BIOMECHANICS OF POSTURAL STABILITY IN THE ELDERLY

by

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Abstract

Falls cause substantial death and morbidity in the elderly. Fall risk depends on ability to maintain balance during daily activities and ability to recover balance following a perturbation such as a slip or trip. To guide the design of fall prevention programs, we need an improved understanding of the biomechanical variables that govern ability to recover balance. The aims of this thesis were to determine (1) the relative importance of strength versus speed-of-response variables in explaining age differences in balance recovery performance with the ankle strategy, and (2) the association between variables related to ability to recover balance and variables related to ability to maintain balance.

To address Aim 1, young and elderly women were supported in a forward leaning position by a horizontal tether and instructed to recover an upright vertical stance by contracting their ankle muscles. The maximum initial lean angle where they could recover balance without release of the tether (which depends primarily on strength) was 19.6% smaller for elderly than young. The maximum initial lean angle where they could recover balance after the tether was suddenly released (which depends on strength and speed-of-response) was 36.1% smaller for elderly. Moreover, between-group differences in performance were related to both strength and speed-of-response. Peak ankle torque was 7.7% smaller in elderly than young during tether release trials, reaction time was 27% slower in elderly, due to a lengthened muscle response latency, and rate of ankle torque generation was 15.6% slower in elderly. These results suggest that exercise-based

fall prevention programs should include balance and agility training, in addition to strength training.

To address Aim 2, the same elderly subjects participated in postural steadiness experiments, where the amplitude, velocity, and frequency of their centre-of-pressure displacement were measured during quiet stance. Postural steadiness during quiet stance and ability to recover balance with the ankle strategy were not associated, perhaps because postural steadiness during quiet stance is controlled partly by anticipatory strategies, while balance recovery following a perturbation is governed by reactive strategies. These results support the need to measure both balance recovery and postural steadiness in balance assessments of the elderly.

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Table of Contents

List of Figures

List of Tables

Chapter 1
General Introduction and Literature Review

1.1 Introduction

One-third of community-dwelling elderly, and one-half of nursing home elderly, will fall at least once per year (1-3). Falls and their related injuries are a major source of morbidity and mortality among elderly individuals. Falls are the number one cause of accidental injury, and the number-two cause of accidental death in the elderly (4), and they account for 84% of injury-related hospitalizations and approximately 40% of nursing home admissions (5). Moreover, in 1995, fall-related injuries cost Canadians $3.6 billion (6). The most serious type of fall-related injury in terms of numbers, cost, and morbidity is hip fracture. Over 23,000 hip fractures occur annually in Canada, at an estimated cost of $1.0 billion (7,8), while in the United States, falls lead to 300,000 hip fractures per year, at an estimated cost of US$10 billion. Without improvements in prevention, hip fracture incidence is expected to increase 4-fold by the year 2041, given the aging of the population and the fact that risk for hip fracture increases exponentially with age (7,8). Falls are also a major cause of upper extremity fractures, head injuries, and vertebral fractures in the elderly (4). Consequently, reducing the incidence and severity of fall-related injuries has been identified as a critical national health priority (6).

This thesis provides improved understanding of the cause and prevention of falls and their related injuries through biomechanical research techniques. The results substantially enhance our understanding of the biomechanical and neuromuscular

variables that govern age-related declines in postural stability, and will ultimately improve our ability to develop fall prevention interventions based on exercise training and rehabilitation.

1.2 Literature Review

1.2.1 Fall Risk Factors

An elderly individual's risk for falls associates with a variety of sensory, motor, cognitive, and psychosocial variables (9). These include impairments in muscle strength, joint movement, proprioception, tactile sensation, reaction time, balance, gait, vision, cognition, neurological disease such as Parkinson's and stroke, and medication use (2,9-11). These and other behavioural and environmental variables impact on two ultimate determinants of fall risk: 1) one's frequency of loss-of-balance episodes, and 2) one's ability to recover balance after such events (12). Accordingly, fall prevention programs should target each of these two areas. By identifying the mechanisms responsible for age-related differences in balance recovery ability, and quantifying the relationship between balance recovery ability and risk for loss-of-balance, the current research accomplishes an essential step towards this goal.

1.2.2 Postural Control

To maintain postural control, the central nervous system must integrate a wide array of sensory information from visual, vestibular, and somatosensory inputs and produce appropriate motor commands that direct coordinated patterns of muscle activities to keep the body centre-of-mass (COM) positioned over the base-of-support (BOS) (13,14). The tightly controlled relationship between the COM and the BOS is mediated

via two pathways: 1) the COM position is controlled through feet-in-place strategies (ankle and hip) that involve torque generation in the lower extremity joints and the trunk; and 2) the BOS is altered through change-in-support strategies such as stepping or grasping (13). External perturbations can either displace the COM beyond the BOS (as occurs during a trip or push), or prevent the BOS from being positioned underneath the COM (as occurs during a slip). In the event of a perturbation, if the postural control system is unable to compensate through a feet-in-place or change-in-support strategy, a fall will result (13). The efficacy of these strategies depends in turn on variables related to strength and speed of response to loss of balance. The current thesis provides important new insight on the role of strength versus speed-of-response variables as a cause of age-related declines in postural control. The results will be imperative to guide the direction of exercise-based fall prevention programs since different interventions may be required to target each of these variables.

1.2.3 Balance Recovery

1.2.3.1 Balance Recovery Strategies

The frequency and severity of falls increases with age, and women are affected to a greater extent than men (3). Approximately 40-50% of falls in community-dwelling elderly women occur as the result of a trip or slip (15). An additional 20% occur from loss-of-balance (15). This suggests that the high frequency of falling among elderly individuals is due in part to decreased ability to produce effective balance recovery responses when faced with unexpected perturbations.

Four main types of balance recovery strategies have been described for correction of anterior-posterior perturbations. They are 1) the ankle strategy, 2) the hip strategy; 3)

the stepping or stumbling strategy; and 4) the grasping strategy. The first two strategies are commonly called "feet-in-place" or "fixed support" strategies as they involve regulating movement of the body's COM with respect to a fixed BOS (16). The latter two strategies are termed "change-in-support" strategies as the body's COM is controlled through a change in BOS (16).

The specific balance recovery response that an individual produces depends on the magnitude of the perturbation and the characteristics of the support surface, as well as the individual's age, physical abilities, and perception of her resources in relation to the demands of the situation (14). The ankle strategy is the most commonly used strategy when the perturbation is small in magnitude and slow in velocity, and when the support surface is firm and wide and therefore capable of resisting ankle rotational torques (14). This strategy shifts the body's COM by rotating the body about the ankle joints while the hip and knee remain extended. In contrast, the hip strategy involves flexion or extension at the hips, and the stepping and grasping strategies involve repositioning the BOS under the COM. The hip and stepping strategies are generally recruited, in order, as the magnitude and speed of perturbation increase and the support surface decreases in size or stiffness. Proper execution of any balance recovery strategy depends on the ability to generate an adequate magnitude of stabilizing torque in a fixed amount of time and requires the correct temporal relationships between onsets of muscle activities. For instance, in the ankle strategy there is a typical distal-to-proximal activation of ankle, thigh, and trunk muscles, and delays as small as 20 msec in the onset of ankle muscle activity have been shown to result in destabilization (14).

This thesis examined age-related differences in balance recovery ability with the ankle strategy. The ankle strategy is an attractive model for examining such differences because of its relatively simple biomechanical characteristics: movement, and therefore muscle activities and torques, occur primarily about the ankle joints. This allows for direct examination and measurement of the contribution of strength versus speed-of-response to age-related declines in balance recovery ability.

1.2.3.2 Biomechanics of Balance Recovery

An important prerequisite to the development of fall prevention programs is an improved understanding of the biomechanical and neuromuscular variables that govern ability to recover balance. A large number of laboratory-based studies of balance recovery ability in young and elderly subjects have been conducted (12,17-28). These studies have used a variety of perturbation types (platform translations, release from incline, waist pulls), magnitudes, and directions (forward, backward, sideways). Furthermore these studies have employed a variety of different balance recovery strategies (feet-in-place, stepping, grasping), and have placed varying amounts of constraints on the recovery response. Some protocols have allowed fully natural movement patterns (20,21), while others have emphasized more controlled responses, including single step responses and precise body postures (12,18,27).

Despite their methodological differences, the results of these investigations, taken collectively, suggest that two types of variables are important in governing balance recovery ability: 1) those related to strength (peak magnitude of lower extremity joint torques during specific recovery responses); and 2) those related to speed-of-response (reaction time, rate of joint torque generation, and speed of movement execution).

However, there are few existing techniques for directly comparing the relative effects of strength versus speed-of-response on recovery ability, which limits our capacity to identify and target patient-specific causes of postural instability through exercise-based fall prevention interventions.

Previously, Robinovitch et al. (12) compared the relative importance of strength versus speed-of-response variables to balance recovery ability with the ankle strategy in healthy, young subjects. Using a combination of experimental and mathematical modelling techniques, they demonstrated the strong effect on recovery ability of latencies in the onset and rate of torque generation (speed-of-response variables), as well as the magnitude of torque development (strength variable). The research presented in this thesis extends their findings by comparing the recovery ability of young and elderly subjects. This comparison provides an improved understanding of the relative importance of strength versus speed-of-response in age-related declines in balance recovery performance.

1.2.3.3 Role of Strength Variables in Balance Recovery and Falls

Age-related declines in muscle strength after the sixth decade of about 30% are well-documented (29,30), and reduced lower limb strength, as detected by decreases in peak torque generation under both isometric and isokinetic conditions, is associated with decreased balance ability and increased fall risk (2,10,11,23,31,32). One reason why strength may affect balance is through its effect on BOS; functional BOS is smaller in elderly than in young because elderly have less ankle strength (33). Consequently, much attention has been directed toward fall prevention interventions that improve lower limb muscle strength.

Exercise programs have been shown to cause significant increases in strength (34-36), but programs involving only resistance training, and no specific balance training, appear unable to decrease the incidence of falls. To date, the largest research effort in this area has been the Frailty and Injuries: Cooperative Studies of Intervention Techniques (FICSIT) trials (37), a set of eight clinical trials assessing the effects of exercise on falls in older adults. Analysis of FICSIT data showed that interventions which focused on enhancing muscle strength provided small or insignificant reductions in fall rates, while those that incorporated balance training such as Tai Chi reduced falls by up to 25% (37).

It has been suggested that a possible reason for the lack of association between improvements in strength (e.g., following an exercise intervention) and reductions in fall incidence is that normal age-related loss of muscle strength does not greatly impair the ability to execute compensatory postural reactions, and that relatively healthy elderly individuals maintain the necessary muscle strength to produce adequate amounts of stabilizing torque during feet-in-place and stepping reactions (13,38,39). However, there also appears to be a strength threshold beneath which balance recovery abilities are greatly impaired (32,38). Furthermore, declines in rate of strength development occur with age, and these declines may be relatively non-responsive to normal strength training. By improving our understanding of the role of strength versus speed-of-response in age-related declines in balance recovery ability, this thesis provides previously unavailable data on the variables governing balance recovery performance, which should enhance the quality of our clinical interventions.

1.2.3.4 Role of Speed-of-Response Variables in Balance Recovery and Falls

1.2.3.4.1 Rate of Torque Generation

Balance recovery manoeuvres require the ability to generate adequate, but not necessarily maximal, torque in a limited amount of time (39). Therefore, it may be the magnitude and rate at which torque can be generated during involuntary, reactive tasks, more than the maximum voluntary strength of an individual, that limits balance recovery ability (38,40,41). To this end, Thelen et al. (29) found that older adults had maximum rates of torque development that were between 30 and 40% slower than young adults (29). These declines in rate of torque generation appeared to be linked to declines in maximum voluntary isometric strength (29). Impaired ability to rapidly develop ankle torque may limit the ability of older adults to recover balance and perform other time-critical actions that demand moderate to large strengths (41). The present thesis examined the functional consequence of age-related declines in strength and rapid torque development by determining whether these declines translated to age differences in balance recovery performance.

1.2.3.4.2 Reaction Time

In addition to rate of torque generation, reaction time is another speed-of-response variable that will determine the magnitude of corrective torque that can be generated during the recovery process, and age-related declines in balance recovery ability may be partly mediated through increased reaction time. It is well-established that reaction time increases with age by approximately 25% from the twenties to the sixties (42) and is an independent risk factor for falls (11,43). In the context of balance recovery tasks, reaction time is typically defined as the interval between the onset of a postural perturbation and the onset of a corrective response, usually detected by an increase in force or torque

generation (12,25,26). In time-critical tasks, such as balance recovery, even slight delays in response initiation may be functionally important and mean the difference between successful recovery and a fall.

Significant differences in reaction time, on the order of 10-20 ms, have been observed between young and elderly women and men during stepping responses after release from forward lean (25,26). However, these age differences were not considered biomechanically important because they constituted only a small proportion (approximately 2%) of the total time for step completion (approximately 500 ms). This conclusion is debatable, given that predictions from mathematical models show isolated increases in reaction time cause near linear decreases in maximum recovery angle (12). Previous studies also show that balance recovery ability can be strongly influenced by small but diffuse neuromuscular deficits (14,18). Moreover, reaction time strongly discriminates fallers from non-fallers (11,14).

Reaction time depends on the specific experimental conditions. Total reaction times during voluntary rapid plantar flexion movements in response to auditory stimuli have been observed in the range of 153 – 176 ms, with faster values observed for men than women, and younger adults than older adults (29,44). In balance recovery paradigms, we expect faster total reaction times because these responses are "automatic postural responses" (14,25,26), meaning they are slower than stretch reflexes but faster than voluntary reactions. Automatic postural responses are thought to be medium to long latency reflex loops that involve supraspinal pathways (14,45,46) and are therefore characterized by faster onsets of muscle contraction than voluntary movements. In fact,

total reaction times of 60-100 ms have been observed in stepping and sway-based balance recovery (12,25,26).

Reaction time can be divided into two components: (1) muscle response latency (14,45,47); and (2) electromechanical delay (EMD) (44,48,49). Muscle response latency is the time interval between the presentation of a stimulus and the onset of EMG activity and is a measure of central or neural delay, representing the time for signal perception, transmission and integration of the stimuli within the central nervous system, as well as the processing of appropriate motor commands to the muscle effectors (48).

Evidence suggests that muscle response latency lengthens with age (45,50). The cause of age-related delays in muscle response latency is unknown; however, it is unlikely that these delays can be accounted for by a single mechanism (14). It is possible that lengthened muscle response latencies in the elderly may be due primarily to peripheral sensory loss and afferent and efferent conduction impairment(s). Patients with neuropathy in the feet secondary to diabetes mellitus exhibit postural response latencies in the gastrocnemius following surface translations that are 20-30 msec longer than in healthy controls, which illustrates the importance of somatosensory information from the feet, and nerve conduction in the lower leg, in triggering postural muscle responses (46,51). Patients with profound vestibular deficits, on the other hand, exhibit altered postural response magnitudes but normal response latencies, suggesting that vestibular inputs are not used to trigger these responses (47). Alternatively, lengthened muscle response latencies in the elderly may be due primarily to central processing deficits. Since these medium latency responses involve supraspinal pathways, they may be sensitive to cognitive function and to the intactness of cortical and subcortical integrative

systems for coordinating visual, vestibular, and somatosensory inputs. For example, studies have revealed that performing secondary attentional tasks impairs the postural control of elderly individuals (52).

It is also unclear whether muscle response latency is amenable to change. Early evidence from Horak and Nashner (53) suggested that prior experience and repeated exposure to a perturbation could reduce response latency. However, later work failed to support this idea. For example, Horak et al. (14) showed that central set (expectation and practice) was able to modify postural response magnitudes but unable to reduce postural muscle response latencies. More recently, Dickstein et al. (51) showed that light touch of the fingertips, although able to improve postural stability during quiet stance in diabetic patients with peripheral neuropathy, was not able to reduce muscle response latency following platform translation in these same subjects. Whether exercise that is targeted at speed and agility training can improve muscle response latencies in the elderly has been largely unexplored and is therefore still unclear. In one exercise study, there was a strong trend indicating that elderly subjects who participated in either agility or resistance training had faster upper and lower extremity reaction times following 6 months of training (54).

Electromechanical delay, the other component of total reaction time, is the time interval between the onset of EMG activity and mechanical force production (48,49), or movement onset (44), and has been associated with the time required for the propagation of action potentials along the sarcolemma and t-tubules, the release and accumulation of Calcium ions in the cytosol, the formation of cross-bridges, and subsequent tension

development in the contractile component and stretching of the series elastic component (44,49).

1.2.3.5 Age-Related Changes in Muscle

There is considerable evidence of the physiological mechanisms underlying declines with age in the peak magnitude and rate of force that can be developed in muscle. Aging beyond the sixth decade is marked by a reduction in the number of excitable motor neurons and motor units, and subsequent motor unit remodeling (denervation/reinnervation of muscle fibers) (30,55-57). Remodeling can result in type II (fast) fibers being reinnervated by axonal branches of motor neurons that primarily innervate type I (slow) fibers (55) and a subsequent loss in ability to produce fast, powerful contractions (56). There is also slowing of motor unit firing rates with age, especially during maximal voluntary contractions (56-58), and a consequential loss of type I and (more predominantly) type II muscle fibers (30,55-57). These events leave aged muscle with a smaller percentage of type II fibers, and therefore, a decreased ability to generate force rapidly. The ability to develop force quickly in lower extremity muscles is further impaired by an increase in muscle twitch contraction durations and relaxation times, which has been linked to prolonged cross-bridge cycling of aging myosin, and impaired Calcium release and uptake in the sarcoplasmic reticulum (30,55,57). Loss of muscle mass also has substantial effects on strength (59). Indeed, from the sixth decade onwards muscle strength declines steadily at approximately 1-1.5% per year (30). However, age-related declines in strength are considerably less for eccentric (lengthening) contractions than for concentric and isometric contractions (30,60). The mechanisms that account for the maintenance of eccentric strength need further

explanation, but age-related changes in muscle mass, contraction speed, and connective tissue seem to contribute (30). This selective preservation of eccentric strength has important implications for balance recovery, as the crucial initial phase of recovery (where downward motion of the body is halted) involves eccentric muscle contractions.

1.2.4 Postural Steadiness

In our conceptual model of fall risk (p.2), frequency of loss-of-balance episodes, in addition to balance recovery ability, is an important risk factor for falls. Frequency of loss-of-balance depends on an individual's ability to maintain balance. During upright stance, balance maintenance involves keeping the body COM within the limits of stability defined by the BOS through movement of the foot centre-of-pressure (COP). Thus, loss-of-balance will occur if the COM moves outside the BOS, in which case an effective balance recovery strategy will be necessary to avoid falling. Balance maintenance is often assessed through measurement of postural steadiness. Postural steadiness can be defined and measured in a variety of ways, including movement of a single point of the body (e.g., shoulder, waist), movement of the COP as recorded from a force plate, or movement of the whole body COM, as determined through inverse dynamics with position data.

Numerous studies have shown that postural sway increases with age (61-65), but that age differences are sensitive to subject recruitment and the difficulty of the balance task. Generally, the difference in sway between healthy young and healthy elderly subjects is less than between healthy young and elderly with co-morbidities, and age differences are more obvious as the difficulty of the balance task increases (e.g., when sensory information is removed or the BOS is narrowed) (66). Furthermore, elderly

fallers tend to have greater sway than elderly non-fallers (62,67), and the sway patterns of elderly fallers and non-fallers differ. The sway pattern of fallers tends to be non-enclosed, such that sway moves progressively in one direction (61). Prospective studies have shown that postural sway during quiet stance is predictive of fall risk (11,68), and lateral sway with eyes closed has been identified as one of the stronger predictors (68).

The usefulness of postural sway as an index of balance and predictor of fall risk has been challenged however (69). First, although convention indicates that greater sway amplitudes and velocities imply poorer stability, some studies have failed to support this idea. For example, Parkinsonian patients exhibit normal sway during quiet stance, possibly because of increased stiffness of the neuromuscular system, even though they have difficulty with balance control in other situations such as gait (70). Also, medial-lateral COP speed was faster among subjects who recovered successfully after tripping than among subjects who failed to recover (69). Second, it has been argued that quiet stance does not present the same level of neuromuscular demands on the postural control system as balance recovery tasks present (69). Compared to maintenance of quiet stance, balance recovery requires larger muscle strengths, and faster reaction times and rates of torque generation. Thus, postural steadiness during quiet stance may not adequately characterize an individual's balance capabilities. At the same time, postural steadiness during quiet stance and balance recovery ability may rely on similar fundamental sensory and motor variables such that performance in one situation is predictive of performance in the other. The results of this thesis provide important new information about the relationship between postural steadiness during quiet stance and balance recovery ability

that will guide the development of balance and fall risk assessment tools for future research.

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Chapter 2
Mechanisms Underlying Age-Related Declines in Ability to Recover Balance with the Ankle Strategy [1]

2.1 Abstract

Falls cause substantial death and morbidity in the elderly. An important prerequisite to the development of fall prevention interventions is improved understanding of the biomechanical and neuromuscular variables that govern ability to recover balance. In the present study we examined the relative importance of strength versus speed-of-response variables in explaining differences in balance recovery performance with the ankle strategy between young and elderly women. Twenty-five young (19-36yr) and 25 community-dwelling elderly women (66-90yr) participated in balance recovery experiments. Subjects were supported in an inclined standing position by a horizontal tether and instructed to recover an upright vertical standing position by contracting their ankle muscles. The maximum initial lean angle where they could recover balance without release of the tether (which depends primarily on strength) was 19.6% smaller for elderly than young. The maximum initial lean angle where they could recover balance after the tether was suddenly released (which depends on both strength and speed-of-response) was 36.1% smaller for elderly than young. Moreover, between-group differences in performance were related to both strength and speed-of-response. Peak ankle torque was 7.7% smaller in elderly than young during tether release trials;

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reaction time was 27% slower in elderly, due to a lengthened muscle response latency; and rate of ankle torque generation was 15.6% slower in elderly. These results suggest that age-related declines in ability to recover balance associate with variables related to strength and speed-of-response, and that exercise-based fall prevention programs should include balance and agility training, in addition to strength training.

Key words: accidental fall, postural balance, reaction time, strength, biomechanics.

2.2 Introduction

Falls are the leading cause of injury and injury-related death in the elderly [1]. While fall risk depends on a variety of sensory, motor, cognitive, psychosocial, and environmental variables [2,3], it depends ultimately on one's frequency of imbalance episodes, and one's ability to recover balance after such episodes [4]. Therefore, fall prevention programs should target each of these areas.

An important prerequisite to the development of fall prevention programs is an improved understanding of the biomechanical and neuromuscular variables that govern ability to recover balance. Both strength [5] and speed of response to loss of balance [6-8] are known to decline with age, and these changes lead to increased fall risk [2,3,9]. However, the relative importance of strength versus speed-of-response variables in explaining age-related changes in ability to recover balance is unknown. It is important to identify the declines with age in the relative contribution to balance recovery of each of these variables, since fundamentally different types of interventions (e.g., exercise programs) may be required to target each of these variables.

Previous studies suggest that strength variables, such as peak attainable torque, and speed-of-response variables, such as reaction time and rate of torque development, are important in determining young subjects' ability to recover balance [4,10,11]. In this study we investigated whether ability to recover balance declines with age, and, if so, whether this is due to declines in speed-of-response variables or strength variables. To address this question, we conducted balance recovery experiments with young and elderly women and measured the maximum forward lean angle where subjects could recover a stable upright stance by contracting their ankle muscles (ankle strategy). We examined this index of balance recovery ability when subjects self-initiated their recovery versus recovered balance after being suddenly released from a forward leaning position. The former parameter, which was termed "maximum static recovery angle", should depend primarily on factors related to muscle strength, while the latter parameter, which was termed "maximum dynamic recovery angle", should depend on factors related to both muscle strength and speed-of-response. The percent difference between young and elderly in the maximum static recovery angle was therefore used to determine the effect of declines in strength, while the percent difference between young and elderly in the maximum dynamic recovery angle was used to determine the effect of declines in strength and speed-of-response. In addition, we also examined performance variables from the recovery trials, including peak ankle torque, reaction time, and rates of ankle torque generation and decline, and we associated these variables with the maximum dynamic recovery angle to further determine the effect of declines in strength and speed-of-response.

2.3 Methods

2.3.1 Subjects

Twenty-five young adult women of mean age 25 ± 4 (SD) yr (range: 19–36 yr), and 25 elderly adult women of mean age 78 ± 7 yr (range: 66–90 yr) participated in this study. Young subjects were recruited through notices at Simon Fraser University, while elderly subjects were recruited from an education-based fall prevention program that operated between autumn 2002 and winter 2003 in local seniors' centres. To be eligible for the study, elderly subjects must have reported at least one fall, as described in the FICSIT definition [12], either in the 12 months prior to their entry into the fall prevention program, as recorded on their intake questionnaire, or since their entry into the program, as recorded on their monthly fall passports. These subjects represented a relatively high-risk group for future falls.

Subjects were initially interviewed in-person and individuals were excluded if they had conditions that would prevent them from performing the experiments, or possessed known risk factors for falls that would represent confounding variables in the analysis, since our intent was to determine factors that affect risk for unexplained falls and balance impairments. These included terminal illness, inability to stand unassisted for 10 minutes, physical deformity of the lower limbs or spinal column which affect stability (including dramatically different leg or foot lengths, foot deformities, tendon abnormalities, and scoliosis), blindness, Parkinson's disease, stroke, peripheral neuropathy or plantar foot ulcers, profound and recurring episodes of vertigo, dizziness, or loss of consciousness in the past 3 months, and use of psychotropic medications (hypnotics, anxiolytics, antidepressants, and antipsychotics) [13]. Subjects were excluded

after an on-site evaluation if they had (1) moderate to severe dementia (Folstein Mini-Mental State Examination score less than 21) [14]; (2) major uncorrected deficits in visual acuity (Snellen score worse than 20/15 at 5 feet); (3) evidence of profound vestibular deficits (gross instability when standing on foam with eyes closed); or (4) evidence of profound somatosensory deficits (measured by big toe position sense and monofilament to the dorsum of the foot). Each subject was paid $10/hour for her participation. Each subject provided informed written consent, and the experiment was approved by the University Research Ethics Board at Simon Fraser University.

2.3.2 Ancillary Measurements

Elderly subjects visited the laboratory for two sessions (usually no more than one week, and at the most, 2 weeks apart), while young subjects visited once. During the first session for elderly subjects, and the first half (approximately 1.5 hours) of the session for young subjects, we acquired ancillary measures to characterize the samples. We measured Get-up and Go time (6 meters) [15] (young=9.8±1.5 sec; elderly=14.2±2.7 sec), Functional Reach [16] (young=38.1±4.4 cm; elderly=27.4±5.9 cm), Folstein Mini-Mental State (scored 0-30) [14] (young=29±1; elderly=27±2), and Activities-specific Balance Confidence (scored 0-100%) [17] (young=98±3%; elderly=79±16%).

2.3.3 Experimental Procedures

The experimental methods are similar to those reported by Robinovitch et al. [4]. We conducted balance recovery experimental trials where we measured the maximum recovery angle from which subjects were able to recover balance (a stable upright vertical stance) primarily by contracting the muscles spanning the ankle joint, a technique called

the "ankle strategy" [18,19]. To conduct the experimental trials, the subject stood on a force plate with her feet a comfortable width apart and arms crossed against her chest. We then inclined the subject into a stationary forward leaning posture via a horizontal tether that attached at one end to an electromagnetic brake (Warner Electric model PB500, South Beloit, IL), and at the other end to a chest harness worn by the subject (Fig.2-1). As a safety precaution, we also attached an overhead fall restraint tether to the chest harness. We instructed the subject to rise into a vertical standing position by contracting the muscles spanning the ankles, keeping the knees and hips fully extended; the heels were allowed to leave the ground. Practice trials were provided until the subject understood the requirements of the ankle strategy and was comfortable leaning into the tether before release. This was typically achieved in three to six practice trials.

To determine the effect on recovery ability of the magnitude versus speed of torque development, the maximum recovery angle was measured under "static" and "dynamic" conditions. In static trials, the subject attempted to voluntarily rise into a standing position when the tether was not released (Fig.2-2A). Following a "ready" cue from the subject, the investigator provided an auditory tone, signaling the subject to begin the recovery process. In dynamic trials, the subject attempted to recover a stable upright position after the tether was suddenly released, which caused a near-step increase in gravitational torque acting to rotate the body downward (tether release time ~15ms) (2-2B). To increase the unexpectedness of the perturbation, the investigator inserted a random time delay of 1-10 sec between receiving the "ready" cue from the subject and the time of release. In both static and dynamic conditions, the first trials involved small lean angles of ~2°, where the subject could recover easily. We then iteratively adjusted

the length of the tether and the corresponding lean angle until we identified the maximum recovery angle (with a resolution of 7 mm in tether length, and ~0.3° in lean angle), beyond which the subject could no longer recover balance in at least three of five repeated trials. The order of presentation of the two different experimental conditions was counterbalanced across subjects to minimize order or learning effects. Before each trial, the subject was instructed to maintain her gaze forward and at eye level on a black X attached to the wall approximately 10 feet in front of her.

Subjects were barefoot during the trials and wore tight-fitting shorts and shirt. To offset the possibility of subject fatigue, rest breaks of approximately 30 sec duration were provided between trials, and 5 minute sitting breaks were provided between the different conditions. In static trials, we monitored tether force to ensure there was no pushing against the tether during recovery which could aid performance. During dynamic trials, we controlled the magnitude of ankle torque before release, which can affect recovery ability [4]. Since ankle torque scales with body height and body weight [4], we aimed to keep the ratio of ankle torque before release divided by the product of subject body height and body weight consistent across subjects and close to its average baseline value measured during quiet standing. This was achieved by using a computer monitor to provide visual feedback to the subject and investigator of foot centre-of-pressure (COP) position (an indicator of baseline ankle torque). We set a target COP position for each subject so that the ratio of COP divided by body height matched (to within ± 10 mm) the average value measured during quiet stance in a subset of six subjects (Appendix A).

During each trial we measured the magnitude and point of application of foot-floor reaction forces and moments at a rate of 960 Hz with a force plate (model 6090H,

Bertec, Worthington, OH). We also measured body segment movements with a 60 Hz, seven-camera motion measurement system (ProReflex, Qualisys Inc., Glastonbury, CT) that recorded the position of 16 markers attached to the skin and clothing overlying the right and left fifth metatarsal, lateral malleolus, lateral femoral condyle, anterior superior iliac spine, pisiform bone, radial head, acromion process, and the head and sacrum. Muscle activities in the tibialis anterior and soleus of the dominant leg were measured through surface electromyography (16-channel Myosystem 1200, Noraxon, Inc.), sampled at 960 Hz. In dynamic trials, the instant of tether release was detected as the onset of a sharp decline in the tension (≥2 N) measured by a load cell (Sensotec, model 31) located in series with the tether.

2.3.4 Data Analysis

We calculated the body lean angle $\theta(t)$, defined as the angle from the vertical to a line connecting the midpoint of the two lateral malleolus markers to the midpoint of the two acromion markers [4] (Fig.2-1). We also calculated time-varying ankle plantar-flexor torque $T_a(t)$ based on the location and magnitude of vertical and horizontal components of foot reaction force:

$$T_a(t) = F_z(t)x(t) - F_x(t)z(t) - W_f x_f(t)$$

where F_z is the resultant vertical force acting on the foot (defined positive if upward), x is the horizontal distance (anterior/posterior) from the ankle joint marker to where F_z acts (defined positive if anterior to the ankle), F_x is the resultant horizontal force in the sagittal plane acting on the foot (defined positive if directed posteriorly), z is the vertical height of the ankle joint marker above the ground, W_f is the weight of the

foot (0.0145 x body weight) [20], and x_f is the horizontal distance from the ankle marker to the centre-of-mass of the foot, assumed to be 0.5 of foot length [20]. In our calculation of ankle torque, we neglected inertial forces associated with angular acceleration of the feet, which have be shown to be negligible [4].

EMG recordings were high pass filtered to remove motion artifact (4th order Butterworth, cutoff frequency = 40 Hz), rectified, and low pass filtered (4th order Butterworth, cutoff frequency = 20 Hz) to determine the envelope of signal intensity. In dynamic trials, the onset of increased EMG activity in the soleus was determined as the time that signal intensity rose 3 SDs above the mean value measured in the 500 msec preceding tether release [21-23].

From each of the three maximum recovery trials per condition, we calculated the following dependent variables: (a) the maximum recovery angle (θ_{max}) calculated as the average value of $\theta(t)$ over the 500 msec interval preceding tether release in dynamic trials, and as the maximum value of $\theta(t)$ in static trials; and (b) the peak ankle torque (T_{max}) generated during balance recovery (Fig.2-3). For each subject, we also calculated the ratio of maximum dynamic recovery angle divided by maximum static recovery angle to reflect the percent decline in recovery ability due primarily to finite reaction times and rates of torque generation [4].

In dynamic trials we also calculated 1) the magnitude of ankle torque before release (T_o), calculated as the average value of $T_a(t)$ over the 500 msec preceding release (Fig.2-3); 2) the muscle response latency, defined as the interval between tether release and the onset of increased EMG activity in the soleus; 3) the electromechanical delay

(EMD), defined as the interval between onset of increased EMG activity in the soleus and increased ankle torque, which was identified as the instant when $T_a(t)$ exceeded T_o by 5 Nm (always outside baseline variability); 4) the total reaction time (Δt), defined as the sum of the muscle response latency and EMD; 5) the rate of ankle torque generation following release (C), defined as the slope of a straight line joining torque-time values at the instant $T_a(t)$ exceeded T_o by 5 Nm to the instant $T_a(t)$ equaled 85% of the difference between T_o and $T_{\max}$; and 6) the rate of ankle torque decline (D) following $T_{\max}$, defined as the slope of a straight line joining torque-time values at the instant of $T_{\max}$ to the instant $T_a(t)$ declined by 85% of the difference between $T_{\max}$ and the minimum ankle torque during recovery. For the rate of ankle torque generation, the 85% value was chosen instead of $T_{\max}$ because it always reflected a point on the initial smooth rise of the torque time curve, while $T_{\max}$ was sometimes offset from the initial rise due to small oscillations (secondary peaks) in $T_a(t)$. For consistency we also chose the 85% value to define the rate of ankle torque decline. We normalized values of T_o, $T_{\max}$, C, and D by the product of body mass (in kg) multiplied by body height (in m). Values of $\theta_{\max}$, T_o, $T_{\max}$, muscle response latency, EMD, Δt, C, and D used in statistical analyses were averages over three repeated trials for each subject.

2.3.5 Statistics

We used independent samples *t* tests to determine whether average values of $\theta_{\max}$ and $T_{\max}$ in the static and dynamic trials were different between young and elderly. We also used independent samples *t* tests to determine whether reaction time (Δt), muscle

response latency, EMD, rate of ankle torque generation (C), and rate of ankle torque decline (D) in the dynamic trials, as well as the ratio of dynamic divided by static maximum recovery angle, were different between young and elderly. We used Pearson product moment correlation coefficients to test for association between maximum dynamic recovery angle, maximum static recovery angle, reaction time, rate of ankle torque generation, and peak ankle torque. Finally, we constructed three forward stepwise regression models to determine predictors of maximum dynamic recovery angle (θ_{max}) for 1) the entire sample, 2) young subjects, and 3) elderly subjects. The variables we entered in these models (based on the correlation coefficients) were reaction time (Δt), muscle response latency, peak ankle torque in dynamic recovery trials (T_{max}), rate of ankle torque generation in dynamic trials (C), and maximum static recovery angle (θ_{max}). Variables were entered at a significance level of $P \leq .05$ and removed at $P \geq .10$.

2.4 Results

2.4.1 Effect of Age on Ability to Recover Balance

Ability to recover balance declined with age (Fig.2-4, Table 2-1). In particular, the average maximum static recovery angle was 19.6% smaller in elderly than in young (mean difference=3.2 deg; 95% CI: 2.0 to 4.3 deg, t=5.56, df=39.4, $P<.001$), and the average maximum dynamic recovery angle was 36.1% smaller in elderly than in young (mean difference=2.6 deg, 95% CI: 1.7 to 3.4 deg, t=5.84, df=48, $P<.001$). In turn, the ratio of dynamic divided by static maximum recovery angle was 20.5% smaller in elderly than in young (young=0.44±0.08 vs. elderly=0.35±0.12, mean difference=0.09; 95% CI: 0.03 to 0.15, t=3.15, df=40.4, $P=.002$).

2.4.2 Effect of Age on Strength and Speed-of-response

Peak ankle torque (parameter T_{max} shown in Fig.2-3) was 4.8 % smaller for elderly than young in static recovery trials, although this difference failed to reach statistical significance (mean difference=0.045 Nm/(kg·m), 95% CI: -0.01 to 0.10 Nm/(kg·m), t=1.65, df=33.6, *P*=.054), and 7.7% smaller for elderly than young in dynamic recovery trials (mean difference=0.08 Nm/(kg·m), 95% CI: -0.01 to 0.17 Nm/(kg·m), t=1.83, df=48, *P*=.037) (Table 2-1). In dynamic trials, it took elderly subjects longer than young subjects to initiate ankle torque development. In particular, reaction time (parameter Δt in Fig.2-3) was 27.0% slower for elderly than young (mean difference=-27 msec, 95% CI: -37 to -17 msec, t=-5.54, df=48, *P*<.001). This was due to a significantly longer muscle response latency (mean difference=-25 msec, 95% CI: -36 to -14 msec, t=-4.51, df=48, *P*<.001), as electromechanical delay was not different for elderly and young subjects (mean difference=-2 msec, 95% CI: -10 to 6 msec, t=-0.48, df=48, *P*=.316). Once torque generation was initiated, rate of ankle torque generation (parameter *C* shown in Fig.2-3) was 15.6% slower for elderly than young (mean difference=1.03 Nm/(sec·kg·m), 95% CI: -0.25 to 2.30 Nm/(sec·kg·m), t=1.62, df=48, *P*=.057), and rate of ankle torque decline (parameter *D* shown in Fig.2-3) was 13.8% slower for young than elderly (mean difference=-0.09 Nm/(sec·kg·m), t=-1.39, df=48, *P*=.085); however, neither of these differences reached statistical significance.

2.4.3 Correlation Between Balance Recovery Ability, Strength, and Speed-of-response

The maximum dynamic recovery angle was correlated with reaction time in dynamic trials (r=-0.657, *P*<.001), with the peak ankle torque in dynamic trials (r=0.571,

$P<.001$), with the rate of ankle torque generation in dynamic trials (r=0.321, $P=.011$), and with the maximum static recovery angle (r=0.669, $P<.001$) (Fig.2-5). These correlations persisted, with three exceptions, when young and elderly data were analyzed separately. For young subjects, maximum dynamic recovery angle was correlated with reaction time in dynamic trials (r=-0.389, $P=.027$), and with the peak ankle torque in dynamic trials (r=0.471, $P=.009$), but not with the rate of ankle torque generation (r=0.235, $P=.129$), or with the maximum static recovery angle (r=0.285, $P=.084$). For elderly subjects, maximum dynamic recovery angle was correlated with reaction time in dynamic trials (r=-0.483, $P=.007$), with the peak ankle torque in dynamic trials (r=0.611, $P=.001$), and with the maximum static recovery angle (r=0.515, $P=.004$), but not with rate of ankle torque generation (r=0.254, $P=.110$).

2.4.4 Regression Analysis

Stepwise linear regression revealed that maximum static recovery angle, reaction time, and peak ankle torque during dynamic recovery trials together explained 62% of the variance in maximum dynamic recovery angle ($P<.001$). Maximum static recovery angle accounted for 45% of the variance. After accounting for maximum static recovery angle, reaction time explained another 12% of the variance, and then peak ankle torque explained a further 5%. Among young and elderly subject groups separately, maximum static recovery angle was the only significant predictor of maximum dynamic recovery angle, explaining 22% ($P=.017$) and 37% ($P=.001$) of the variance for young and elderly respectively.

2.4.5 Magnitude of Ankle Torque Before Release

On average, the magnitude of ankle torque before release in the dynamic trials (parameter T_o shown in Fig.2-3), normalized for body height and body weight, was not different from normalized ankle torque during quiet stance for young (mean dynamic=0.37±0.14 Nm/(kg·m) vs. mean quiet stance=0.33±0.09 Nm/(kg·m), P=.059) or elderly (mean dynamic=0.30±0.18 Nm/(kg·m) vs. mean quiet stance=0.35±0.02 Nm/(kg·m), P=.194) subjects. Further, normalized T_o was not different between young and elderly subjects (P=.141).

2.5 Discussion

In this study, we found that ability to recover balance using the ankle strategy was affected by age. Elderly women had a 36.1% smaller maximum dynamic recovery angle, and a 19.6% smaller maximum static recovery angle than young women (Table 2-1, Fig.2-4). This in agreement with previous studies that reported age-related reductions in ability to recover balance with the stepping strategy [24,25].

Our results also suggest that age-related declines in ability to recover balance associate with variables related to both speed-of-response and strength. Reaction time was 27.0% slower in elderly than in young (a 27 msec difference), and there was a trend for rate of ankle torque generation to be slower in elderly than in young. Maximum dynamic recovery angle was correlated with both reaction time and rate of ankle torque generation, and reaction time was a significant predictor of maximum dynamic recovery angle. Furthermore, maximum dynamic recovery angle, and the ratio of dynamic divided by static maximum recovery angle were smaller in elderly than young, both suggesting

that speed-of-response requirements have a larger influence on the balance recovery ability of elderly than of young.

Although the percent differences in peak ankle torque between young and elderly were not as large as the differences in reaction time or rate of ankle torque generation, strength appears to be an important variable governing age-related declines in balance recovery ability. Peak ankle torque during dynamic trials was correlated with, and a significant predictor of, maximum dynamic recovery angle, and when elderly subjects were analyzed separately, maximum dynamic recovery angle was correlated more strongly with peak ankle torque than with reaction time. Furthermore, maximum static recovery angle, an index of strength, was smaller in elderly than young, and it was the strongest predictor of maximum dynamic recovery angle in the sample as a whole, and when young and elderly were analyzed separately.

Our results show that age-related declines in reaction time are functionally significant, despite appearing to be small. The average difference in reaction time between young and elderly subjects in our study was 27 msec, which is similar to the 13-23 msec differences in reaction time that Wojcik et al. [24] reported for their young and elderly female subjects during stepping tasks. In the context of their stepping tasks, where recovery times averaged 500 msec, Wojcik et al. concluded that age-related delays in reaction time of 20-25 msec are not functionally significant. However, in the context of our feet-in-place balance recovery task, where recovery occurs more quickly (<200 msec) than during stepping, we have shown that age-related delays in reaction time are functionally significant in determining ability to recover balance. Maximum dynamic recovery angle was negatively correlated with reaction time, in agreement with

predictions from mathematical models that show isolated increases in reaction time cause near linear decreases in maximum recovery angle [4]. Previous studies also show that balance recovery ability can be strongly influenced by small but diffuse neuromuscular deficits [18,26]. Moreover, reaction time strongly discriminates fallers from non-fallers [3,18].

Our results for muscle response latency and electromechanical delay suggest that age-related neural changes in the sensing of stimuli and processing of motor commands (represented by the muscle response latency), rather than age-related changes in muscle contraction mechanics (represented by the electromechanical delay), may govern age-related declines in balance recovery ability. Average muscle response latencies for the soleus were 25 msec slower in elderly than in young, which is similar to the 21 msec difference that Thelen et al. [27] observed between young and elderly women during rapid isometric contractions of the soleus. In contrast, we found that average electromechanical delay was not significantly slower in elderly than in young. The muscle responses that we observed had latencies longer than the monosynaptic stretch reflex but shorter than volitional reaction times, and therefore reflect medium latency postural responses [8,28].

Localizing the neurological cause of age-related slowing of muscle response latency may help to identify the feasibility of improving reaction time in the elderly and the most promising interventions for achieving this. Previous research has established that peripheral sensory loss and afferent and efferent conduction impairment(s) result in lengthened muscle response latencies [22,23], while vestibular deficits affect the magnitude but not the timing of postural responses [29]. Alternatively, lengthened muscle

response latencies in the elderly may be due to central processing deficits. These medium latency responses involve supraspinal pathways, and may be sensitive to cognitive function and the intactness of cortical and subcortical integrative systems for coordinating visual, vestibular, and somatosensory inputs. For example, studies have revealed that secondary attentional tasks impair the postural control of elderly individuals [30]. Thus, to improve our understanding of age-related declines in ability to recover balance, future research should assess how proprioceptors, cutaneous afferents, and central processing affect recovery through manipulations of somatsosensory inputs and secondary cognitive tasks.

Our results show that elderly not only take longer to generate torque in response to a postural perturbation (indicated by a longer reaction time), but once torque onset has occurred, they also develop torque at a rate that is approximately 16% slower than young. While these age differences in rate of torque generation were not statistically significant (P=.057), we consider the magnitude of the percent difference to be functionally relevant, and presumably with a larger sample size, we would have observed statistical significance. Our results for rate of torque generation are consistent with the well-documented loss of fast twitch (Type II) muscle fibers that occurs with age [5,31-33]. Furthermore, Thelen et al. [7] previously demonstrated that maximum rate of ankle torque generation was 36% slower in older females than in younger females. The discrepancy in magnitude probably relates to differences in the nature of the task: Thelen et al. measured rate of torque generation under voluntary isometric conditions while subjects lay supine, whereas we measured rate of torque generation during upright stance in response to a postural perturbation.

There are certain limitations to this study. First, we excluded individuals with major neurological, musculoskeletal, cognitive, or sensory deficits, and thus our results may have limited applicability to individuals with profound cognitive or neuromuscular impairment. Another limitation of the study is that the magnitude and direction of the balance perturbation were somewhat predictable to the subject. We examined only forward loss-of-balance and instructed subjects to recover using the ankle strategy rather than allowing them to choose between feet-in-place or stepping responses [34,35]. This is substantially different from a real-life loss of balance episode when the timing, magnitude, and direction of external perturbations are more unexpected and movement planning is more limited. However, by measuring performance in a specific balance recovery task, we were able to minimize behavioural influences and clearly identify biomechanical contributions to performance.

A third limitation of the study is that subject performance may have been influenced by partial reliance on the hip strategy. We visually inspected each experimental trial during data collection and repeated any trials with noticeable knee or hip flexion. In post-hoc analysis we calculated peak hip flexion rotations during recovery. For dynamic trials, these rotations ranged between 1.6 and 21.1 deg and averaged 6.1 deg in young and 9.5 deg in elderly. These hip flexion rotations did not associate with maximum dynamic recovery angle for either young (R=0.16, P=.051) or elderly (R=0.01, P=.603) subjects. Furthermore, the trend for greater hip flexion rotations in the elderly could not invalidate the main conclusions of this study. For static trials, peak hip flexion rotations ranged between 0.05 and 11.2 deg and averaged 1.2 deg in young and 1.4 deg in elderly. Again, these rotations also did not associate with maximum static recovery angle

for either young (R=0.12, *P*=.092) or elderly (R=0.01, *P*=.591) subjects. This is not to say that trunk and hip muscles did not contribute to the balance recovery response. Subjects had to stabilize the hip and minimize movement between the trunk and lower extremities. Therefore, the ability to coordinate trunk and ankle muscle contractions may have limited performance in both static and dynamic conditions.

In summary, we have shown that age-related declines in ability to recover balance with the ankle strategy are accounted for by a lengthening of reaction time, in particular muscle response latency, and decrements in rate of ankle torque generation and peak ankle torque. Thus, age-related declines in ability to recover balance associate with variables related to both speed-of-response and strength. These results complement growing clinical evidence that exercise programs to prevent falls should include balance and agility training, in addition to strength training [36-38].

2.6 Acknowledgements

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2.7 Figures

Figure 2-1 Balance recovery experiment.
Subjects were held by a tether in an initially inclined position and instructed to recover a stable upright position by contracting their ankle muscles. Their "maximum static recovery angle" was the maximum initial forward lean angle (θ_o) where they could accomplish this task when the tether was not released. Their "maximum dynamic recovery angle" was the maximum θ_o where they could recover balance after the tether was suddenly released.

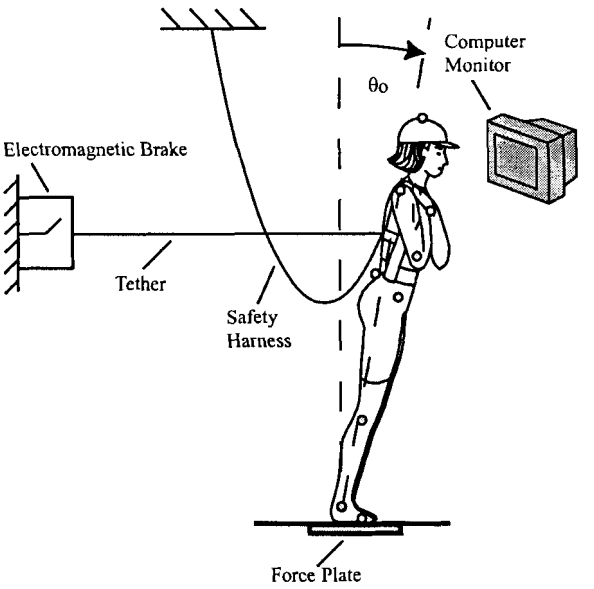


Figure 2-2 Typical variations in kinematic and kinetic parameters for a 22 year old young subject (heavy traces) and a 66 year old elderly subject (light traces) in (A) static recovery trial and (B) dynamic recovery trial.
(A) In the static trial each subject self-initiated her recovery into a stable upright position by increasing her ankle torque. This caused the body lean angle to decrease and the tether force to decline. The young subject generated a greater peak ankle torque than the elderly subject, and a greater maximum recovery angle. (B) In the dynamic trial, tether release (indicated by the dashed line) was followed by a sharp decline in tether force. After release, the body rotated downward (lean angle increased), and after a time delay of 88 ms for the young subject and 107 ms for the elderly subject, ankle torque began to increase and halt downward rotation of the body, which allowed for return to upright posture. The soleus muscle response latency was longer for the elderly subject, and consequently the onset of ankle torque generation was later. Also, the rate of ankle torque generation and the peak ankle torque were smaller for the elderly subject. Note that while peak ankle torque was greater in the dynamic trial than the static trial for both subjects, their maximum recovery angles were smaller in the dynamic trials.

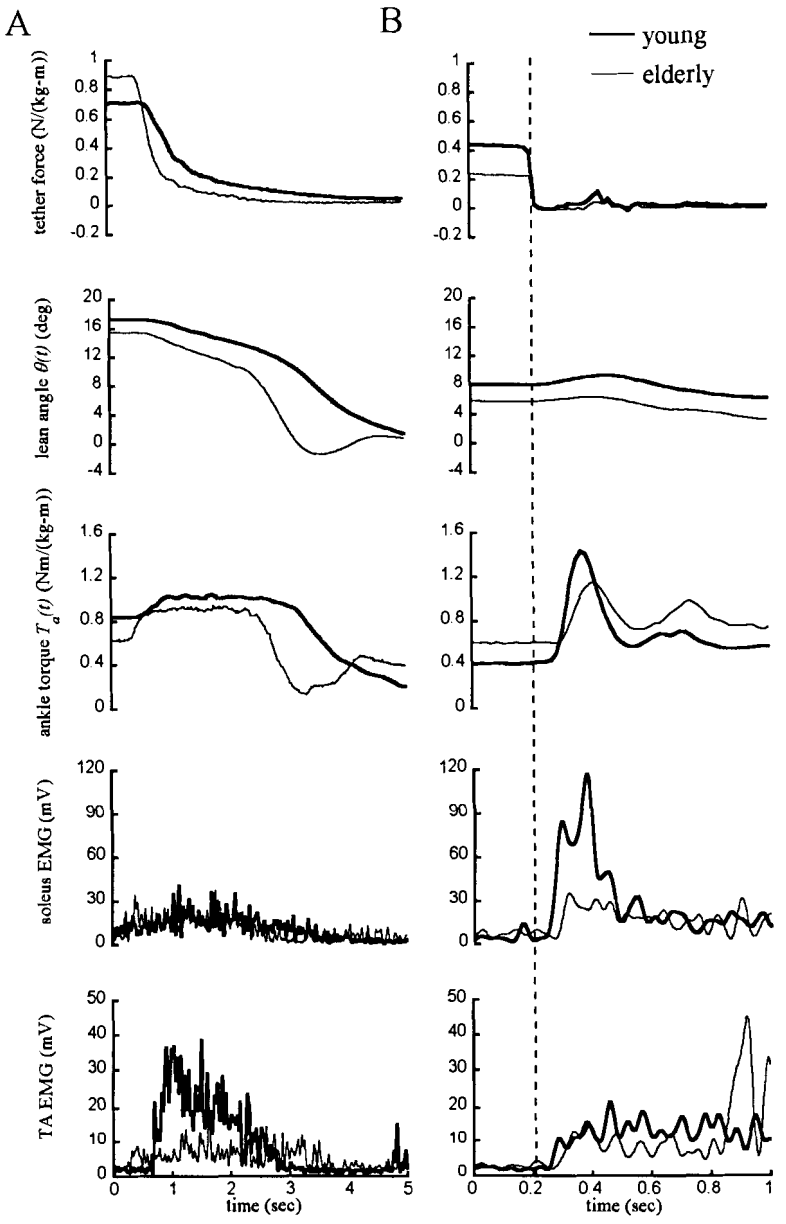


Figure 2-3 Ankle torque profile characteristics and measured reaction time components.
We detected tether release (shown by the earliest vertical dashed line) by a sharp decline in tether force. Before tether release, subjects adjusted their ankle torque to match the average value during quiet standing (T_o) (lower panel). Following release, there was a reaction time (Δt) before the onset of increased ankle torque generation. Ankle torque was generated at a rate C and reached a peak, $T_{\max}$, before declining at a rate D. Total reaction time (Δt) was composed of muscle response latency and electromechanical delay (EMD). Onset of increased EMG activity in the soleus is shown by the second vertical dashed line, and onset of increased ankle torque production is shown by the third vertical dashed line. We regarded muscle response latency, EMD, Δt, C, and D as speed-of-response variables, and $T_{\max}$ as a strength variable.

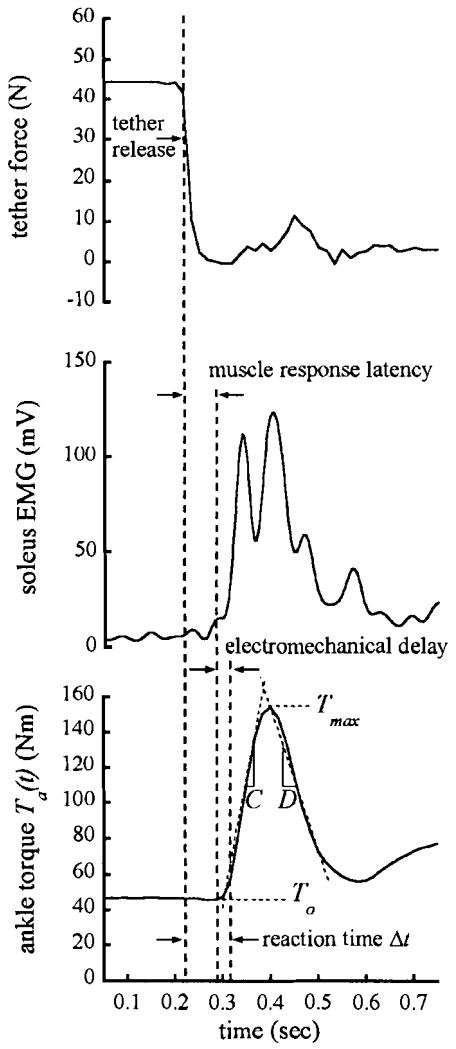


Figure 2-4 Maximum recovery angles under static conditions (squares) and dynamic conditions (circles) for both young (filled symbols) and elderly (unfilled symbols) subjects.
Lines show average values. The average maximum static recovery angle was 19.6% smaller in elderly than in young, and the average maximum dynamic recovery angle was 36.1% smaller in elderly than in young.

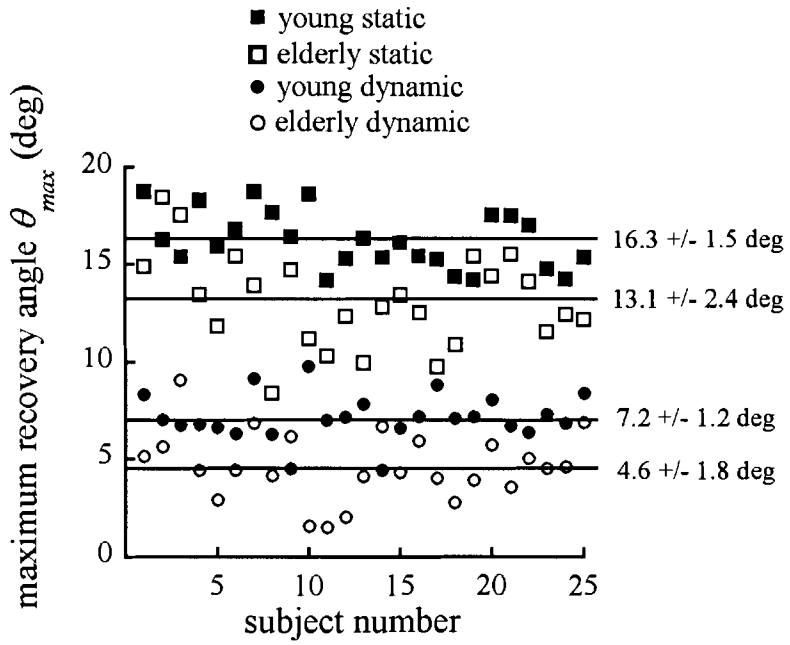
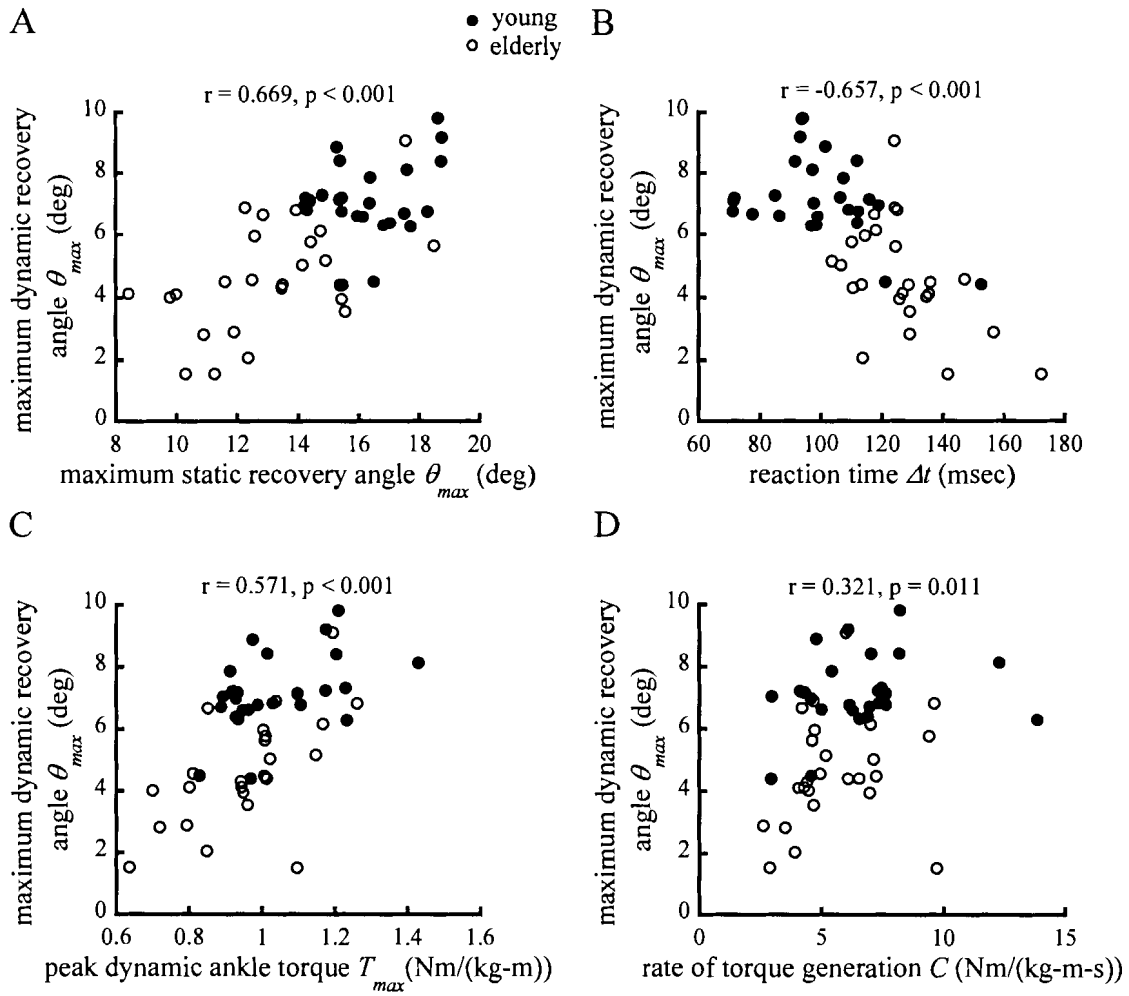


Figure 2-5 Correlation between maximum dynamic recovery angle and (A) maximum static recovery angle, (B) reaction time, (C) peak ankle torque from the dynamic trials, and (D) rate of ankle torque generation, indicating that ability to recover balance is associated with both strength and speed-of-response variables.
Young (filled circles), elderly (unfilled circles).



2.8 Tables

Table 2-1 Dependent variables by age category.

Dependent Variable	Young (n = 25)	Elderly (n = 25)	*P*-value
Maximum *static* recovery angle, θ_{max}, (deg)	16.3 ± 1.5	13.1 ± 2.4	<.001
Maximum *dynamic* recovery angle, θ_{max}, (deg)	7.2 ± 1.2	4.6 ± 1.8	<.001
Peak *static* ankle torque, T_{max}, (Nm)*	95.1 ± 17.5 (0.93 ± 0.06)	87.5 ± 20.9 (0.89 ± 0.12)	.054
Peak *dynamic* ankle torque, T_{max}, (Nm)*	106.4 ± 27.3 (1.04 ± 0.15)	94.3 ± 23.4 (0.96 ± 0.16)	.037
Rate of ankle torque generation, C, (Nm/sec)*	672.6 ± 293.0 (6.60 ± 2.49)	541.2 ± 204.4 (5.57 ± 1.98)	.057
Rate of ankle torque decline, D, (Nm/sec)*	57.2 ± 27.2 (0.56 ± 0.24)	63.1 ± 23.3 (0.65 ± 0.25)	.085
Total reaction time, Δt, (msec)	100 ± 18	127 ± 16	<.001
Muscle response latency (msec)	73 ± 22	98 ± 16	<.001
Electromechanical delay (msec)	27 ± 14	29 ± 15	.316

Notes:
Cell entries show mean ± one SD.
* Cell entries show mean variable values followed in parentheses by mean normalized values, where normalized value = variable value/[body mass (in kg) x body height (in m)].

2.9 Appendix

In dynamic recovery trials, we set each subject's baseline COP position (and thus ankle torque) to the value typically observed during quiet upright stance (when no tether or lean existed). If we neglect the small contribution to equilibrium of the horizontal force at the foot-floor interface [4], the ankle torque during quiet standing (with no tether applied) (T_o) is given by $T_o = BW \times COP = BW[5/9BH \times \sin\theta_o]$, where BW is body weight in N, COP is the horizontal position of foot centre-of-pressure measured in the anterior-posterior direction with respect to the ankle in m, $5/9BH$ is the height of the body's centre-of-mass above the ground in m [20], and θ_o is the body's angle of

inclination during standing in deg. Therefore, we could individually set the target COP for each subject based on her body height if we knew a typical value for θ_o during standing. In pilot experiments involving 4 elderly subjects (aged 71-82 years) we measured an average θ_o of 5.2 deg, based on an average body height of 1.60 m and COP position of 0.081 m. Similarly, in pilot experiments with 2 young subjects (mean age = 23) we measured an average θ_o of 3.3 deg, based on an average body height of 1.69 m and COP position of 0.054 m. Therefore, the overall average value of θ_o (combining young and elderly subjects) was 4.2 deg. Thus, we set a target $COP = 5/9BH \times \sin(4.2^{\circ})$, and instructed subjects to adjust the anterior-posterior excursion of their COP to within ± 10 mm of this target value.

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Chapter 3
Postural Steadiness during Quiet Stance Does Not Associate with Ability to Recover Balance in Older Women[2]

3.1 Abstract

Fall risk depends on ability to maintain balance during daily activities, and on ability to recover balance following a perturbation such as a slip or trip. It is unclear whether similar neuromuscular variables govern these two domains of postural stability. To address this issue, we examined the association among elderly women between postural steadiness during quiet stance and ability to recover balance during a simulated trip. Twenty-five elderly women (mean age 78 ± 7 (SD)yr) participated in balance experiments. We measured the amplitude (RANGE, RMS), velocity (MVEL), and frequency (MFREQ) of centre-of-pressure displacement (postural steadiness) in the anterior-posterior (AP) and medial-lateral (ML) directions during quiet stance with eyes open on a rigid support surface, and with eyes closed on a foam support surface. We also measured the maximum forward inclination angle (θ_{max}) where subjects could be released and recover balance using the ankle strategy, and corresponding values of reaction time, rate of ankle torque generation, and peak ankle torque. Balance recovery variables were not strongly or consistently correlated with postural steadiness variables. θ_{max} was significantly correlated with only three postural steadiness variables: MFREQ-

[2] The following chapter will be submitted to the Journals of Gerontology: Medical Sciences under the co-authorship of Dr. Stephen Robinovitch.

AP during quiet stance on a rigid surface with eyes open (r=0.36, p=.041), and RMS-ML and MVEL-ML during quiet stance on a foam surface with eyes closed (r=0.41, p=.038 and r=0.49, p=.015). Rate of ankle torque generation was moderately correlated with RANGE, RMS, and MVEL of centre-of-pressure displacement in the AP and ML directions (r=0.48-60) during quiet stance on a foam surface with eyes closed. Reaction time and peak ankle torque during balance recovery were not significantly associated with any postural steadiness variables. These results suggest that postural steadiness during quiet stance is not predictive of balance recovery following a trip. This may be because postural steadiness during quiet stance is controlled in part by anticipatory strategies while balance recovery following a perturbation is governed largely by reactive strategies. Therefore, clinical balance assessments and fall prevention programs should target each of these components of postural stability.

Key words: accidental falls, slips and trips, postural balance, reaction time, biomechanics, aging, hip fracture.

3.2 Introduction

Falls are the leading cause of injury and injury-related death in the elderly (1). While risk for falls associates with a variety of sensory, motor, cognitive, and psychosocial variables, it depends ultimately on ability to maintain balance during daily activities, and on ability to recover balance following a perturbation such as a slip or trip. Ability to maintain balance during daily activities is often characterized by measuring postural steadiness during quiet stance, while ability to recover balance following a perturbation is commonly assessed through measuring performance after a simulated slip or trip.

A few studies have reported a lack of association between postural steadiness during quiet stance and ability to recover balance following a perturbation (2, 3). There are at least two possible explanations for this lack of association. First, it seems there are differences in the control strategies for quiet stance and balance recovery. The control of quiet stance has been explained through feedback models (4, 5) and stiffness control (6, 7), and recent evidence also suggests that quiet stance is controlled at least in part by anticipatory strategies, whereby modulation of muscle activity in the ankle extensors precedes variations in centre-of-mass (COM)position (8, 9). Balance recovery ability, on the other hand, is governed largely by reactive mechanisms, whereby muscle activity is triggered by somatosensory information in response to a postural perturbation (10, 11). Second, balance recovery would seem to tax the neuromuscular system more heavily than quiet stance by requiring closer to threshold values for reaction time, peak torque, and rate of torque generation.

On the other hand, comparisons between postural steadiness during quiet stance and ability to recover balance after a perturbation may have been complicated by fundamental differences in the mechanics of the balance tasks. While quiet stance is dominated by ankle strategies (8), researchers who have examined the association between postural steadiness and ability to recover balance have commonly used balance recovery tasks that involve stepping strategies (3). Stepping strategies require substantial hip and knee mechanics, in addition to ankle mechanics. In the current study, we deliberately selected two "feet-in-place" tasks that were similar in their kinematics and that should be governed by similar muscle groups and sensory organs. The main

difference between the quiet stance and balance recovery tasks selected for the current study was the absence or presence of a perturbation.

Thus, the purpose of this study was to determine whether, among elderly women, there is association between variables related to postural steadiness during quiet stance and variables related to ability to recover balance using an ankle strategy. To address this question, we conducted balance experiments with 25 elderly women (mean age 78 ± 7 (SD)yr). We measured the amplitude (RANGE, RMS), velocity (MVEL), and frequency (MFREQ) of centre-of-pressure (COP) displacement (postural steadiness) in the anterior-posterior and medial-lateral directions during quiet stance with eyes open on a rigid support surface, and with eyes closed on a foam support surface. We also measured the maximum forward inclination angle (θ_{max}) where subjects could be released and recover balance using the ankle strategy, and corresponding values of reaction time, peak ankle torque, and rate of ankle torque generation. Correlations between postural steadiness variables and balance recovery variables were therefore used to determine the association between these two components of postural stability.

3.3 Methods

3.3.1 Subjects

Twenty-five community-dwelling elderly women of mean age 78±7 yr participated in the study. Subjects were recruited from an education-based fall prevention program that operated in local seniors' centres between September 2002 and December 2003. Subjects reported at least one fall in the 18 months prior to their participation in our study.

Subjects were initially interviewed and excluded if they had conditions that would prevent them from performing the experiments, or possessed known risk factors for falls or balance impairments. These included inability to stand unassisted for 10 minutes, terminal illness, blindness, Parkinson's disease, stroke, peripheral neuropathy, vertigo or dizziness, and use of psychotropic medications (12). Subjects were excluded after an on-site evaluation if they had moderate to severe dementia as indicated by a Mini Mental State Exam score (13) less than 21, Snellen visual acuity score with corrective lenses worse than 20/15 at 5 feet, gross instability when standing on foam with eyes closed, or impaired big toe position sense or monofilament sensation on the dorsum of the foot. Each subject was paid $10/hour for her participation. Each subject provided informed written consent, and the experiment was approved by the University Research Ethics Board at Simon Fraser University.

3.3.2 Balance Recovery Experiment

In balance recovery trials, the subject stood on a force plate with her feet a comfortable width apart and arms crossed against her chest. We used a horizontal tether and chest harness to incline the subject into a stationary forward leaning position (Fig.3-1A). We instructed the subject to rise into a vertical standing position by contracting the muscles spanning the ankles and keeping the knees and hips fully extended, a technique called the "ankle strategy" (14, 15). The heels were allowed to leave the ground. To train appropriate body posture and movement, each subject was given three to six practice trials where she leaned into the tether, allowed it to hold part of her body weight, and returned to an upright vertical stance.

We then measured the maximum recovery angle from which subjects were able to recover balance (a stable upright vertical stance) after the tether was suddenly released, which caused a near-step increase in gravitational torque acting to rotate the body downward (tether release time ~15ms). The first trials involved small lean angles of ~2°, where the subject could recover easily. We then iteratively adjusted the length of the tether and the corresponding lean angle until we identified the maximum recovery angle (with a resolution of 7 mm in tether length, and ~0.3° in lean angle), beyond which the subject could no longer recover balance in at least three of five repeated trials. To increase the unexpectedness of the perturbation, the investigator inserted a random time delay of 1-10 sec between receiving a "ready" cue from the subject and the time of release. Before each trial, the subject was instructed to maintain her gaze forward and at eye level on a black X attached to the wall approximately 10 feet in front of her.

Subjects were barefoot during the trials and wore spandex (or similar) shorts and shirt. Rest breaks of approximately 30 sec duration were provided between trials, and 5 minute sitting breaks were provided every 10-15 trials. During trials, we controlled the magnitude of ankle torque before release, which can affect recovery ability (16). Since ankle torque scales with body height and body weight (16), we aimed to keep the ratio of ankle torque before release divided by the product of subject body height and body weight consistent across subjects and close to its average baseline value measured during quiet standing. This was achieved by using a computer monitor to provide visual feedback to the subject and investigator of foot COP position (an indicator of baseline ankle torque). We set a target COP position for each subject so that the ratio of COP

divided by body height matched (to within ± 10 mm) the average value measured during quiet stance in a subset of six subjects.

During each trial we measured the magnitude and point of application of foot-floor reaction forces and moments at a rate of 960 Hz with a force plate (model 6090H, Bertec, Worthington, OH). We also measured body segment movements with a 60 Hz, seven-camera motion measurement system (ProReflex, Qualisys Inc., Glastonbury, CT) that recorded the position of 16 markers attached to the skin and clothing overlying the right and left fifth metatarsal, lateral malleolus, lateral femoral condyle, anterior superior iliac spine, pisiform bone, radial head, acromion process, and the head and sacrum. The instant of tether release was detected as the onset of a sharp decline in the tension (≥ 2 N) measured by a load cell (Sensotec, model 31) located in series with the tether.

We calculated the body lean angle $\theta(t)$, defined as the angle from the vertical to a line connecting the midpoint of the two lateral malleolus markers to the midpoint of the two acromion markers (16) (Fig.3-1A). We also calculated time-varying ankle plantarflexor torque $T_a(t)$ based on the location and magnitude of vertical and horizontal components of foot reaction force (16). The baseline ankle torque before release was calculated as the average value of $T_a(t)$ over the 500 msec preceding release.

From each of the three maximum recovery trials, we calculated four outcome variables to characterize the balance recovery response. We calculated (a) the maximum recovery angle (θ_{max}), defined as the average value of $\theta(t)$ over the 500 msec interval preceding tether release; (b) the peak ankle torque generated during balance recovery (Fig.3-1B); (c) the reaction time (Fig.3-1B), defined as the interval between tether release

and the instant $T_a(t)$ exceeded baseline ankle torque by 5 Nm (always outside baseline variability); and (d) the rate of ankle torque generation following release (Fig.3-1B), defined as the slope of a straight line joining torque-time values at the instant $T_a(t)$ exceeded baseline ankle torque by 5 Nm to the instant $T_a(t)$ equaled 85% of the difference between baseline ankle torque and peak ankle torque. For rate of ankle torque generation, the 85% value was chosen instead of peak ankle torque because it always reflected a point on the initial smooth rise of the torque time curve, while peak ankle torque was sometimes offset from the initial rise due to small oscillations in $T_a(t)$.We normalized values of baseline ankle torque, peak ankle torque, and rate of ankle torque generation by the product of body mass (in kg) multiplied by body height (in m). Values of θ_{max}, peak ankle torque, reaction time, and rate of torque generation used in statistical analyses were averages over three repeated trials for each subject.

3.3.3 Postural Steadiness Experiment

Each subject also participated in measures of postural steadiness during quiet stance under various support surface and vision conditions. In these measures, the subject was barefoot and instructed to stand "as still as possible" near the centre of a force plate with her feet a comfortable width apart, arms at her sides, and vision directed straight ahead (Fig.3-2A). Trials were acquired with the subject standing on the rigid ground with her eyes open (EOR), and on foam (open cell foam rubber measuring 42 cm x 31 cm x 10 cm with density 44.1 kg/m^3) with her eyes closed (ECF). These two conditions of quiet stance were chosen specifically because of their similarities with the balance recovery trials. In the EOR condition, the sensory information available most closely matched that

available in the balance recovery task (i.e., visual, somatosensory, and vestibular feedback were available). In the ECF condition, deprivation of visual and somatosensory information provided a challenging balance task that may have elicited recovery-type responses. Trials were 15 seconds in duration and a rest break of approximately 30 seconds was provided between conditions. The EOR trial always preceded the ECF trial.

We calculated 4 measures of postural steadiness from the COP sway in the anterior-posterior (AP) and medial-lateral (ML) directions (using formulae by Prieto et al., 1996 (17)) (Fig.3-2B). These were peak-to-peak range of sway distance (RANGE), root mean square distance of sway from the mean COP location (RMS), mean velocity of sway (MVEL), and mean frequency of sway (MFREQ) (2, 17).

In all trials, we used a force plate (model 6090H, Bertec, Worthington, OH) to measure (at 540 Hz) the position of the COP between the feet and the ground. The COP time series were then filtered using a fourth order Butterworth low pass filter with 5 Hz cut-off frequency, and the AP and ML components were computed.

3.3.4 Data Analysis

After confirming normality of data, we used Pearson product moment correlation coefficients to test for association between balance recovery variables and postural steadiness variables. A total of five data points (and no more than one for a given variable) from the postural steadiness data were identified as outliers (defined as values >3SDs above or below the mean) and removed before statistical analyses were conducted. All analyses were conducted with statistical analysis software (SPSS 11.0, Inc., Chicago, IL) using a level of significance of $p<0.05$.

Four subjects were unable to complete the ECF condition without opening their eyes or using the support of an experimenter's hand. The data from these four subjects were not used in the analysis of the ECF condition. Furthermore, equipment malfunction prevented us from recording the postural steadiness data from one subject. Therefore, data from this subject were excluded, yielding a final sample size of n=24.

3.4 Results

3.4.1 Correlation Between Balance Recovery Variables and Postural Steadiness Variables

Means and standard deviations for each of the balance recovery and postural steadiness variables are presented in Table 3-1. Balance recovery variables were not strongly correlated with postural steadiness variables. Maximum recovery angle (θ_{max}) was significantly correlated with only three postural steadiness variables (Table 3-2, Fig.3-3). θ_{max} was correlated with MFREQ-AP in the EOR condition (r=0.36, p=.041), and with RMS-ML and MVEL-ML in the ECF condition (r=0.41, p=.038 and r=0.49, p=.015). Reaction time and peak ankle torque during balance recovery were not significantly associated with any postural steadiness variables. Rate of ankle torque generation was moderately correlated with RANGE, RMS, and MVEL of postural sway in the AP and ML directions in the ECF condition (r=0.48-0.60), as well as with RMS-AP in the EOR condition (r=-0.42, p=.019).

3.5 Discussion

In this study we found that among community-dwelling elderly women, who had a history of falls, steadiness during quiet stance and ability to recover balance with the

ankle strategy were not associated. These results suggest that postural steadiness during quiet stance is not predictive of balance recovery following a trip.

Our results are consistent with previous investigations that examined associations between postural steadiness during quiet stance and ability to recover balance following a perturbation. Maki et al. (2) measured induced sway during continuous platform translations as an index of ability to recover balance following a destabilizing perturbation, and found that it was not associated with postural steadiness during quiet stance. Further, Owings et al. (3) measured ability to recover balance with a single step following tether release and ability to recover balance after an unexpected trip during gait and found that neither measure was associated with postural steadiness during quiet stance.

Larger values for sway RANGE, RMS, MVEL and MFREQ generally indicate poorer stability (2, 18-21); thus, we would expect negative correlation coefficients between postural steadiness variables and balance recovery variables. (2). Instead, we found that RMS-ML and MVEL-ML of sway during quiet stance on foam with eyes closed were positively correlated with maximum recovery angle, suggesting that larger amplitudes and velocities of sway under conditions of sensory deprivation associate with improved ability to recover balance following a perturbation. A similar phenomenon was observed, whereby elderly subjects who were able to recover their balance after tripping demonstrated faster speeds of lateral COP displacement during quiet stance than subjects who could not recover their balance after a trip (3). The present study suggests this relationship may relate to the capacity to generate ankle torque rapidly. Previous literature has shown that ability to recover balance using the ankle strategy is enhanced

by faster rates of ankle torque generation (16, 22). The current results suggest that rate of ankle torque generation also associates with postural steadiness during quiet stance on foam with eyes closed. Increased RANGE, RMS, and MVEL of sway in the AP and ML directions during quiet stance associated with increased rates of ankle torque generation during balance recovery trials.

One reason for the lack of association between postural steadiness during quiet stance and ability to recover balance following a simulated trip is that these two components of postural stability appear to be governed by different neuromuscular and sensory variables. Previous research has demonstrated that ability to recover balance using the ankle strategy is associated with peak ankle torque, rate of ankle torque generation, and reaction time, indicating the importance of ankle strength and speed-of-response to balance recovery performance (16). In contrast, the current results show that postural steadiness during quiet stance does not associate with these same neuromuscular variables. These results imply that balance recovery taxes the neuromuscular system more heavily than quiet stance.

Futhermore, it seems there are differences in the control strategies for quiet stance and balance recovery. Some have suggested that quiet stance is controlled primarily by feedback mechanisms (4, 5, 23). For instance, Peterka (23) proposed a feedback model whereby sensory information is used continuously to sense deviations in body posture away from upright and drive commands to various muscles to generate corrective torques. These corrective torques are proportional to body sway angle (position), sway angular velocity (derivative), and the integral of body sway angle (integral) and are applied at a time delay that accounts for conduction, processing, and muscle activation

times. With this feedback model, Peterka generated stabilogram diffusion functions (24) with features very similar to actual stabilogram diffusion functions calculated from experimental data.

In contrast to a feedback model, Winter et al. (6, 7) introduced the concept of stiffness control for quiet stance. They suggested the CNS sets ankle joint stiffness through appropriate muscle tone, and in turn, the ankle muscles act like springs to cause the COP to move in phase with the COM. Stiffness control provides almost instantaneous corrective responses as it does not rely on reactive mechanisms such as sensory input from the visual, vestibular, or somatosensory systems. Winter et al. argued that a reactive control schema (feedback system) would be too slow, given the time required for afferent and efferent processing and muscle responses, to explain the short 4 ms lag between the COM and the COP that is observed experimentally. Winter et al. further argued against reactive control because head accelerations, joint displacements, and joint velocities during quiet stance are usually below threshold values for activating vestibular and somatosensory receptors.

Stiffness control of upright stance has subsequently been heavily debated. Morasso and Schieppati (25) and Morasso and Sanguineti (26) argued that muscle stiffness alone cannot stabilize the human upright posture, and furthermore that the tight temporal tracking of COM by COP is a consequence of the dynamics of an inverted pendulum and not dependent on the control strategy for quiet stance. While they acknowledged the important contribution of ankle stiffness in stabilizing upright stance, they proposed that stiffness control was used together with active control of sway, whereby sensory information drives the modulation of muscle activation during stance in

anticipation of body sway. Evidence from Gatev et al. (8) and Masani et al. (9) supports the role of active anticipatory strategies in the control of upright posture. During quiet stance, they observed that variations in muscle activity in the ankle extensors preceded variations in the COM position (8, 9). Velocity information, likely derived from somatosensory afferents, appeared to play a critical role in mediating this anticipatory control (9). Thus, active control requires the availability of reliable sensory information on which to base predictive processes (or accurately "tune" an internal model), and this is consistent with clinical findings of altered sway patterns and increased reliance on ankle stiffness to control quiet stance in specific patient populations with sensory deficits. Active control does not introduce long delays since sensory information is used in an anticipatory fashion; therefore, active control is still compatible with the small time lag observed between COM and COP.

While the control of quiet stance is not yet fully understood, the evidence provided above suggests that it is governed at least partly by anticipatory strategies. In contrast, balance recovery ability is governed largely by reactive mechanisms, whereby muscle activity is triggered by somatosensory information in response to a postural perturbation (10, 11). This difference in the control of quiet stance and balance recovery may account for the lack of association between postural steadiness during quiet stance and ability to recover balance after a perturbation.

It seems therefore that fundamental differences in the control of postural steadiness during quiet stance and balance recovery following a perturbation (predictive vs. reactive) and in the level of neuromuscular demands for each of these tasks (low for quiet stance compared to balance recovery) account for the lack of association between

postural steadiness during quiet stance and ability to recover balance following a perturbation. If there was any overlap between postural steadiness and balance recovery ability, we should have been able to detect it since we limited our investigation to two tasks that were dominated by ankle strategies and therefore had similar mechanics.

Our results support the need to measure both balance recovery ability and postural steadiness in balance assessments of the elderly. While previous studies have shown that postural steadiness during quiet stance is a significant predictor of fall risk, the association is moderate at best (18, 27). Furthermore, performance of the postural control system is task specific, such that simple standing balance tests may not detect balance impairments in older adults (28-30). For example, such tests are unable to assess the strength or speed with which an individual responds to destabilizing perturbations, which should contribute to fall risk (2).

We acknowledge certain limitations of this study. First, we had a relatively small sample size. However, with 24 subjects we had approximately 80% power to detect an r of 0.50 (R^2=0.25) at the α =0.05 level of significance (31, 32); anything smaller is of questionable clinical relevance. A further limitation is that we focused on postural steadiness and balance recovery from a standing position, using the ankle strategy. Other investigators have previously examined balance recovery by stepping, and we felt it was important to equate the mechanics of the postural steadiness and balance recovery tasks as much as possible to compare the amount of overlap in these two components of postural stability.

In summary, we found that, among community-dwelling elderly women, postural steadiness during quiet stance and ability to recover balance following a simulated trip

were not associated. This lack of association may be because postural steadiness during quiet stance is controlled in part by anticipatory strategies while balance recovery following a perturbation is governed primarily by reactive strategies. Therefore, clinical balance assessments and fall prevention programs should target both components of postural stability.

3.6 Acknowledgements

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3.7 Figures

Figure 3-1 Balance recovery experiment.
(A) Subjects were held by a tether in an initially inclined position and instructed to recover a stable upright posture by contracting their ankle muscles. Their "maximum recovery angle" (θ_{max}), was the maximum forward lean angle (θ_o) where they could accomplish this task when the tether was released. (B) Measured neuromuscular variables. We detected tether release (shown by the earliest vertical dashed line) by a sharp decline in tether force. Following release, there was a reaction time before the onset of increased ankle torque generation (shown by the second vertical dashed line). Ankle torque was generated at a specific rate (rate of ankle torque generation), and reached a peak value (peak ankle torque) before declining.

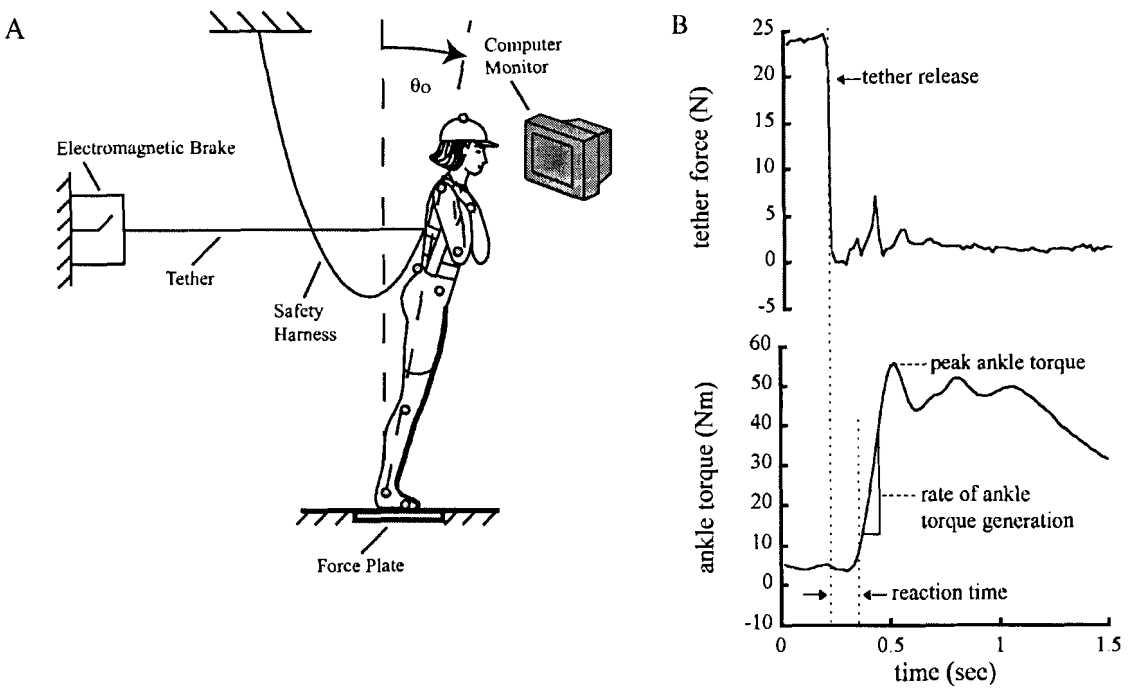


Figure 3-2 Postural steadiness experiment.
(A) Subjects were instructed to stand "as still as possible" for each trial. (B) The stabilogram from one subject for quiet stance on a rigid surface with eyes open (left trace) and for quiet stance on a foam surface with eyes closed (right trace). The x-axis indicates sway of the COP in the medial-lateral (ML) direction. The y-axis indicates sway of the COP in the anterior-posterior (AP) direction.

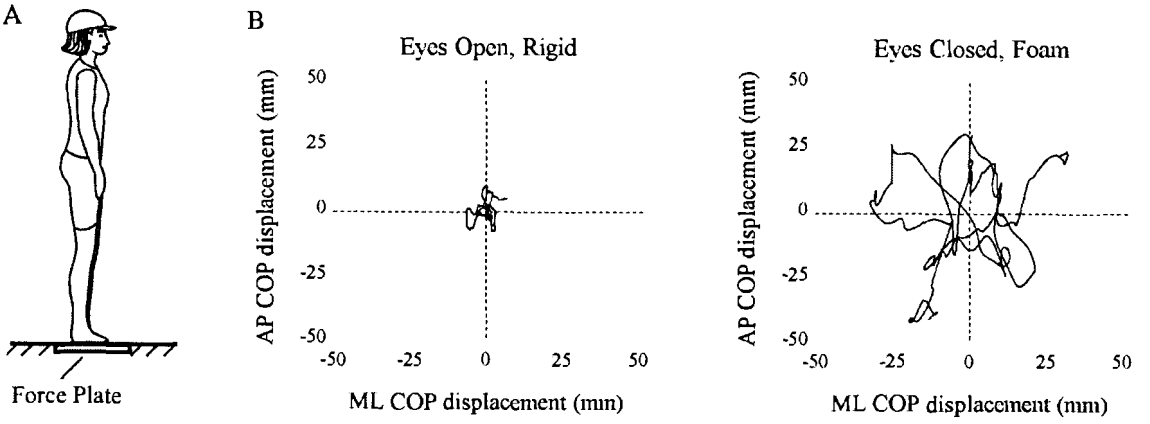
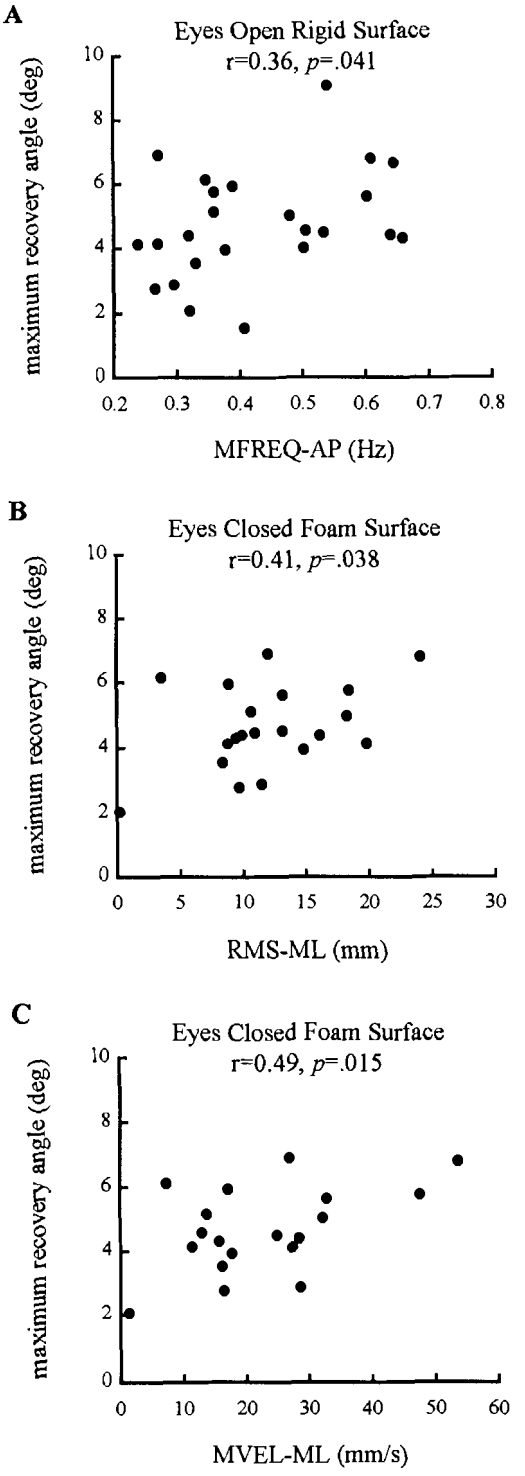


Figure 3-3 Correlation between maximum recovery angle and postural steadiness variables. (A) MFREQ-AP during quiet stance on a rigid surface with eyes open, (B) RMS-ML during quiet stance on a foam surface with eyes closed, and (C) MVEL-ML during quiet stance on a foam surface with eyes closed.



3.8 Tables

Table 3-1 Means and standard deviations (Mean ± SD) for postural steadiness and balance recovery variables.

	Eyes Open Rigid	Eyes Closed Foam
	Mean ± SD	Mean ± SD
Postural Steadiness		
RANGE-AP (mm)	17.0 ± 6.2	66.4 ± 19.4
RANGE-ML (mm)	8.8 ± 4.9	54.8 ± 26.9
RMS-AP (mm)	3.6 ± 1.2	15.0 ± 4.8
RMS-ML (mm)	2.2 ± 1.4	12.1 ± 5.5
MVEL-AP (mm/s)	6.3 ± 1.7	39.4 ± 14.5
MVEL-ML (mm/s)	4.3 ± 2.2	23.0 ± 12.8
MFREQ-AP (Hz)	0.43 ±0.14	0.62 ± 0.20
MFREQ-ML (Hz)	0.50 ± 0.23	0.42 ± 0.12
Balance Recovery		
Maximum recovery angle (deg)	4.77 ± 1.69	
Reaction time (msec)	125 ± 13	
Peak ankle torque (Nm/(kg·m))	0.95 ± 0.16	
Rate of ankle torque generation (Nm/(kg·m·s))	5.40 ± 1.81	

Notes:
AP = Anterior-Posterior
ML = Medial-Lateral

Table 3-2 Correlation between postural steadiness variables and balance recovery variables.

	Maximum recovery angle		Reaction time		Peak ankle torque		Rate of ankle torque generation	
	r	*p*	r	*p*	r	*p*	r	*p*
Eyes Open - Rigid								
RANGE-AP (mm)	0.07	.378	0.18	.197	-0.16	.229	-0.34	.055
RANGE-ML (mm)	-0.02	.472	0.01	.483	-0.11	.312	0.02	.472
RMS-AP (mm)	-0.14	.260	0.15	.237	-0.25	.119	-0.42	.019*
RMS-ML (mm)	0.20	.175	0.00	.500	-0.04	.424	0.00	.494
MVEL-AP (mm/s)	-0.22	.159	0.06	.400	-0.31	.078	-0.25	.121
MVEL-ML (mm/s)	0.12	.285	0.07	.370	-0.04	.423	-0.06	.387
MFREQ-AP (Hz)	0.36	.041*	-0.09	.344	0.22	.147	0.26	.114
MFREQ-ML (Hz)	-0.23	.146	-0.10	.323	0.02	.463	-0.07	.374
Eyes Closed - Foam								
RANGE-AP (mm)	0.35	.065	0.09	.359	0.12	.303	0.54	.007*
RANGE-ML (mm)	0.37	.054	0.05	.412	0.17	.232	0.48	.015*
RMS-AP (mm)	0.31	.092	0.10	.333	0.09	.356	0.55	.006*
RMS-ML (mm)	0.41	.038*	0.06	.407	0.21	.192	0.52	.009*
MVEL-AP (mm/s)	0.28	.123	0.22	.182	0.29	.116	0.57	.006*
MVEL-ML (mm/s)	0.49	.015*	0.03	.456	0.35	.063	0.60	.003*
MFREQ-AP (Hz)	0.13	.292	-0.02	.474	0.15	.263	0.37	.057
MFREQ-ML (Hz)	0.29	.114	0.02	.468	0.29	.115	0.26	.146

Notes:
AP = Anterior-Posterior
ML = Medial-Lateral
r = Pearson product moment correlation coefficient
p = significance level
* = significant correlation at $p<.05$

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Chapter 4
General Conclusions

4.1 Overview

Falls and their related injuries are a serious health concern among the elderly population, and reducing their incidence and severity is a national health priority in Canada. This thesis examined two different aspects of balance that contribute to fall risk: ability to recover balance following a destabilizing perturbation, and postural steadiness during quiet stance.

In Chapter 2, it was shown that age differences in ability to recover balance with a feet-in-place strategy depend on variables related to both strength and speed-of-response. The maximum static recovery angle, which depends primarily on strength, was 19.6% smaller in elderly than young, illustrating the effect of age-related declines in strength. The maximum dynamic recovery angle, which depends on both strength and speed-of-response, was 36.1% smaller in elderly than young, illustrating the additional effect of age-related declines in speed-of-response. The effect of age-related declines in strength were shown further by the differences in peak ankle torque generated by young and elderly subjects. In static recovery trials, elderly subjects generated 4.8% less peak ankle torque than young, and in dynamic trials, they generated 7.7% less peak ankle torque. Perhaps even more striking were the age-related differences in speed-of-response variables measured during dynamic recovery trials. Reaction time was 27.0% slower in elderly than young, due to a significantly longer muscle response latency, and rate of

ankle torque generation was 15.6% slower. These results were supported by a correlation analysis, which revealed that ability to recover balance after a destabilizing perturbation was associated with peak ankle torque, reaction time, and rate of ankle torque generation. These results suggest that age-related declines in ability to recover balance associate with variables related to strength and speed-of-response, which complements growing clinical evidence that exercise programs to prevent falls in the elderly should incorporate balance and agility training, in addition to strength training.

In Chapter 3, it was shown that, among community-dwelling elderly women who had a history of falls, postural steadiness during quiet stance and ability to recover balance following a simulated trip were not associated strongly or consistently. Maximum recovery angle from balance recovery trials was significantly correlated with only three postural steadiness variables. Reaction time and peak ankle torque during balance recovery were not significantly associated with any postural steadiness variables. Rate of ankle torque generation was moderately correlated with seven of 16 postural steadiness variables. This lack of association may be because postural steadiness during quiet stance is controlled in part by anticipatory strategies while balance recovery following a perturbation is governed primarily by reactive strategies. Therefore, clinical balance assessments and fall prevention programs should target both components of postural stability.

4.2 Limitations

In balance recovery studies there are often tradeoffs between measuring the effects of biomechanical contributions to performance and measuring natural responses. One of the limitations of this study was that we examined only forward loss-of-balance

and instructed subjects to recover using the ankle strategy rather than allowing them to choose a natural response (e.g., feet-in-place or stepping responses). This is substantially different from a real-life loss of balance episode. Our intention was to compare the relative contributions of strength versus speed-of-response variables to age-related declines in ability to recover balance. To do so, we felt it was important to strictly limit the recovery strategy. If we had simply instructed subjects to prevent a fall, we anticipated that it would be difficult to define an index of recovery ability, and separate the contributions to performance of behavioural (e.g., strategy, fear, efficacy, and habit) versus biomechanical (e.g., strength and speed-of-response) variables. By providing subjects with explicit instructions on the desired balance recovery strategy, we were able to minimize behavioural influences and clearly identify biomechanical contributions to performance. As a result, we gained an improved understanding of the role of specific declines in physical capacity (biomechanical contributions) in explaining age-related declines in balance recovery performance. The advantage of studies that measure natural responses, on the other hand, is that the results may have greater generalizability to real-life loss-of-balance events. However, it is not usually possible in such studies to separate behavioural and biomechanical contributions to performance. Both types of studies are valuable and provide complementary information. In future research, it would be useful to examine behavioural and neuromuscular contributions to balance declines in the elderly, by comparing naïve and performance-based responses.

Another limitation of balance recovery studies, which limits their application to real-life loss of balance episodes, is that the perturbation is always somewhat expected. In the tether release paradigm used for this thesis, the magnitude and direction of the

balance perturbation were somewhat predictable to the subject. Platform perturbation experiments avoid this problem, as the magnitude and direction of perturbation can be varied without the subject's awareness. However, both tether release and platform experiments are limited by the subject's expectation of balance perturbation. Thus, responses in the natural environment, when loss-of-balance is more unexpected, may be quite different from responses measured in the laboratory, when subjects can focus their attention on balancing and have time to plan movement strategies to accommodate upcoming balance perturbations. Some slipping experiments have minimized expectation of loss-of-balance by inserting random slip trials into a series of normal walking trials (1-3); however, these experiments can also be influenced by subject anticipation (4). In choosing the appropriate paradigm for a set of experiments, it is important to remember the impending fall situation that is being simulated. For example, one of the advantages of the tether release is that it simulates a trip by inducing forward movement of the centre-of-mass (COM) with respect to a stationary base-of-support (BOS). In contrast, platform translations more closely simulate slips, where the BOS cannot be established under the COM.

Despite the aforementioned limitations, balance recovery experiments provide important information to guide the development of improved techniques to prevent falls and fall-related injuries among the elderly. The results presented in Chapter 2 substantially enhance our understanding of the biomechanical and neuromuscular variables that govern age-related declines in postural stability and balance recovery ability. The results presented in Chapter 3 demonstrate there are multiple components to balance and postural stability which the postural control system must regulate. With

effective knowledge exchange strategies, these results should ultimately improve our ability to develop exercise-based fall prevention interventions.

4.3 Future Work

As the population ages, falls and their related injuries will become an increasingly serious health problem affecting the elderly population, and consequently, an enormous economic burden for the Canadian healthcare system. Research has the potential, and perhaps the responsibility, to reduce the incidence and severity of falls and fall-related injuries through identifying their cause and prevention. Given the complex and multi-factorial nature of falls in the elderly, research programs in the area are considerably varied. For example, qualitative researchers are addressing questions such as the social and personal impacts of hip fracture, while biomedical engineers are testing the strength of the proximal femur in realistic fall configurations and developing systems to reduce impact forces at the hip during a fall. Biomechanics researchers are examining mechanisms that account for age-related declines in balance, clinical researchers are finding ways to improve balance and mobility and reduce fall risk through exercise interventions, and health promotion researchers are learning how to implement fall prevention programs to ensure their sustainability within the community. Given the numerous different perspectives by which we can address falls prevention, we are beginning to see the emergence of transdisciplinary research teams that combine expertise in areas such as biomedical engineering, biomechanics, orthopaedics, neuroscience, physiology, exercise, bone health and osteoporosis, geriatrics, sociology, and psychology. In the future, such teams will be critical to effectively address falls prevention.

4.3.1 Biomechanics Research

Biomechanics research has an important role to play in understanding the cause and prevention of falls and fall-related injuries. Schultz (5) makes a critical point when he states:

> Changes in cognition and central nervous system processing; changes in visual, vestibular and proprioceptive sensing abilities; the effects of physical inactivity and disuse; the effects of neurologic and musculoskeletal pathologies; the effects of medications; and the effects of motivation and fear all ultimately express themselves as changes in the biomechanics of physical-task performance.

Thus, a large proportion of the physiological and psychological changes that occur with age and increase one's fall risk are manifest as impairments in physical-task performance. Biomechanics research, which is able to precisely quantify physical task-performance through the study of human motion and the forces that cause motion, is therefore positioned perfectly to provide new insight into the cause and prevention of a vast array of mobility problems that plague the elderly population, including balance impairments and falls.

By quantifying physical capabilities, one role of biomechanics research is to help explain clinical observations of balance in the elderly population. For example, why can elderly individuals not recover from the same magnitude of perturbations as young individuals? Why do elderly tend to sway more during quiet stance? Why can elderly not stand on one foot, or stand in tandem stance as long as young? Answers to these questions, garnered through biomechanical studies, can help direct the development and evaluation of exercise-based fall prevention programs. Without a clear understanding of the mechanisms underlying balance and mobility problems in the elderly, which is

afforded by biomechanics research, it is not difficult to imagine that the efficiency of clinical trials research would suffer, as a vast number of interventions would be necessary to identify those with the most benefit. Thus, there is an important link between laboratory and clinical research. Clinical work benefits from the evidence base established through laboratory research, and laboratory research often gains inspiration from real world clinical problems. Although it is a large undertaking, future research needs to strengthen this link, so that knowledge gained in the laboratory has a positive impact on the health of aging Canadians.

4.3.2 Age-Related Changes in Balance and Falls

The conceptual model of fall risk that underpins this thesis is that an individual's risk for falling depends ultimately on two factors: 1) the frequency of loss-of-balance episodes, and 2) the ability to recover balance after such events. How often an individual loses balance (or in other words, their ability to maintain balance) depends partly on their steadiness during quiet stance. Cross sectional studies have confirmed that elderly fallers have greater sway than elderly non-fallers (6,7) and than young subjects (7-11), and prospective studies have shown that greater amounts of sway during quiet stance increases risk for falls (12,13).

A small number of studies have examined differences in balance recovery ability between elderly fallers and non-fallers (6,12), but work in this area is still lacking. In future studies, it would be useful to examine differences in balance recovery ability between fallers and non-fallers across a number of different perturbation paradigms (e.g., tether release, platform translation, hold and release, and postural stress test). Given the results of the current thesis, we hypothesize that differences in recovery ability between

elderly fallers and non-fallers associate more strongly with variables related to speed-of-response than with those related to strength. If cross-sectional work suggests differences exist between elderly fallers and non-fallers, a prospective trial, with falls, and even fall-related injuries, as the outcome(s) would be beneficial. A prospective trial would determine how well balance recovery performance could predict falls and related injury. Since ability to recover balance determines whether a loss-of-balance episode results in a fall, it is foreseeable that balance recovery performance would explain additional variance in predictive models of fall risk, on top of that already explained by postural steadiness during quiet stance. Further support for this idea comes from Chapter 3, where it was shown that postural steadiness during quiet stance is not associated with ability to recover balance, suggesting that these two domains of balance stress different components of postural stability.

One of the main findings in Chapter 2 was that muscle response latency was slower in elderly than young. An important goal for future studies is to identify the neurological mechanisms underlying the age-related slowing of muscle response latency. It is possible that lengthened muscle response latencies in the elderly may be due primarily to peripheral sensory loss and afferent and efferent conduction impairment(s) (14,15). Alternatively, lengthened muscle response latencies in the elderly may be due primarily to central processing deficits (16). Thus, to improve our understanding of age-related declines in ability to recovery balance, future research should assess how proprioceptors, cutaneous afferents, and central processing affect recovery through manipulations of somatsosensory inputs and secondary cognitive tasks. Localizing the neurological cause of age-related slowing of muscle response latency may help to

identify the feasibility of improving reaction time in the elderly and the specific types of interventions that may be most effective.

In the context of the present research, one way to test the central versus peripheral contributions to muscle response latency would be to compare the muscle response latencies of young and elderly subjects on the static recovery task, as long as subjects were instructed to recover their balance as quickly as possible after hearing the auditory cue. Compared to the dynamic recovery task examined in the current work, this experimental paradigm would remove the time required for detection and processing of somatosensory information, and therefore the muscle response latency would be more dependent on central processing. If age-related differences in muscle response latency were observed, it would suggest that central processing delays are at least partly responsible. On the other hand, if there were no age differences in muscle response latency during the static recovery trials, it would suggest that the lengthened muscle response latencies we observed in elderly subjects during dynamic recovery trials are likely due to age differences in peripheral processing of somatosensory information.

We tested a relatively healthy group of elderly subjects in the current study. It is important for future research to determine the relative importance of declines in strength and speed-of-response in ability to recover balance among frail elderly, who are at higher risk for falls and injury. It is reasonable to hypothesize that the observed age-related differences in ability to recover balance would be exacerbated in a group of frail elderly, and muscle weakness and corresponding declines in lower extremity strength may play a more substantial role in explaining declines in ability to recover balance in this frail population.

4.3.3 Exercise

Exercise is an attractive intervention for preventing falls, since the benefits of exercise extend far beyond fall prevention to overall musculoskeletal, cardiovascular, and mental health spectrums. Many studies have shown that exercise can positively affect fall risk factors (17-22), and reduce the number of falls (23-28). Only one study however has shown a reduction in the risk of injurious falls following an exercise intervention (26). Thus, the next large undertaking for falls researchers is to conduct an intervention trial with fall-related injuries, including hip fracture, as the primary outcomes. This type of study would need to be a large-scale, multi-centre trial in order to achieve adequate statistical power, given that fall-related injuries, and hip fracture in particular, are relatively rare events. Evidence suggests that such an intervention should include balance training and muscle strengthening, and should be individually focused, targeted at high risk fallers, and delivered in the home by a health professional (23-28). Given that such an intervention would be time and resource intensive, it would be critical to conduct an economic evaluation to determine cost effectiveness.

In the meantime, it is worthwhile pursuing smaller-scale exercise interventions related to balance recovery ability, which is one risk factor for falls that has not been well-addressed in exercise intervention trials to date. It would be beneficial for future studies to address whether, among elderly fallers, a multi-factorial exercise program can enhance speed-of-response and strength variables, and improve ability to maintain and recover balance. There is still substantial uncertainty surrounding whether reaction time (a speed-of-response variable) is amenable to change through an exercise program, so the results of this type of study would be highly valuable.

4.4 Conclusions

We have identified that age-related differences in ability to recover balance are governed by declines in strength and speed-of-response. These results suggest that fall prevention interventions should target both types of neuromuscular variables. We have also identified, that among community-dwelling elderly women, ability to recover balance is not associated with postural steadiness during quiet stance. These results suggest that traditional balance assessments, which measure steadiness during quiet stance, may be inadequate for fully characterizing an individual's balance capabilities and impairments. Future work should continue to address the cause and prevention of falls with the goal of improving quality of life for our elderly population.

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Appendix A
Research Ethics Approval

SIMON FRASER UNIVERSITY

OFFICE OF RESEARCH ETHICS

BURNABY, BRITISH COLUMBIA
CANADA V5A 1S6
Telephone: 604-291-3447
FAX: 604-268-6785

October 28, 2002

Ms. Dawn Mackey
Graduate Student
Kinesiology
Simon Fraser University

Dear Ms. Mackey:

Re: The Neuromuscular and Sensorimotor Basis of Balance Recovery

I am pleased to inform you that the above referenced Request for Ethical Approval of Research has been approved on behalf of the Research Ethics Board. The approval for this project is for the term of the period of the grant, as defined by the funding agency. If this project does not receive grant support, the term of the approval is twenty-four months from the above date.

Any changes in the procedures affecting interaction with human subjects should be reported to the Research Ethics Board. Significant changes will require the submission of a revised Request for Ethical Approval of Research. This approval is in effect only while you are a registered SFU student.

Your application has been categorized as 'minimal risk" and approved by the Director, Office of Research Ethics, on behalf of the Research Ethics Board in accordance with University policy R20.0, http://www.sfu.ca/policies/research/r20-01.htm. The Board reviews and may amend decisions made independently by the Director, Chair or Deputy Chair at its regular monthly meetings

"Minimal risk" occurs when potential subjects can reasonably be expected to regard the probability and magnitude of possible harms incurred by participating in the research to be no greater than those encountered by the subject in those aspects of his or her everyday life that relate to the research.

Best wishes for success in this research.

Sincerely,

Dr. Hal Weinberg, Director
Office of Research Ethics

c: Stephen Robinovitch, Supervisor
/bjr