population (Campbell et al., 1989). Unfortunately, because we

selected a relatively healthy and mobile population of elderly participants we ran the risk of seeing relatively few differences from younger subjects. However, despite this limitation, my results still provided evidence that the elderly women used alternative strategies to the young to maintain their COG positions. And furthermore, correlations between performance measures and basic strength measures suggested that greater changes in stability and postural strategy would have been seen if I had used a more frail elderly population (i.e. reduced hip, ankle and shoulder extensor strength).

4.4 Future Directions

The current research provides new information regarding postural control during weight acceptance and lends support to past research concerning the role of anticipatory responses in the ability to minimize perturbation effects on movement. However, it also brings up many new questions and possibilities for further inquiry. For example, how is control of perturbation onset and rate used to manipulate torque development and COP movement? Limiting the control of the rate of weight acceptance appeared to increase the magnitude of the disturbance to the COG, as well as increase the demands on the joint torques and COP movement. On the other hand, limiting the control of the perturbation onset only had an effect on the joint torque demands and COP positioning associated with postural control. It has been theorized, and much support has been found, that postural control is attained through anticipatory responses that are initiated by the motor control system and based on proprioceptive feedback

and the biomechanical requirements of the intended movement (Kandel et al., 1991; Nashner & McCollum, 1985). However, future studies could help us understand how the perturbation onset and rate are incorporated into these organized anticipatory responses. Furthermore, more studies could examine how control and knowledge over other characteristics of the perturbation (i.e. duration, intensity) is used to elicit anticipatory responses.

Second, what differences in muscle activation patterns are seen with self-initiated versus investigator-induced or even unexpected weight acceptance? My preliminary examination of COP and COG displacement onset suggested that there may have been greater delays in COP responses in elderly compared to younger women. Variations in COP initiation suggest differences in postural muscle activation timing. Variations in muscle activation could indicate a change in postural strategy which may influence the propagation of the torques created about the ankle, hip and shoulder joints. Muscle activation patterns are believed to be controlled through motor control centers located in the central nervous system (Kandel et al., 1991). Thus, an increased understanding of muscle activation patterns, and therefore a paralleled increased understanding of torque development, would help define the role of higher level controls systems. Moreover, the knowledge gained from this study could also be strengthened through greater understanding of the role that co-contraction plays in both the upper and lower extremities during weight acceptance to the outstretched hands.

Third, is the lack of differences seen between young and elderly due to a healthy elderly population or to constraints (related to the protocol) on the postural control strategies? Although few differences were seen in the postural stability of young versus elderly female subjects in trials that were completed, the results and supplementary analysis suggest that differences may arise with a more frail population. In addition, further understanding of various age-related deficits on postural stability during weight acceptance to the outstretched hands could be obtained by using populations with neuromuscular deficits or disorders (including fallers, nursing-home residents or Parkinsonian patients), which could lend insight into how postural control normally relies on these neuromuscular components.

Finally, how do the same changes in the conditions of weight acceptance (i.e. differences in anticipation) affect the postural stability of the whole-body during other less constrained tasks which involve weight acceptance? I chose to examine the postural control associated with weight acceptance alone, and not the transfer components normally seen in daily activities that include weight acceptance. Furthermore, I chose to examine this weight acceptance when the subjects was standing upright and reaching forward. Similar studies could be conducted to understand how strategies vary when the reaching component is included, when the weight acceptance occurs at the side of the body, or even when the weight is only accepted with one hand.

4.5 Conclusion

The studies in this thesis have shown that the ability to minimize the disruption to the COG during weight acceptance in the outstretched hands depends on the individual's ability to anticipate and control the rate and onset of weight acceptance and on the initial position of the object with respect to the individual. More specifically, the strategies employed to reduce the effects of the perturbation on the whole-body COG vary with age-related deficits in mobility and depend, in part, on the individual's need to either reducing the effort demanded at joints or increasing the range of object locations in which stability can still be maintained. Unfortunately, the use of a health elderly subject population prevented detection of age-related deficits on ability to maintain stability during weight acceptance. However, correlations between performance on the task and standard tests of mobility, postural stability and strength indicate that differences could have been seen in a frailer population. Future work will aim to expand the biomechanical understanding and the motor control contributions to weight acceptance, as well as the abilities and performances of other populations and related tasks.

APPENDIX A

SIMON FRASER UNIVERSITY

INFORMED CONSENT BY SUBJECTS TO PARTICIPATE

IN A RESEARCH PROJECT OR EXPERIMENT

The University and those conducting this project subscribe to the ethical conduct of research and to the protection at all times of the interests, comfort, and safety of subjects. This form and the information it contains are given to you for your own protection and full understanding of the procedures. Your signature on this form will signify that you have received a document which describes the procedures, possible risks, and benefits of this research project, that you have received an adequate opportunity to consider the information in the document, and that you voluntarily agree to participate in the project.

Having been asked by Stephen Robinovitch, Ph.D. of the School of Kinesiology of Simon Fraser University to participate in a research project experiment, I have read the procedures specified in the attached document "Information Sheet for Subjects".

I understand the procedures to be used in this experiment and the personal risks to me in taking part.

I understand that I may withdraw my participation in this experiment at any time.

I also understand that I may register any complaint I might have about the experiment with the researcher named above or with Dr. John Dickinson, Director, School of Kinesiology at Simon Fraser University.

I may obtain copies of the results of this study, upon its completion, by contacting: Stephen Robinovitch, Ph.D., School of Kinesiology, Simon Fraser University, Burnaby, B.C.; telephone: 604-291-3566.

I have been informed that the research material will be held confidential by the Principal Investigator.

I understand that my supervisor or employer may require me to obtain his or her permission prior to my participation in a study such as this.

I agree to participate in an experimental session of approximately 5 hours with a 30 minute break, at a time to be arranged. The session will be carried out at the Centre for Injury Prevention and Mobility at Simon Fraser University.

NAME (please type or print legibly): _____

ADDRESS: _____

SIGNATURE: _____ WITNESS: _____

DATE: _____

ONCE SIGNED, A COPY OF THIS CONSENT FORM AND A SUBJECT FEEDBACK FORM SHOULD BE PROVIDED TO THE SUBJECT.

Information Sheet for Subjects

Purpose and Background: Stephen Robinovitch, Ph.D., in the School of Kinesiology at Simon Fraser University is conducting a research study to help understand movement strategy during reaching **and weight acceptance**. I am being asked to participate in this study because I meet the inclusion criteria of the study.

Procedures: If I agree to be in the study, the following will occur:

On the first day of the experiment, I will come to the Injury Prevention and Mobility Laboratory at Simon Fraser University, and change my clothes into a light cotton shirt and short pants. My height, weight, and leg length will then be measured. My arm and leg strength will then be measured by having me push as hard as I can against a padded surface. My arm and leg flexibility will then be measured by having me flex and extend each joint as much as I can. My "reaction time" will then be measured by having me press a button with my foot or finger as quickly as possible after I hear a beep. My balance will then be measured while I stand on a rigid platform. My depth perception will be tested using a depth perception apparatus. I will then complete a questionnaire regarding my medical history and status. The estimated time required for this first session is three hours.

On the second day of the experiment, I will again come to the Injury Prevention and Mobility Laboratory at Simon Fraser University. Foam markers will be taped to the skin overlying my feet, knees, hips, shoulders, arms, and top of my head. Sensors that monitor my muscle activity may be placed on some of my leg and trunk muscles. For the first part, I will participate in a series of reaching trials watching a target in front of me. Once I hear a beep, I will attempt to contact the target with my dominant hand at either a natural pace or as quickly as possible. In some trials, the target will be moving while in others, it will be stationary. For the second part, I will participate in trials where I will be instructed to walk or run from a starting location to one or more stations in the laboratory, where I must grasp targets with one or both hands. Between trials, I will be given a rest break of approximately 30 seconds. In some trials, I may be required to wear a harness that will be attached to a tether. This tether will normally be slack, and will not impair my movements. However, it will catch me if I accidentally lose my balance.

For an additional experiment, I will perform a series of reaching and lifting tasks. During these trials, I will be asked to reach forward and either grasp or lift an object, while maintaining my balance and keeping my feet stationary on the ground. The mass of the object will be between 10 grams and 3 kg. The horizontal distance between the object and my feet may vary between 10 cm and my maximum forward reach distance), and the vertical height of the object may vary between ground height and just above the height of

my head. Also, I may be asked to conduct three types of trials. During the voluntary lifting trials, I will be asked to reach forward and voluntarily lift the object. During voluntary release trials, I will reach forward and lightly grasp the object. At my own discretion I will be asked to release the support, by pressing a fingertip button, to accept the weight in my outstretched hands. During the unexpected trials, I will reach forward and lightly grasp the object, which is initially supported by a rope. After a short delay, the investigator will release the support, and my task will be to accept the weight of the object and hold it stationary. I will take several breaks between trials to minimize my risk for discomfort or muscle strain. The time required for the entire session should be no more than 3 hours.

Risks and Discomforts: There are few risks and discomforts associated with the experiment. However, I may feel fatigued at some point during the session. **I may also experience some muscle strain in my trunk or limbs.** If this occurs, we can stop the experiment to take a break or end the testing.

Benefits: These studies are being performed to advance our knowledge of reaching behavior in humans. They are not intended to diagnose or reduce my own risk for disease or injury.

Cost/Payment: In return for my time as a participant in this study, I will receive \$10 per hour, starting from the moment I arrive at the Laboratory to the moment I depart the Centre. The payment will be sent to me approximately three weeks after the experiment, in the form of a check.

Questions: This study has been explained to me by Dr. Robinovitch or a co-investigator and my questions were answered. If I have any other questions about the study, I may call Dr. Robinovitch at (604) 291-3566.

If I have any comments or concerns about participation in this study, I should first talk with the researchers. If for some reason I do not wish to do this, I may contact Dr. John Dickinson, Director of the School of Kinesiology. I may reach this office between 9:00 AM and 5:00 PM, Monday through Friday, by calling (604) 291-4062.

Confidentiality: My confidentiality will be strictly maintained by the study investigators. The results of my test session may be included with results from other studies and be reported in a scientific meeting and/or journal. However, my name or likeness will not be revealed in any published reports about this study.

IF I WISH TO PARTICIPATE, I SHOULD SIGN THE CONSENT FORM.

PARTICIPATION IN THIS STUDY IS VOLUNTARY AND I AM FREE TO WITHDRAW AT ANY TIME WITHOUT PENALTY.

APPENDIX B

NATURAL COP LOCATION WITH REACHING

Purpose

The purpose of this sub-study was to generate appropriate COP positions to be used as target COP locations during the weight acceptance reaching protocol.

Methods

Eight young participants [mean age (SD): 26.13+/-3.80yrs] were recruited to complete the sub-study. Of these eight subjects, 3 were female and 5 were male.

Participants stood on a force plate with the object suspended on a rope in front of them. Subjects were asked to bend forward and lightly wrap their hands around the handlebars of the object and were further instructed to bend only at the hips and ankles while keeping their knees straight. Subjects were informed that they were allowed to raise their heels off the ground but were not permitted to take a step. The position of the object varied in height and horizontal distance using the same combinations used in the weight acceptance reaching protocol (low, med, high heights and far and near distances). Subjects were instructed (and monitored) not to apply pressure to the object. They were also informed that they would not be accepting the weight of the object at any time and to

remain relaxed. No feedback was provided to the subject regarding the position of their COP. Trials were repeated three times in a random order. Force plate data was then collected for 5 seconds.

Force plate data collected for each of the trials was run through MATLAB programming (The MathWorks, Natick, MA, USA) to calculate the COP from the ground reaction forces ($COP = My / Fz$). The resultant COP were averaged across trials and subjects for each of the relevant positions and normalized to body height.

Results

Results are presented in the table B.1 and used to set participant target COP from real-time force plate data during weight acceptance protocol collection.

Table B.1 Mean COP(m/BH) for a given position.

		Low 60%BH	Medium 80%BH	High 100%BH
Near	45%BH	0.059	0.056	0.071
Far	50%BH	0.062	0.065	0.072

APPENDIX C

Dorsiflexion Strength

Only elderly women were tested for dorsiflexion strength, using the standardized method outlined in Stephen Lord's *Fall Screen Physiological Profile Assessment Testing Instructions Manual* (S. Lord, 2003; S. R. Lord, Menz, & Tiedemann, 2003). The elderly participant sat in a chair (height 43cm) with her arms across her chest and her dominant foot place on a slanted surface, which was attached underneath to a vertical force scale. A strap was secured around her fifth metatarsal and her ankle was lined up with a bolt (about which the surface pivoted). The subject's knees were flexed between 120° and 130° and her ankles dorsiflexed at approximately 100°. The elderly woman was asked to flex her foot towards her shins, against the resistance of the force scale, until she was told to relax (approximately 10s). Maximum force pulled on the force scale was recorded. The subject completed the test three times and the average was recorded.

Hip Flexion Strength

The participant stood facing away from a wall with her hands lightly resting on the back of a chair for added stability. An inextensible tether was connected in series with a load cell (capture rate = 960Hz, models 31, Honeywell Sensotec Inc., Columbus, OH, USA) and secured at one end around the ankle of the participant's dominant leg. The other end of the tether was attached to the wall.

The subject was instructed to apply maximal force against the tether by flexing her leg, while keeping her knees straight, when cued by the investigator. The hip flexion was held for 10s. The subject was further instructed to remain standing upright, using the chair for balance but not support, and to prevent herself from leaning into the chair for extra strength. The subject completed the test three times and the average was recorded.

Hip Extension Strength

The participant stood facing a wall with her hands lightly resting on the back of a chair for added stability. An inextensible tether was connected in series with a load cell (capture rate = 960Hz, models 31, Honeywell Sensotec Inc., Columbus, OH, USA) and secured at one end around the ankle of the participant's dominant leg. The other end of the tether was attached to the wall. The subject was instructed to apply maximal force against the tether by extending her leg, while keeping her knees straight, when cued by the investigator. The hip extension was held for 10s. The subject was further instructed to remain standing upright, using the chair for balance but not support, and to prevent herself from leaning into the chair for extra strength. The subject completed the test three times and the average was recorded.

Shoulder Extension Strength

The participant sat in a standard chair with her dominant arm extended at 90° in front of her, palm down. An inextensible tether was connected in series with a load cell (capture rate = 960Hz, models 31, Honeywell Sensotec Inc.,

Columbus, OH, USA) and secured at one end around the palm of the extended hand. The other end of the tether was attached to the floor directly below the subject's hand. The participant was instructed to apply maximal force against the tether by extending her arm, while keeping the elbow and wrists straight, when cued by the investigator. The shoulder extension was held for 10s. The subject was further instructed to keep her other hand on her lap. The subject completed the test three times and the average was recorded.

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