NEUROMUSCULAR AND BEHAVIOURAL INFLUENCES ON BALANCE AND FALLS

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ABSTRACT

Falls are the number one cause of accidental injury and exert an especially heavy toll on the elderly. Most falls involve a common sequence of events. First, a particular event results in loss of balance. Second, there is a failed attempt at recovering balance, and finally, once the fall is "recognized" to be inevitable, attempts may be made to lessen its severity. My thesis research focuses on neuromuscular and behavioural aspects of balance maintenance and protective responses during falls. In Chapter 2 and 3, I describe efforts to develop and apply a novel technique to determine how risk for imbalance during daily activities depends on behavioural versus neuromuscular factors. Specifically, I developed the "Reach Utilization Test" to determine whether the tendency to approach imbalance is different between young and elderly women who resided either in nursing homes (Chapter 2) or in the community (Chapter 3). In Chapters 4 to 7, I focus on the measurement and analysis of protective responses during sideways falls. In Chapter 4, I describe the results of experiments to determine whether unexpected sideways falls in young adults elicit a common sequence of responses that might protect against hip fracture. I used a novel experimental paradigm which challenged participants to focus on maintaining balance after experiencing a single large perturbation, which in the vast majority of cases elicited a sideways fall. In Chapter 5, I describe the results of experiments to test whether the ability of humans to alter their body configuration during the fall depends on the time when the response is initiated. I hypothesized that a critical time window exists, beyond which one is unable to avoid hip impact. In Chapter 6, I describe the results of experiments to determine whether individuals are able to accurately recall the details of their falls. I addressed this question by interviewing young adults immediately after they experienced an unexpected sideways falls in our laboratory. Finally, in chapter 7, I describe results from a modelling study to address how fall severity, depend on the cause of the fall (slip vs. trip).

Keywords: biomechanics; injury prevention; falls prevention; hip fracture, balance/posture; fall protective responses, reaction time

Subject Terms: Falls (Accidents) in old age -- Prevention; Hip joint -- Wounds and injuries -- Prevention; Hip Fractures -- prevention & control; Accidental Falls -- prevention & control; Aged.

"Do not go where the path may lead,

go instead where there is no path and leave a trail"

Ralph Waldo Emerson (Philosopher)

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CHAPTER 1 INTRODUCTION

1.1 Epidemiology of falls

Falls and fall related injuries are a major health problem in the elderly population. It is estimated that over 30 percent of community dwelling individuals and 50 percent of nursing home residents over age 65 fall each year (Speechley & Tinetti, 1991). Consequences of falls include fracture of the hip, spine, arm, pelvis, and wrist, head concussions, bruises, and lacerations. However, even when no injury occurs, falls can have psychological effects on the life of the individual. "Fear of Falling" results not only in restricted activity but increased dependency on others and a decrease in social interaction (Howland et al., 1998; Howland et al., 1993). Fall related injuries bring not only suffering to the individuals, but also represent a huge cost to the society. Injuries associated with falls in the elderly account for more than $1 billion dollars in annual health care costs in Canada (National Advisory Council on Aging, 1999). This is three times the cost of injuries related to motor vehicle accidents in this age group. Approximately 90 percent of hip fractures are due to falls (Grisso et al., 1991), and in 1995 alone, there were over 23,000 cases of hip fractures in Canada. These numbers are projected to increase nearly four-fold by the year 2041, given the aging of the population and the fact that fracture risk increases exponentially with age (The Hygeia Group, 1998). Furthermore, approximately 20% of older adults hospitalized for a hip fracture die within a year, and only 50% are able to return to their homes or live independently again (Tinetti, and Williams, 1997; Zuckerman, 1996). Clearly, developing improved strategies

to reduce the frequency and consequences of falls in the elderly is a critical health priority.

1.2 Cause and risk factors for falls

While considerable evidence exists of the circumstances and risk factors surrounding falls, significant challenges still remain in our understanding of their cause and prevention. In biomechanical terms, most falls can be defined as loss of a stable upright posture due to movements (and lack of appropriate corrective actions) which displaces the body's center of gravity beyond its base of support. Most falls have no apparent link to environmental hazards (Morfitt, 1983), but instead result from failed attempts at performing activities of daily living such as walking, turning, rising, and bending (Nevitt & Cummings, 1993b; Parker, Twemlow, & Pryor, 1996; Tinetti, Doucette, & Claus, 1995). While slips and trips are common self-reported causes for falls, similar in frequency are claims of "loss-of-balance," "leg gave way," "changed posture," or "don't know the cause" (Blake et al., 1988; Brocklehurst, Exton-Smith, Lempert Barber, Hunt, & Palmer, 1978; Cumming & Klineberg, 1994). A variety of factors predispose individuals to falls, including medication use, stroke, neurological disease such as Parkinson's syndrome, and impairments in muscle strength, joint movement, balance, gait, vision, hearing, and cognition (see reviews in (Rubenstein, Josephson, & Robbins, 1994; Tinetti, 1994).

1.3 Biomechanics of falls

The mechanics of falling can be divided into four stages (see Figure 1-1; adapted from (W. C. Hayes et al., 1996)): (1) a *balance maintenance stage*, which involves keeping the body's center of gravity within its base of support, (2) an *initiation stage*, which involves loss-of-balance and potential attempts to regain upright posture; (3) a *descent stage*, where movements may be attempted in preparation for landing; and (4) a *contact stage*, where impact occurs between the body parts and the ground, resulting in the generation of reaction forces and absorption and/or dissipation of the body's kinetic energy.

This thesis addresses unexplored aspects of balance maintenance (stage 1) and fall descent mechanics (stage 3), which has traditionally received relatively little attention when compared to the balance recovery and impact stages. In particular, I sought to examine the influence of neuromuscular and behavioural variables on (a) movement strategies to maintain balance during the task of reaching and (b) protective responses to avoid injury in the event of a fall. The factors associated with balance maintenance are introduced in section 1.4 and investigated in chapters 2 and 3. The factors that influence the descent stage are introduced in section 1.5 and investigated in chapters 4, 5, 6, and 7.

1.4 Balance maintenance

1.4.1 Role of perception in movement planning

Two sets of constraints determine our range of possible movement at a given instant. The first is environmental, and includes the location and structural properties of objects near or contacting our body. The second is intrinsic, and includes our muscle

strength, body size, flexibility, and movement speed. A reasonable assumption is that to plan successful movements we must accurately perceive such constraints, and develop motor plans which are compatible with these. While for many movements we are unaware of the computations involved in this, we become particularly aware of them while planning challenging movements, such as jumping a large puddle without getting our feet wet, or crossing the street with traffic approaching in the distance. For elderly individuals, this might extend to more common activities, such as stair climbing, rising from a chair, reaching to pick up an object, or turning a corner.

The majority of data on the accuracy of self-perceived motor abilities has been generated from research on "affordances," which Gibson defined as the functional utility the environment offers an individual at a given instant (Gibson, 1966, 1979). The affordance provided by an object is determined not only by its geometry, material properties, and location in space, but also by the physical capabilities and geometry of the individual. In terms of motor control, affordances determine not only one's range of feasible movements, but also the need to switch from one movement pattern to another to accomplish a specific task. Investigators have shown that such transitions, or critical points (Kelso, 1995; Warren, 1984; Warren, and Whang, 1987), occur at specific ratios of object size to body size. For example, the critical height one perceives as climbable scales with leg length (Mark, 1987; Meeuwsen, 1991; Warren, 1984), and the critical width of an aperture one perceives as passable scales with shoulder width (Warren & Whang, 1987). Furthermore, perceived and actual affordances generally show strong agreement in healthy, young individuals (Meeuwsen, 1991; Warren, 1984; Warren & Whang, 1987),

supporting the notion that accurate perception of affordances plays an important role in the successful planning and execution of movement.

An important question then is whether elderly individuals maintain an accurate awareness of their (declining) abilities, or whether a mismatch develops between perceived and actual abilities, leading to impaired movement planning and increased risk for injury. While few studies have directly addressed this issue, an area of related research concerns self-efficacy (or confidence) in one's ability to perform activities of daily living (Powell & Myers, 1995; Tinetti, 1994). Evidence suggests that self-efficacy strongly influences physical function and activity level, independent of sensory, motor, and cognitive status (Tinetti, 1994). However, self-efficacy provides no direct information on the accuracy of perceived abilities, and thus fails to establish whether an individual's confidence in their abilities is justified. In that regard, valuable companion parameters to self-efficacy would be direct comparisons between perceived and actual motor abilities.

1.4.2 Influence of movement planning on balance maintenance

A variety of evidence supports the contention that motor planning influences both functional mobility and risk for falls. For example, studies have shown that impairments in strength and flexibility increase one's risk for fall (Whipple, Wolfson, & Amerman, 1987), and correlate with impairments in gait, stair-climbing, and rising from a chair (Bassey et al., 1992; Schenkman, Hughes, Samsa, & Studenski, 1996; Schultz, 1992). However, biomechanical studies suggest that the strength and flexibility required for standing balance (Alexander, Shepard, Gu, & Schultz, 1992; Gu, Schultz, Shepard, &

Alexander, 1996) and activities such as rising from a chair (Schultz, Alexander, & Ashton-Miller, 1992) are generally well-within the range of values measured in elderly subjects. Studies also indicate that the influence of strength on gait speed and other physical performance measures is relatively modest (Brown, Sinacore, & Host, 1995; Wolfson et al., 1996). For example, in examining the effect of exercise therapy on gait speed, Buchner *et al.* (1996) found that strength accounted for 23 percent of the variation in gait speed at study onset. However, changes in gait speed occurring over the intervention period, while correlated with changes in health status and depression, were unrelated to changes in leg strength. Similarly, a recent cross-sectional study by Ferrucci and co-workers (1997) of 985 women enrolled in the Womens Health and Aging Study (WHAS) found relatively modest correlations (partial R^2 values of less than 0.20) between lower extremity strength and walking speed, sit-to-stand time, and balance.

Buchner and colleagues (1996) as well as Leidy (1994) have suggested that the relatively modest ability of pure motor capacities to predict task performance arises from the fact that, while motor capacities limit movement possibilities, motor planning and intent dictate the portion of such capacities which daily movements utilize.

The relative importance of these two factors, motor capacity and motor planning, in the etiology of falls is difficult to determine, in part because of the lack of techniques for evaluating a given individual's tendency to approach imbalance while performing daily activities (Maki, 1997; Rosengren, McAuley, & Mihalko, 1998; Tinetti, Mendes de Leon, Doucette, & Baker, 1994). This, in turn, limits our ability to design improved techniques for preventing falls and mobility disorders in the elderly.

To address this, I developed (Chapter 2) and improved upon (Chapter 3) a novel technique to determine how risk for imbalance during daily activities depends on behavioural (e.g., risk taking) versus neuromuscular factors. Specifically, the "Reach Utilization Test" was developed to determine whether differences exist between young and elderly women in tendency to approach imbalance during a forward reaching task. I considered that, on the one hand, general cautiousness or fear of falling might cause older adults to underestimate their abilities and utilize a smaller percentage of their maximum attainable reach. On the other hand, declines in cognitive ability or a reluctance to admit disability might cause older individuals to overestimate their abilities and utilize a greater percentage than young of maximum attainable reach (Robinovitch & Cronin, 1999). The first two chapters attempt to answer this question by comparing young subjects to two distinct elderly populations, nursing home residents (Chapter 2) and community-dwelling (Chapter 3).

1.5 Fall descent

1.5.1 Role of Fall Mechanics in Determining Hip Fracture Risk

An important question for hip fracture prevention strategies concerns the factors that separate injurious and non-injurious falls. Investigators have examined this through questionnaire regarding fall characteristics in individuals who fell and fractured, versus those who fell and did not fracture (Greenspan et al., 1998; Greenspan, Myers, Maitland, Resnick, & Hayes, 1994; Keegan, Kelsey, King, Quesenberry, & Sidney, 2004; Nevitt & Cummings, 1993b; Parkkari et al., 1999; Schwartz, Kelsey, Sidney, & Grisso, 1998; Wei, Hu, Wang, & Hwang, 2001). These include questions about the direction of the fall (e.g., forward, sideways, backwards, or straight down), body parts that received the impact of

the fall, and whether an attempt was made to break the force of the fall (e.g., with the outstretched hand). Several studies have also included comparison of bone mineral density (BMD) measures in fractured and non-fractured patients.

These studies indicate that hip fracture risk is increased over 30-fold by landing on or near the hip (Hayes, Myers, Morris, Gerhart, Yett, Lipsitz, 1993; Nevitt & Cummings, 1993b; Schwartz et al., 1998). In contrast, a single standard deviation decrease in femoral bone density increases fracture risk by 2 to 3-fold. Falling sideways causes a 5 to 6-fold increase in hip fracture risk (Greenspan, Myers, Maitland, Kido, Krasnow, and Hayes, 1994; Nevitt & Cummings, 1993b; Schwartz et al., 1998). When compared to injurious falls, non-injurious falls are more likely to involve impact between the ground and the hand or knee, which reduces hip fracture risk by approximately 3-fold (Nevitt & Cummings, 1993b; Schwartz et al., 1998). Risk for injury during a fall is also increased by lower limb weakness, which increases hip fracture risk over 5-fold (Schwartz et al., 1998; Tinetti, Doucette, Claus, & Marottoli, 1995), and upper limb weakness, which increases fracture risk 2-fold (Nevitt & Cummings, 1993b). Thus, in contrast to the traditional view of hip fractures as a consequence of osteoporosis, these data suggest that fracture risk is dominated by the direction of the fall, the configuration of the body at impact, the intactness of specific fall protective responses (such as braking the fall with the outstretched hand), and the neuromuscular status of the faller.

To design effective hip fracture prevention strategies, we must consider two sets of risk factors: those that lead to an increased propensity for falling, and those that increase the risk for hip fracture in the event of a fall. As reviewed by Tinetti (1994), risk factors

for falls include increased age, medication use, neurological disease such as Parkinson's syndrome, stroke, and impairments in muscle strength, joint movement, balance, gait, vision, hearing, and cognition. Many of these factors also increase one's risk for hip fracture, including impaired vision, use of specific medications, lower limb weakness, and neurological disease (Felson, 1989; Grisso et al., 1991; Ray, 1989).

1.5.2 Safe landing responses in young adults

However, important questions remain regarding why the elderly are more prone to falls that result in hip fracture. Investigators have shown the energies available in a fall from standing height greatly exceed those required to fracture either the young or elderly proximal femur ex-vivo (Courtney, Wachtel, Myers, and Hayes, 1995; Lotz, 1990). Then why is hip fracture from a standing-height fall a rare occurrence in the young (Robinson, Court-Brown, McQueen, & Christie, 1995) (even among athletes who undergo regular falls onto hard surfaces) and why do only 1-2% of falls result in hip fracture in the elderly? Also, why do fall-related wrist fractures far outnumber hip fractures in the young, while this trend is reversed in the very old (Praemer, 1992)? Such age-related changes in fracture incidence suggest that key differences exist between young and elderly individuals in their capacity not only to prevent falls, but also to "land safely" during a fall.

Few data exist on movement patterns during the descent and pre-impact stages of falls. Experimental measures of impact velocities during falls from standing were first reported by van den Kroonenberg and co-workers (1996), who measured planar motions of various body segments as young athletes executed self-launched, sideways falls onto a

gymnasium mattress. Reported hip contact velocities (which averaged 2.8 ± 0.4 (S.D.) m/s) were considerably lower than those predicted by rigid link pendulum models (Robinovitch, Hayes, & McMahon, 1991), suggesting that energy was absorbed through lower extremity muscle contraction during descent. Energy absorption can also be achieved through impact to the outstretched hands. Anecdotal support for this notion includes the high frequency of fall related wrist fractures in young adults (Owen, 1982), which suggests that hand impact is common, and results from several studies showing there is sufficient time during descent for individuals to move their hand(s) into a protective position and activate upper extremity muscles to break a fall (DeGoede, Ashton-Miller, Liao, & Alexander, 2001b; Dietz, and Noth, 1978; Kim & Ashton-Miller, 2003; Robinovitch, Normandin, Stotz, & Maurer, 2005). More direct support is provided by the results of Hsiao and Robinovitch (1998), who found that during unexpected sideways falls initiated by sudden translation of the support surface, young adults tended to avoid hip impact by rotating their trunk forward during descent to land on both outstretched hands. Unfortunately, this study involved a small sample size (n=6) and a small number of sideways falls (n=13). Furthermore, since each participant underwent multiple trials, the results may have reflected a learning effect. However, conflicting results were reported by van den Kroonenberg et al. (1996), who found that most of their young adult participants did not impact the ground with the outstretched hand during self-initiated sideways falls, despite being instructed to do so. In their study, hip impact tended to occur before contact of the arm or hand. Sabick et al. (1999) also reported this tendency of the arm to strike the ground after hip impact. A problem with each of these

previous studies concerns the self-initiated nature of the falls, which allowed for pre-planning of a landing strategy which is rarely the case during falls in daily life.

Given the conflicting nature of this evidence, it is still unknown whether unexpected sideways falls in young adults elicit a common sequence of protective responses that are known to protect against hip fracture. In Chapter 4, I address this issue by using a novel experimental paradigm which challenged participants to try and maintain balance after experiencing a large unexpected perturbation, which in the vast majority of cases elicited a sideways fall. I focused specifically on the following questions: (1) what is the frequency of impact to the lateral aspect of the hip during unexpected sideways falls in young adults? and (2) what is the frequency of impact to the upper extremity and knees during such falls?

1.5.3 Time demands of fall protective responses

Another important question regarding hip fracture prevention is whether individuals are able to modify the landing position of the body during the descent phase of a fall. Previous experiments showed that young individuals can avoid hip impact by rotating forwards of backwards during descent (Hsiao & Robinovitch, 1998; Robinovitch, Inkster, Maurer, & Warnick, 2003). However, the same limitations outlined on section 1.5.2 also apply to these studies making it difficult to compare real-life falls where one rarely has the ability to plan a descent strategy before imbalance occurs.

Furthermore, the effectiveness of a specific safe landing strategy may depend on time delays in selecting and initiating the response. In Chapter 5, I address this issue by testing whether individuals' ability to alter their body configuration while falling depends

on time during descent when the response is initiated. I hypothesized that a critical time window exists following the onset of the fall, beyond which one is unable to avoid hip impact. This study provides important evidence of the strict time demands that govern safe landing responses, to help guide the development of exercise interventions to prevent fall-related injuries in older adults.

1.5.4 Accuracy of self reported fall characteristics

Since most falls are unwitnessed (Hayes et al., 1993; Nurmi, Sihvonen, Kataja, & Luthje, 1996; Wagner, Capezuti, Taylor, Sattin, & Ouslander, 2005), the validity of questionnaire data of the type discussed in section 1.5.1 depends on the accuracy of elderly participants' recall of events that may have happen days or even weeks after the incident had occurred. For example, in a study by Nevitt and Cummings (1993b), the average interval between a hip or a wrist fracture and interview was 2.1 months (SD = 1.4). For falls without fracture, the average interval between fall and interview was 1.4 months (SD = 1.2). In a study by Schwartz and colleagues (1998), the median duration of time from a fall to interview was 1.4 months for individuals who suffered a hip fracture and 2.1 months for controls. Greenspan and colleagues (1994) obtained information about the characteristics of the fall within 5 days for participants with hip fractures and 10 days for controls. Moreover, Wei and colleagues (2001) interviewed patients with hip fracture within 3 days of hospital admission, and individuals who fell without hip fractures were interviewed within 2 weeks after confirmation of the fall.

An important question is whether individuals can accurately recount the cause and circumstances of their fall. In Chapter 6, I address this issue by interviewing the young

adults that participated on the "balance competition" (Chapter 4) immediately after they experienced their unexpected sideways falls. I focused specifically on the following questions: (1) how accurate are individuals in recalling the direction of fall, the attempts they made to recover balance, and the occurrence of impact to specific body parts? (2) Is the accuracy of recall of impact different depending on the body part? and (3) Is there a relationship between self-reported confidence and actual recall accuracy?

1.5.5 Mathematical models of fall descent

Several investigators have used mathematical models to predict fall descent and impact dynamics for forward falls (Zhou, Draganich, & Amirouche, 2002), backwards fall (Sandler & Robinovitch, 2001) and sideways fall (Lo & Ashton-Miller, 2008; van den Kroonenberg, Hayes, & McMahon, 1995). Sandler and Robinovitch's (2001) modeling study indicates that lower extremity joint torques during the descent phase of a fall can provide 79 percent attenuation in the body's vertical kinetic energy at impact, and 48 percent attenuation in the downward velocity of the pelvis at impact. A more complex model developed by Lo and Ashton-Miller (2008), indicated that peak hip impact forces can be reduced up to 56% by an arrest strategy combining flexion of the lower extremities, and progressive ground contact with the side of the body.

In Chapter 7, I present the results of modelling efforts to address previously unexplored questions: What is the effect on impact severity of slipping versus tripping and how does the nature of the balance perturbation influences the protective effect of lower extremity torque generation during descent?

1.6 Summary of goals

The present thesis describes a set of laboratory-based experiments to enhance our understanding of the biomechanics of balance maintenance and fall protective responses.

In Chapter 2 and 3, I describe a novel technique (the "Reach Utilization Test") to determine whether differences exist between young and elderly populations in tendency to approach imbalance during a forward reaching task.

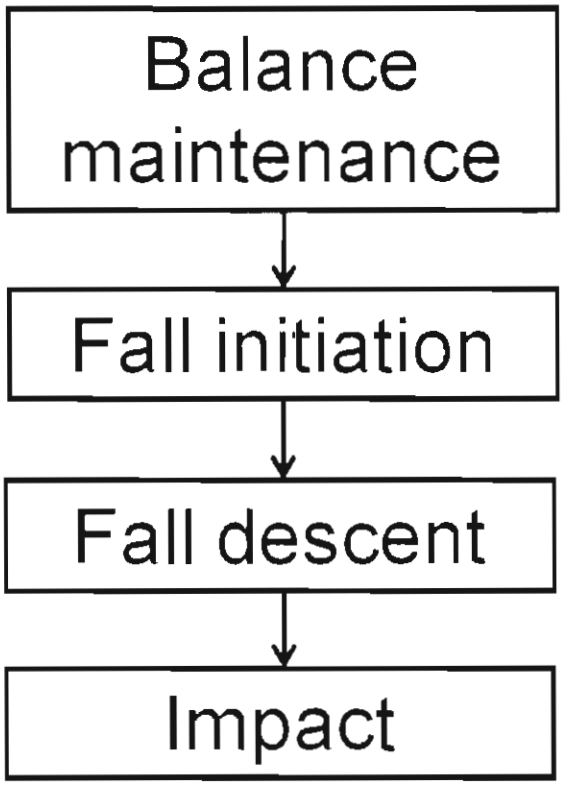
In Chapters 4 to 7, I focused on the measurement and analysis of protective responses during sideways falls. In Chapter 4, I describe a novel experimental paradigm to determine whether unexpected sideways falls in young adults elicit a common sequence of protective responses that may protect against hip fracture.

In Chapter 5, I describe the results of an experiment to test whether the ability of young women to avoid hip impact (by rotating forward or backward during descent) is influenced by the time, relative to the onset of the fall, when this response is initiated.

In Chapter 6, I describe the results of a questionnaire study to determine whether young individuals are able to accurately recall the details of a fall immediately after they experienced an unexpected perturbation in our motion analysis laboratory.

Finally, in Chapter 7, I describe results from a modelling study to address how fall severity, and one's ability to influence fall severity through muscle activation during descent, depend on the cause of the fall (slipping versus tripping).

Figure 1-1: Schematic view of the sequence of events involved in a fall.



CHAPTER 2 ELDERLY NURSING HOME AND DAY-CARE PARTICIPANTS ARE LESS LIKELY THAN YOUNG ADULTS TO APPROACH IMBALANCE DURING VOLUNTARY FORWARD REACHING

2.1 Introduction

Falls are a major cause of injury among the elderly, including approximately 90 percent of hip fractures and wrist fractures in this population (Grisso et al., 1991). Most falls in the elderly occur while performing daily activities such as walking, bending, reaching, or turning (Berg, Alessio, Mills, & Tong, 1997a; Campbell, 1989; Hill, Schwarz, Flicker, & Carroll, 1999; Nevitt, Cummings, & Hudes, 1991; Overstall, 1977; Speechley & Tinetti, 1991). One's risk for imbalance during such activities depends on the size of the base of support between the feet and the ground (King, Judge, & Wolfson, 1994), and on one's ability to maintain (or re-establish) the body's centre of gravity within the borders of the base of support. However, the relative importance of these two factors in the etiology of falls is difficult to determine, in part because of the lack of techniques for evaluating a given individual's tendency to approach imbalance while performing daily activities (Maki, 1997; Rosengren et al., 1998; Tinetti et al., 1994). This, in turn, limits our ability to design improved techniques for preventing falls and mobility disorders in the elderly.

In the present study, we focused on the task of reaching, where for any individual there exists a specific reach distance, beyond which imbalance will occur. We were specifically interested in whether there are differences between young and elderly

subjects in the tendency to approach this threshold during voluntary reaching movements. We considered that, on the one hand, general cautiousness or fear of falling might cause older adults to underestimate their abilities and utilize a smaller percentage of their maximum attainable reach (Buchner et al., 1996; Kressig et al., 2001; Maki, 1997; Tinetti et al., 1994). On the other hand, declines in cognitive ability or a reluctance to admit disability might cause older individuals to overestimate their abilities and utilize a greater percentage than young of maximum attainable reach (Robinovitch & Cronin, 1999). To address this question, we developed a new technique for quantifying subjects' tendency to approach their maximum attainable reach, via a parameter we termed the "normalized voluntary reach." We then used this technique to test the following hypotheses (1) that average magnitudes of reach utilization would be different between young adults and elderly adults who resided in nursing homes or participated in elderly day care facilities; and (2) that there would be an association among elderly subjects between reaching ability and reach utilization.

2.2 Materials and Methods

2.2.1 Subjects

Twenty-five elderly subjects and twenty-six young subjects participated in the study. Elderly subjects consisted of 11 males and 14 females, ranging in age from 62 to 87 years (mean = 77 ± 6 yrs), body mass from 49 to 116 kg (mean = 67 ± 14 kg), and body height from 144 to 179 cm (mean = 160 ± 9 cm). They were either residents of nursing homes (n =5) or participants in elderly day-care programs (n=20). Young subjects were all community dwelling, and consisted of 13 males and 13 females, ranging

in age from 19 to 33 years (mean = 23 ± 4 yrs), body mass from 40 to 84 kg (mean = 62 ± 10 kg), and body height from 150 to 201 cm (mean = 169 ± 12 cm).

Individuals were screened by a telephone interview and a variety of ancillary measures to determine whether they meet our inclusion criteria. These were: (1) able to read and understand simple directions in English;(2) able to stand independently for a period of 3 minutes; (3) able to walk continuously and without assistance a distance of 3 meters, as measured through the Get-Up-and-Go test (Mathias, Nayak, & Isaacs, 1986); (4) able to score greater than 15/20 on the Snellen test of visual acuity, with habitual corrective lenses if necessary; (5) no debilitating arthritis; (6) no severe kyphosis; (7) no diagnosed peripheral neuropathy; (8) no major change within the past 3 months in medical status, or change within the past 6 weeks in the use of hypnotic or antipsychotic medications; and (9) able to score greater than 20 points (out of 30) on the Folstein Mini-Mental Status Exam (MMSE) (Cockrell, 1988). This MMSE threshold was selected as a compromise between our competing desires to (a) screen those individuals who (due to severe dementia) would likely be unable to understand instructions, and (b) obtain a range of MMSE scores large enough to examine the relationship between cognitive status and reaching behaviour. The observed range of MMSE scores in our elderly subjects was 20 to 30, with a mean value of 25.6 ± 3.5 (SD).

All subjects provided informed consent, and the experiment was approved by the local Institutional Review Board. During the consent process and experimental measures, subjects were informed that the goal of the study was "to measure movement speeds during reaching." At no time were they provided with additional information regarding the study hypotheses.

2.2.2 Experimental Protocol

The experiment measured each subject's willingness or tendency to approach (during voluntary reaching) his or her maximum attainable reach distance, beyond which imbalance would occur. To conduct the experiment, we positioned the subject standing with the hands at the sides and the toes 10 cm from the near edge of a 90 cm high table that they faced (Figure 2.1). The table was covered with a sheet of white paper of surface area 61 cm by 152 cm, and a set of rails were secured over the paper. A cart having low-friction castors rested on top of the rails, which restricted motion of the cart to be along the subject's anterior/posterior axis. Six differently coloured targets (red, blue, black, yellow, green, and pink) were located on the cart, each of which corresponded to a different combination of reaching distance and direction (as explained below). The targets were arranged in a row to be equidistant to the subject, and 5 cm apart from one another. Each target was a 2 cm diameter sphere secured to the top of a spring-loaded pen that, when the target was struck, made a mark on the paper overlying the table. This mark consisted of an indentation and ink streak the colour of the target (thus allowing for measure of reach distance).

We first conducted static reaching trials to measure the subject's "upright reach" and "maximum attainable reach." Upright reach was defined as the farthest target distant, measured from the heels, that the subject could reach while standing stationary, without bending at the waist, and without leaning forward at the ankle. This provided the origin from which all other reach distances were measured (Figure 2.1). Maximum attainable reach was the farthest target distance from the origin that the subject could reach without taking a step (as defined by movement of the toes) or losing balance.

We then conducted dynamic trials to determine the subject's willingness to approach the maximum attainable reach during voluntary reaching movements. In these trials, we moved the cart by hand back and forth along a path defined at one end by the origin and at the other end by 1.25 times maximum attainable reach. The movement was roughly in the shape of a triangular waveform of frequency 0.125 Hz, so that 8 seconds elapsed between peak-to-peak excursions. Once during each movement cycle, we provided a vocalized cue (by audibly reciting the word "RED," "BLUE," "BLACK," "YELLOW," "GREEN," or 'PINK," corresponding to the colour of the target to be struck) to the subject to reach forward with their dominant hand and strike one of the targets "as soon as they could hit it." We further instructed the subject that their toes must remain stationary during the reaches (lifting of the heels was allowed), and that they should return to a stationary upright position after reaching. Control of the target motion and timing of the verbal cue was approximate and controlled manually by the experimenter.

The distance from the subject to the target bank at the instant of the go cue was randomized, subject to the constraint that four trials occurred at each of six different combinations of target distance and direction of movement. These combinations, along with the slow movement speed of the target, were selected to ensure that, regardless of the subject's reaching speed, in approximately 50 percent of trials the target was unreachable at the instant (and for a substantial interval after) the go cue was presented. The target distances (identified by marks on the table visible to the investigator but not the subject) were 0.5, 0.75, 0.87, 1.12, and 1.25 times maximum attainable reach. For the trials involving target distances of 0.5, 0.75, and 0.87 times maximum attainable reach,

the target bank was always moving away from the subject at the instant of the go cue. For the trials involving a target distance of 1.25 times maximum attainable reach, the target was always moving toward the subject at the instant of the go cue. Finally, for a target distance of 1.12 times maximum attainable reach; trials were acquired with the target both moving toward and moving away from the subject at the instant of the go cue. This combination of close and far targets ensured that subjects remained attentive during the trials, and were provided with a sense of accomplishment through the mix of challenging and easily attainable targets (and thus did thus not "give up" due to the overwhelming difficulty of the task). Occasionally, the subject contacted the wrong target during the trial. When this occurred, the investigator stopped the experiment, and identified the corresponding mark on the paper to be later rejected during data analysis. The trial was then immediately repeated.

2.2.3 Data Analysis

At the end of each session, we carefully removed the white paper overlying the table, which had pen marks on it indicating the origin, maximum attainable reach, and reach distances for each of the 24 trials (four trials times six conditions, with each condition shown in a different colour). The latter were measured as the distance from the origin to the indentation in the paper created by the pen when it first struck the paper. For all subjects, we disregarded the four trials involving target distances of 0.5 times maximum attainable reach, as these involved reach distances well below those observed in other series. For each of the five remaining conditions, we determined each subject's average reach distance (in cm) and normalized reach distance (in percent), where the normalized reach distance was given by (actual reach distance)/(maximum attainable

reach distance)*100. As an indicator of overall behaviour during the trials, we also calculated the upper quartile (75th percentile) of actual reach distance over the 20 trials, which we will refer to as "voluntary reach," and the upper quartile of normalized reach distance over the 20 trials, which we will refer to as the "normalized voluntary reach."

2.2.4 Statistics

We used a two-factor analysis of variance to test whether there was an effect of age (young versus elderly) and gender (male versus female) on mean values of normalized voluntary reach and maximum attainable reach (followed by post-hoc t-tests, where appropriate). We also used correlation to determine whether normalized voluntary reach associated with maximum attainable reach, and with measures of subjects' mental status (the MMSE) and mobility (as quantified by scores on the timed "get-up and go" test). To account for multiple comparisons, we regarded p values from individuals tests to indicate significance if $p < 0.005$. All statistical tests were conducted with statistical analysis software (SPSS Inc., Chicago, IL).

2.3 Results

We found that, when compared to young subjects, elderly subjects had smaller attainable reach and were less likely to utilize their attainable reach. Average values of maximum attainable reach normalized to body height were 30 percent smaller in elderly than young subjects (22 ± 6 percent body height versus 32 ± 4 percent body height; DF = 49 ; $p < 0.001$). Furthermore, average values of normalized voluntary reach were 32 percent smaller in elderly than in young subjects (65 ± 23 percent versus 95 ± 5 percent;

DF = 49 ; $p < 0.001$; Table 2.1). Consequently average values of voluntary reach were 53 percent smaller in elderly than young subjects (14 ± 7 percent body height versus 30 ± 5 percent body height; DF = 49 ; $p < 0.001$).

In each series, mean values of normalized reach were smaller for elderly than for young subjects (DF = 49 ; $p<0.001$; Figure 2.2). The greatest difference between groups occurred in the 1.25 times maximum attainable reach condition (44 ± 23 versus 95 ± 6 percent), while the smallest difference occurred in the 0.87 times maximum attainable reach condition (71 ± 23 versus 97 ± 5 percent). The 0.87 and the 0.75 times maximum attainable reach condition represented conditions where the largest average values of normalized reach were observed in both young and elderly groups.

There was no effect of gender on maximum attainable reach ($p = 0.356$; Table 2.1). However, there were nearly significant associations between normalized voluntary reach and gender ($p = 0.082$) and age * gender ($p = 0.064$), suggesting a trend of greater reach utilization in elderly women than in elderly men.

Reaching ability did not associate with reach utilization (Figure 2.3). There was no correlation between normalized maximum attainable reach and normalized voluntary reach for both elderly ($r = 0.085$; DF = 24; $p = 0.682$) and for young subjects ($r = 0.178$; DF = 25; $p = 0.385$).

Among elderly subjects, Get-Up-and-Go time correlated negatively with normalized voluntary reach ($r = -0.457$; DF = 22; $p < 0.002$; Figure 2.4(a)), but did not correlate with maximum attainable reach ($r = 0.102$; DF = 22; $p = 0.643$; Figure 2.4(b)). Also, MMSE scores among elderly did not correlate with normalized voluntary reach (r =

-0.009; DF = 24; p = 0.985) or maximum attainable reach (r = -0.002; DF = 24; p = 0.993).

2.4 Discussion

We found that elderly subjects were less likely than young to approach their maximum attainable reach during the voluntary reaching task employed in our experiments. Normalized voluntary reach averaged 65 ± 23 percent in elderly and 95 ± 5 percent in young. This indicates that cautiousness reduces elderly subjects' voluntary reach distance by 35 percent on average, and young subjects' voluntary reach by only 5 percent. Our results also suggest that, of the mean difference in voluntary reach between young and elderly subjects of 16 percent body height, neuromuscular constraints account for 9.5 percent body height (or 59 percent of the total), and cautiousness accounts for 6.5 percent body height (or 41 percent).

We also found that the normalized voluntary reach did not correlate with maximum attainable reach, despite the considerable variability in each of these variables among elderly subjects. This suggests that physical capacity and capacity utilization are independent predictors of reaching behaviour in the elderly. Furthermore, we observed a trend (although not statistically significant) for greater cautiousness among elderly men than elderly women in approaching maximum attainable reach. Future studies are required to clarify how, among the elderly, gender associates with tendency to approach imbalance during daily activities. Finally, we found that Get-Up-and-Go times among elderly subjects correlated negatively with normalized voluntary reach but did not associate with maximum attainable reach. This likely relates to the fact that subjects' pace in the Get-Up-and-Go test is self-selected. Therefore, as with their normalized

voluntary reach, behavioural variables such as motivation and fear may strongly influence test performance.

The normalized voluntary reach provides a currently unavailable technique for quantifying the influence on movement patterns of true motor capacities versus behavioural variables. Furthermore, the "low-tech" nature of the measure should allow for easy integration in the clinical setting. However, there are several important limitations to the technique. One of these concerns the possibility that psychological factors such as motivation or fear of falling may have influenced our measured values of maximum attainable reach, as well as voluntary reach (Schieppati, Hugon, Grasso, Nardone, & Galante, 1994). To help offset this possibility, we instructed and encouraged subjects to reach as far as possible during the maximum attainable reach trials. Another potential limitation is that the reaching targets were set at the same height for all subjects, and this may have caused subjects' height to influence their reaching performance. While our focus on normalized voluntary reach (i.e. voluntary reach divided by maximum attainable reach) may have helped to eliminate this confounding factor, it is possible that the "maximum attainable reach" and the "voluntary reach" were affected differently and in a nonlinear way by the table height. In addition, values of normalized voluntary reach may have been influenced not only by cautiousness in approaching a state of imbalance, but also by the accuracy of self-perceived limits of reachability, neural capacity for motor planning, musculoskeletal factors affecting the ability to move the hand rapidly and accurately, neuromuscular factors such as reaction time and somatosensory feedback regarding limb position and contact with the target, and neural control of predictive and reactive postural adjustments. However, because of the slowness of the target speed in

our trials, we believe these factors are far less important than risk taking behaviour in explaining the large differences we observed between young and elderly subjects in normalized voluntary reach. It is also possible that our results were slightly influenced by variations between trials and between sessions in the experimental conditions. For example, the cart was moved by hand, so it is likely that there were slight variations in its speed between trials and between all subjects. Furthermore, the investigator may not have always issued the aural "go" cue at precisely the same time and in the exact same way with respect to when the target passed the mark. Furthermore, variations in the investigator's tone of voice of sense of urgency could have affected performance. Lastly, the investigator was not blinded to the study's objectives and hypotheses, and therefore may have exhibited some unconscious bias in delivering the verbal cue. These issues could be addressed in future studies by, for example, lighting up the target instead of using an aural cue, and using a motor to move the cart. At the same time, we believe that because a single, highly trained investigator acquired all the measures, the between-subject variability in the nature of the cart movement and cueing in the current study was negligible. Certainly, we have no reason to believe that between-subject differences in experimental conditions could have accounted for the large differences we observed between young and elderly in maximum attainable reach and normalized voluntary reach.

Among the additional limitations of this study is the fact that we measured reaching behaviour under a single condition, and different degrees of risk-taking may arise under alternative reaching scenarios. Also, our elderly subjects were nursing homes residents or participants in adult day-care programs, and therefore we cannot be certain about the applicability of our results to community-dwelling elderly (who may differ in both

maximum attainable reach and in tendency to approach maximum reach). Furthermore, while we found that the normalized voluntary reach did not associate with maximum attainable reach, or with MMSE score, further study is required to understand the associations between capacity utilization and traditional measures of behavioural and cognitive status. Moreover, we found that some subjects exhibited values of normalized voluntary reach that were greater than 100 percent (Figure 2.2(a)). This may reflect sub-maximal efforts during the trials used to measure maximum attainable reach. Conversely, it may reflect the ability in some subjects to reach farther under dynamic than static conditions, due (for example) to the influence of joint flexibility or postural sway. On a related note, the fact that our protocol prevented us from explicitly controlling reaching speed might be regarded as a limitation of our study, since reaching speed may slightly affect attainable reach distance (Pai & Patton, 1997). However, based on pilot trials with young and elderly subjects, we have found that maximum forward reach distance is nearly identical under static and dynamic conditions (as the reader can easily verify with some simple reaching exercises). Accordingly, we do not believe that interpretation of our data is complicated by between-subject differences in reaching speed.

We took several precautions to help ensure that we measured natural reaching behaviour during the trials. Of primary importance was keeping subjects blinded about the true study hypotheses, and instructing them that the goal of the study was "to measure movement speeds during reaching." This, along with the instruction to contact the target as soon as they could hit it, caused subjects to believe that our primary focus was to measure movement speed and not reach distance.

We also attempted to minimize the potential effects on reach utilization of both maximum attainable reach and reaching speed, by using a slow target speed and a target path scaled to maximum attainable reach. Our observation of no significant association between normalized voluntary reach and maximum attainable reach indicates that our study design allowed us to isolate behavioural and neuromuscular influences on reaching performance. This was best achieved in the 0.87 times maximum attainable reach condition, which involved the greatest normalized voluntary reach for both young and elderly subjects. In contrast, the greatest differences between young and elderly in reach utilization occurred in the 1.25 times maximum attainable reach condition. In this condition, the target moved within reach shortly after the go cue was presented, and therefore faster responses resulted in greater voluntary reach distances. Accordingly, movement speed seemed to influence voluntary reach distances more strongly in this condition than in the others.

Our results are in agreement with previous studies indicating that elderly individuals have smaller attainable reach distances than young (Cavanaugh et al., 1999; Duncan, Studenski, Chandler, and Prescott, 1992), and reduced functional base-of-support, or ability to maintain the centre of pressure far from the ankle during standing (Endo, Ashton-Miller, & Alexander, 2002; King et al., 1994). For example, King and co-workers (1994) found that functional base of support (which they calculated as the difference between mean centre of pressure location during sustained forward and backward leaning, divided by foot length) was 30 percent lower in elderly subjects than in younger subjects. This is similar to the 30 percent difference in normalized maximum attainable reach we observed between our young and elderly subjects. Our finding that

elderly subjects tend to approach only 65% of their maximum attainable reach distance is consistent with previous studies indicating that fear of falling and cautiousness affect movement speed and task performance, independent of variables such as strength and flexibility (Buchner et al., 1996; Kressig et al., 2001; Maki, 1997; Rosengren et al., 1998). Future studies are required to examine the relationship between fear-of-falling, history of falls, and quantitative measures of risk-taking such as the reach utilization test.

In conclusion, we found that, when prompted to reach as quickly as possible towards targets that moved in and out of reach, elderly nursing home and day care participants were less likely than community-dwelling young adults to approach their maximum attainable reach. We also found that maximum attainable reach did not associate with tendency to approach this limit during the trials. This leads us to hypothesize that each may be an independent predictor of mobility and cause for falls in the elderly.

Table 2-1: Mean parameter values* separated by age and gender

	Young			Elderly		
	Males n=13	Females n=13	Total n=26	Males n=11	Females n=14	Total n=25
Body Height (cm)	176.41 ± 10.44	162.01 ± 8.53	169.21 ± 11.88	164.70 ± 8.35	155.62 ± 6.63	159.61 ± 8.60
Maximum Attainable Reach (percent Body Height)	32.76 ± 3.34	30.23 ± 4.84	32.49 ± 4.27	22.08 ± 5.53	21.97 ± 6.06	22.02 ± 5.71
Normalized Voluntary Reach (percent Maximum Attainable Reach)	95.19 ± 2.72	94.65 ± 6.87	94.92 ± 5.12	55.55 ± 23.74	72.16 ± 21.29	64.85 ± 23.48
Voluntary Reach (percent Body Height)	31.20 ± 3.45	28.66 ± 5.21	29.93 ± 4.52	12.31 ± 6.30	16.02 ± 7.13	14.39 ± 6.90

*Cell entries show mean ± one standard deviation.

Figure 2-1: Experimental setup for measuring reach utilization. The subject stands in front of a table, upon which a cart moves back and forth, towards and away from the subject. The subject is instructed that, upon hearing an aural go cue, he or she should reach forward and strike a spherical target on the moving cart as soon as they could reach it. When the target is struck, a pen makes a mark on the underlying paper, allowing detection of reach distance. The target location at the time of the go cue is randomly varied (between 50, 75, 87, 112, and 125 percent of the subject's maximum attainable reach), so that in some trials it is initially beyond reach, and the subject had to wait for it to move back within reach before striking it. Through repeated trials, the test measures the maximum distance the subject is willing to reach. Upright reach is the longest distance the subject can reach while keeping the feet stationary and trunk upright. Maximum attainable reach is the longest distance the subject can reach while keeping the feet stationary and leaning forward as far as possible. The open arrows indicate the direction of cart movement at the time of the go cue for each target distance. Measures are acquired for each trial of actual reach distance (in cm), and normalized reach (defined as actual reach distance)/(maximum attainable reach)*100. The 75th percentile of these parameters over all trials defines "voluntary reach" and "normalized voluntary reach," respectively.

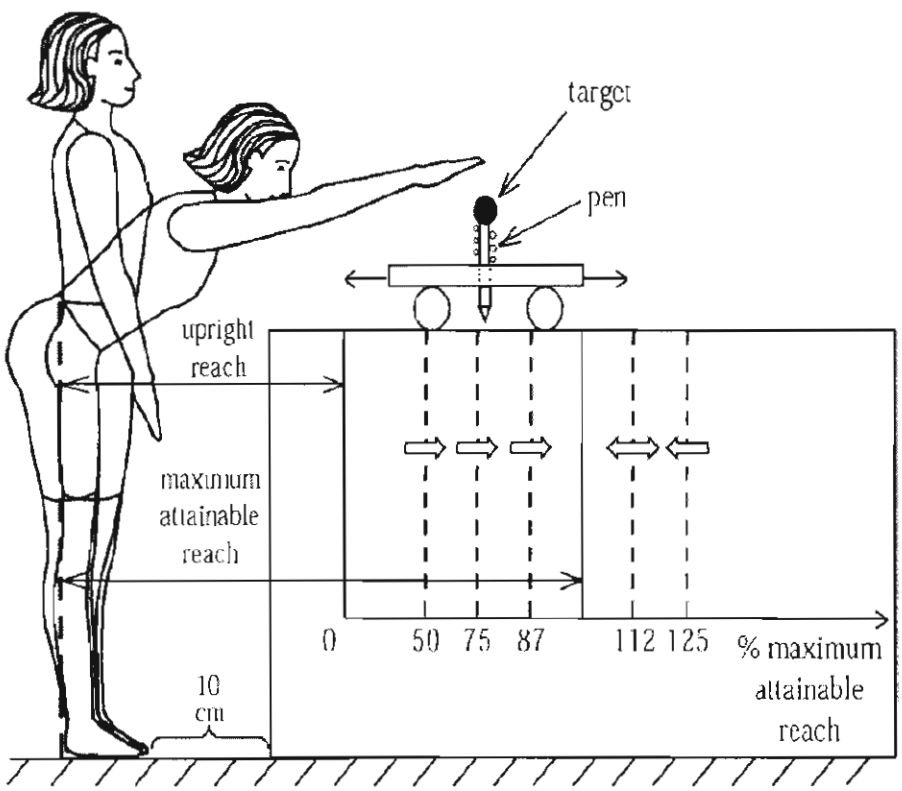
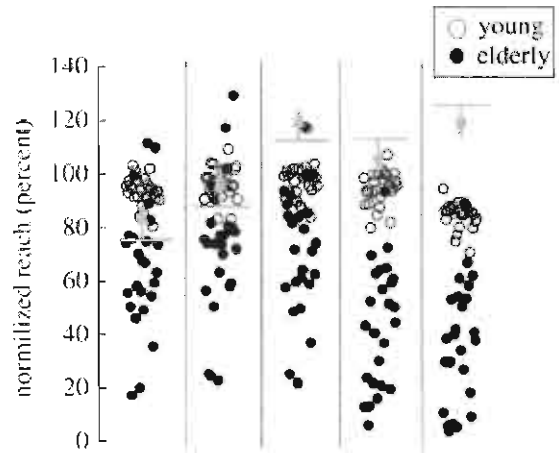
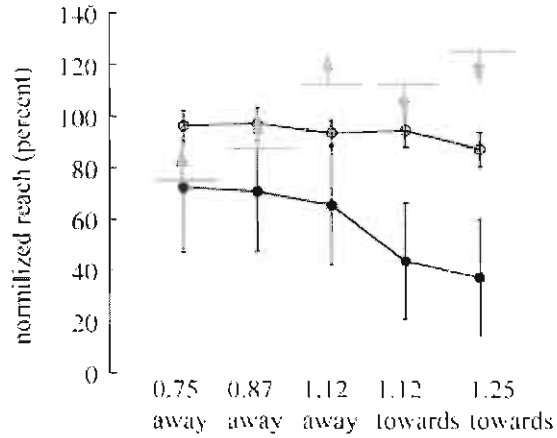


Figure 2-2: Average values of normalized reach in each experimental series. In (A), the graph shows average values of normalized reach for each subject and target condition. In (B), the graph shows average values for each group and target condition. Error bars show one standard deviation. Grey arrows, and labels at the bottom of (B), indicate the location and direction of the target at the instant of the go cue. For each series, normalized reach was smaller for elderly subjects than for young subjects, and elderly subjects exhibited greater between-subject variability in normalized reach. For both young and elderly subjects, the greatest normalized reach occurred when, at the time of the go cue, the target distance was 0.87 times maximum attainable reach.



(A)



(B)

Figure 2-3: Scatter plot of maximum attainable reach versus normalized voluntary reach. There was no association between these variables for both elderly subjects (r = 0.085; p = 0.682) and young subjects (r = 0.178; p= 0.385). This leads us to hypothesize that reaching ability and utilization of reaching ability may be independent predictors of mobility and risk for falls.

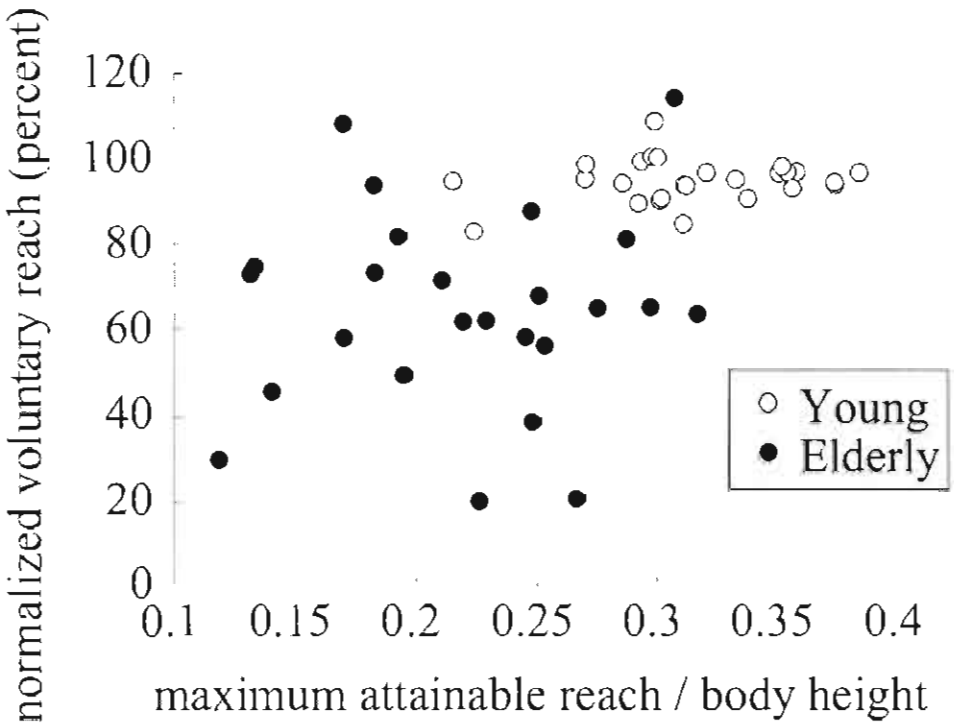
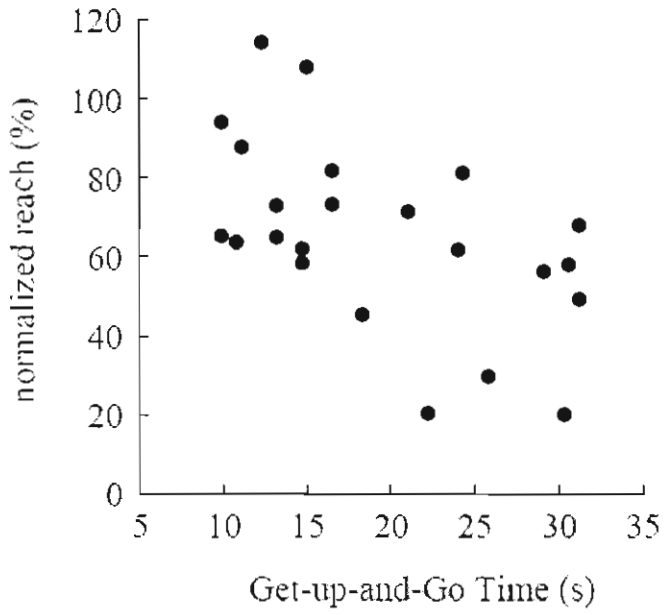
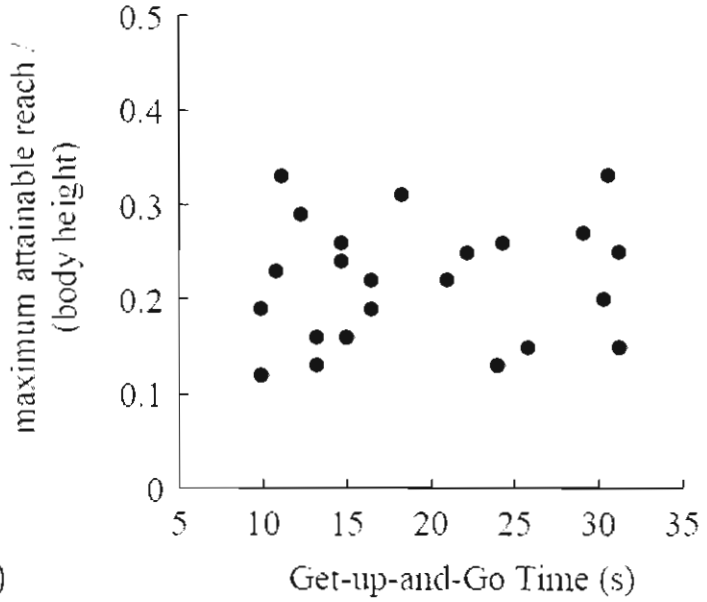


Figure 2-4: Relationship among elderly subjects between Get-up-and-Go time and (a) normalized reach and (b) maximum attainable reach. Get-up-and-Go time (a standard measure of mobility) is the length of time it takes the subject at their chosen speed to: rise from sitting, walk forward 3 m, turn around and walk back, and sit down. This variable correlated significantly with normalized voluntary reach ($r = -0.457$, $p < 0.002$), but not with maximum attainable reach ($r = 0.102$; $p = 0.643$). This indicates the substantial effect in the performance of each task of behavioural (risk taking) tendencies.



(A)



(B)

CHAPTER 3 NEUROMUSCULAR VERSUS BEHAVIORAL INFLUENCES ON REACHING PERFORMANCE IN YOUNG AND ELDERLY WOMEN

3.1 Introduction

Falls are a major health problem in the elderly with over 30% of individuals over age 65 falling at least once a year (Speechley & Tinetti, 1991). While growing evidence exists of the risk factors and circumstances surrounding falls, significant challenges remain in our understanding of their cause and prevention. In biomechanical terms, most falls can be defined as the loss of a stable upright posture due to movements (and lack of appropriate corrective actions) that displace the body's centre of gravity (COG) beyond the functional base of support (FBOS) between the feet and the ground. While slips and trips are common self-reported causes of falls in the elderly, similar in frequency are claims of ''loss-of-balance'', ''leg gave way'', ''changed posture'', or ''don't know the cause'' (Blake et al., 1988; Brocklehurst et al., 1978). Moreover, epidemiological evidence suggests that most falls have no obvious link to environmental hazards (Lyons et al., 2003), but instead result from failed attempts at performing daily activities, such as walking, turning, rising, and bending (Nevitt & Cummings, 1993a; Tinetti et al., 1995). Furthermore, one community-based study involving 1571 elderly found that fallers believed their own risk-taking behaviour (e.g., carrying too many objects, climbing on furniture) was a more common cause of falling than their health or environmental factors (Hornbrook, Wingfield, & Stevens, 1991).

Risk for imbalance and falls during such activities should depend on both the size of the FBOS, and on one's cautiousness in maintaining the COG within the boundaries of the FBOS during daily activities. However, while we know that declines occur with age in the size of the FBOS (King et al., 1994), little is known about whether there are also changes in individual's tendency to move the COG near the borders of the FBOS during daily tasks. Previously, we compared young adults and elderly adults who resided in nursing homes (or participated in elderly day cares) in their tendency to approach imbalance during a voluntary forward reaching task, using an experimental tool we termed the ''Reach Utilization Test'' (Feldman & Robinovitch, 2004). In this test, subjects were required to reach forward from a standing position as quickly as possible to targets that cycled in and out of reach. We found that elderly individuals were less likely than young to approach their maximum attainable reach, beyond which imbalance would occur (voluntary reach averaged 65 ± 23% of maximum attainable reach in elderly, and 95 ± 5% in young).We also found that maximum attainable reach did not associate with voluntary reach.

We interpreted this to reflect greater cautiousness among nursing-home elderly in approaching imbalance, due perhaps to fear of falling (Buchner et al., 1996; Hatch, Gill-Body, & Portney, 2003; Kressig et al., 2001). However, we were unable to test this latter hypothesis, since we did not acquire measures of balance confidence or fear of falling. We also wondered whether our results might have been affected by the fact that we measured maximum attainable reach under static conditions (similar to Functional Reach (FR) (Duncan, Weiner, Chandler and Studenski, 1990)), while the voluntary reach trials were highly dynamic, in the sense that subjects were instructed to grasp the target ''as

soon as they could reach it.'' We were particularly concerned that declines in attainable rates of muscle force generation (and corresponding centre of pressure (COP) velocities) may have reduced reach distance under dynamic conditions, especially for elderly subjects (Kozak, Ashton-Miller, & Alexander, 2003; Stapley, Pozzo, Cheron, & Grishin, 1999; Thomas, Corcos, & Hasan, 2003). Finally, we were concerned that during these initial trials we moved the target back and forth by hand, and reaching performance may have been affected by corresponding irregularities in the target velocity profile.

In the present study, we used an improved version of the "Reach Utilization Test" to determine whether, during voluntary forward reaching movements, differences exist between young and community-dwelling elderly women in tendency to approach imbalance. We hypothesized (based on our previous findings) that elderly subjects would have a significantly smaller voluntary reach distance than young, due to smaller maximum attainable reach (a neuromuscular constraint), and increased cautiousness in approaching maximum attainable reach (a behavioural constraint). We also hypothesized that subjects' cautiousness in approaching their maximum attainable reach would not correlate significantly with actual magnitudes of maximum attainable reach, but that it would correlate significantly with independent measures of balance confidence.

3.2 Materials and Methods

3.2.1 Subjects

Participants included 18 community-dwelling elderly women ranging in age from 70 to 87 years (mean age = 77 ± 5 years (S.D.)), and 18 community-dwelling young women ranging in age from 19 to 34 years (mean age = 22 ± 4 years (S.D.)). Elderly women were recruited through advertisements and posting of flyers in local newspapers

and at senior recreational centres. Young women were recruited through posting of flyers at local universities.

Subjects were initially contacted by telephone and excluded if they reported a major neurological disease (e.g., stroke or Parkinson's Disease) or debilitating orthopedic problems (e.g., severe arthritis), had experienced major injuries within the past year (e.g., bone fracture), had regular episodes of dizziness or fainting, were unable to stand independently and walk a distance of 4.5 m without assistance, or were unable to understand simple directions in English. Subjects were excluded by on-site evaluation if they were unable to score greater than 24 points (out of 30) on the Folstein Mini-Mental Status Exam (MMSE) (Cockrell, 1988), had evidence of major deficits in proprioception (as measured by big toe position sense and monofilament to the dorsum of the foot), or major uncorrected problems in visual acuity (as indicated by a score of less than 20/15 on the Snellen test) and depth perception (as indicated by a score of more than 10 cm on the Howard–Dolman stereopsis test; (Howard, 1919)). We include only females, since elderly women are more likely than men to experience falls and hip fractures, and characterizing the potentially complex effect of gender on performance is beyond the scope of this study. All subjects provided written informed consent, and the experiment was approved by the Research Ethics Committee of Simon Fraser University.

3.2.2 Experimental Protocol

Each subject visited the laboratory on two occasions, typically one week apart. On the first visit, we acquired ancillary measures of sensory, functional, cognitive, and psychosocial status (Table 3.1). We characterized postural steadiness as the average velocity of the COP in the anterior/ posterior direction during quiet stance for 15 s with

eyes open. We measured each subject's Functional Reach, defined as the distance from the subject's heels to the tip of her longest finger, when reaching forwards as far as possible while maintaining the fingertip at the height of the shoulders when standing, and other reaching measures we acquired), the subject was allowed to raise her heels, but was not allowed to move her toes. This is similar to Functional Reach, a clinical measure of FBOS (Duncan, Weiner, Chandler and Studenski, 1990). We also measured subject's performance on the timed Get-Up-and-Go (Mathias et al., 1986) and Sit-to-Stand tests (Bohannon, 1995). We characterized cognitive status with the Folstein Mini- Mental Status Exam (Cockrell, 1988) and balance confidence with the Activity Balance Confidence (ABC) scale (Powell & Myers, 1995).

On the second visit, we conducted the Reach Utilization Test (Feldman & Robinovitch, 2004). This two-part test is described in detail in the following paragraphs. To summarize, it is designed to measure how far subjects are willing to reach (as a percent of maximum attainable reach), when prompted to grasp a moving target (Figure 3.1). Part one of the test measures the subject's willingness or tendency to approach maximum reach, by implicitly encouraging (as opposed to explicitly instructing) subjects to reach as far as they are willing to go. To achieve this, a reaching target moves slowly back and forth, in and out of the subject's reach, and the subject is instructed that, after hearing a go cue, they should reach to grasp the target "as soon as they can reach it." By varying the location of the target at the time of the go cue, the test provides a range of voluntary reach distances, the outer fringe of which represents the farthest distance the subject was willing to reach under these conditions. In part two of the test, the subjects

maximum attainable reach distance is measured, for comparison with voluntary reach distances acquired in part one.

During this test, the subject stood with her feet shoulder width apart and heels aligned in the frontal plane (Figure 3.1), and was instructed to reach forward to grasp and pull down on the bottom edge of a 25 cm wide by 20 cm high cardboard reaching target (as described in further detail below). The midpoint of the cardboard edge was aligned at the height and medio-lateral position of the subject's dominant-side acromion when standing. The target was attached to a compliant spring and thereby exerted a negligible force on the hand when pulled down.

In each trial, we measured the three dimensional positions of 16 skin-surface reflective markers with a seven camera, 60 Hz motion analysis system (QTrac, Qualysis Inc.,Glastonbury, CT). Markers were located at the following sites: top of the subject's head, sacrum, and left and right shoulders (acromion), elbows (lateral epicondyle), wrists (wrist joint), anterior superior iliac spines, knees (lateral condyle), ankles (lateral malleolus), and toes (5th metatarsal). Markers were also located on the floor (in line with the subject's heels), and on the reaching target. We also acquired synchronized measures of the magnitude and location of foot contact forces from a 6 degree of freedom forceplate embedded in the floor (model 6090-15, Bertec,Worthington, OH), and determined the time and position of the target at the instant it was pulled down from a contact switch. Data from the forceplate and contact were acquired at 960 Hz.

In part one of the Reach Utilization Test, we measured the subject's ''voluntary reach distance.'' In these trials, a stepper motor was used to move the target towards and away from the subject, through a saw tooth displacement profile having an amplitude of

18 cm, a mean value of FR, a speed (between peaks) of 4 cm/s, and a period of 18 s (see inset to Figure 3.1). The subject was instructed that, upon hearing the aural go cue (a beep of 200 ms duration) she should ''reach and pull down the target as soon as she could reach it, using a single continuous motion.'' She was also instructed that, in the event the target was too far away to reach, she should ''wait for it to come back, and pull down on it as soon as she could reach it.'' For each subject, we acquired trials at seven different combinations of target distance and direction at the time of go cue (Figure 3.1). These combinations were selected so that, in at least one-half of all trials, the target was unreachable at the instant and for a substantial interval after the go cue was presented. They were: (FR - 16 cm), moving away from the subject; (FR - 8 cm), moving away from the subject; FR, moving away from the subject; FR, moving towards the subject; (FR + 8 cm), moving away from the subject; (FR + 8 cm), moving towards the subject; (FR + 16 cm), moving towards the subject. These combinations were presented in a pseudo-random manner, constrained by the requirements that (a) three trials were conducted for each combination (for a total of 21 trials), and (b) in the first trial, the target was always located beyond reach at the time of the go cue (at (FR + 8 cm), moving away from the subject). This ensured that the first trial measured naive risk-taking behaviour, before the subject had the opportunity to develop familiarity with the nature and difficulty of the task.

The go cue was triggered through a manual button press by the investigator (FF) based on visual inspection of the alignment between arrows located on the target and overhead track (Figure 3.1). Given the slow speed of the target, this technique allows for

high accuracy, with the go target always within ±1 cm of the desired location at the time of the go cue (Figure 3.2).

Subjects were allowed to lift their heels but were not allowed to move their toes (i.e., take a step). Trials where the subject missed the target or lost balance were discarded and repeated at the end of the testing session. As a safety precaution, all subjects wore a fall restraint harness that attached to an overhead support via a tether, which was slack during reaching.

In part two of the Reach Utilization Test, we measured the subject's maximum attainable reach distance under dynamic and static conditions. In these trials, the target was stationary during reaching. In dynamic trials, we instructed the subject to reach forward and pull down the target in a single continuous motion as quickly as possible, and to then return to upright standing. In static trials, we instructed the subject to reach slowly forward and hold the target for two seconds, before returning to an upright stance. In both conditions we conducted multiple trials, starting at an easy distance and moving the target 1 cm farther after each successful trial, until imbalance was detected by the need to take a step. The distance reached by the subject in the trial prior to imbalance was taken as maximum attainable reach. We shall refer to these maximum reach distances as ''MAX_DYNAMIC'' and ''MAX_STATIC''. To minimize the effect of our observation technique on subject's reaching behaviour during the test, we always measured voluntary reach distances before maximum attainable reach. Furthermore, the information sheet given to the subject simply stated that the study's aim was ''to measure movement speeds during reaching,'' and the investigator did not elaborate on the hypothesis to be tested.

3.2.3 Data Analysis

We defined reach distance as the horizontal distance (in cm) from the target location at the instant of the grasp, to the location that the longest finger reached when the subject stood upright in a comfortable stance, and extended the hand forward at shoulder height, while keeping the elbow and wrist fully extended. We shall refer to reach distance in the first trial as "VOL_FIRST" and the 75th percentile over all 21 trials (representing overall behaviour) as "VOL_QUART."

We characterized subject's initial tendency to approach imbalance by the ratio (VOL_FIRST / MAX_DYNAMIC) X 100, which indicates how near VOL_FIRST was to MAX_DYNAMIC. We shall refer to this ratio as UTILIZED_FIRST. We characterized subject's overall tendency to approach imbalance by the ratio (VOL_QUART / MAX_DYNAMIC) X 100, which we shall refer to as UTILIZED_QUART.

We also examined for each trial the maximum anterior displacement of the centre of pressure between the foot and the ground (calculated from forceplate data) and the maximum anterior displacement of the whole-body centre of gravity, estimated from marker position data and anthropometric relations provided by Dempster (Winter, Patla, and Frank, 1990). We also estimated hip flexor/extensor torques and ankle dorsoflexor / plantarflexor torques using inverse dynamics. Figure 3.3 shows temporal variations in these parameters for typical young and elderly women, under both VOL_FIRST and MAX_DYNAMIC conditions.

3.2.4 Statistics

To test our primary hypothesis, we used a two-sided independent-samples t-test to determine whether significant differences existed between elderly and young women in UTILIZED_FIRST and UTILIZED_QUART. We also examined (with independent-samples t-test) whether significant differences existed between elderly and young women in reach distances, COP and COG displacements, peak hip extensor torques, and peak ankle plantarflexor torques in each of the VOL_FIRST, MAX_DYNAMIC, and MAX_STATIC conditions. To test our second hypothesis, we used Pearson's correlation coefficient to determine whether UTILIZED_FIRST and UTILIZED_QUART correlated significantly with MAX_DYNAMIC, and with ancillary measures of subjects' balance confidence, cognitive status, and mobility. We regarded $p < 0.05$ to indicate significant effects. We used parametric tests for hypothesis testing after confirming through one sample Kolmogorov–Smirnov tests that all dependant variables were normally distributed, including the ratios UTILIZED_FIRST ($p = 0.829$ for young and $p = 0.746$ for elderly), UTILIZED_QUART ($p = 0.970$ for young and $p = 0.939$ for elderly), and MAX_DYNAMIC/MAX_STATIC ($p = 0.967$ for young and $p = 0.560$ for elderly). All statistical tests were conducted with statistical analysis software (SPSS Inc., Chicago, IL).

3.3 Results

Elderly women had a smaller maximum attainable reach than young women, and were less willing in voluntary trials to approach their maximum attainable reach (Table 3.2; Figure 3.4). Average values of MAX_DYNAMIC were 11% smaller in elderly than in young (young = 26.9 ± 3.4 cm; elderly = 23.9 ± 5.1 cm; mean difference = 3.0 cm; S.E.

of the difference = 1.44 cm; d.f. = 34; p = 0.022). Moreover, values for the UTILIZED_FIRST were 19.6% smaller in elderly women than young (young = 84.2 ± 11.3%; elderly = 64.6 ± 13.6%; mean difference = 19.6%; S.E. of the difference = 4.16%; d.f. = 34; p = <0.001).

Both groups became more confident over multiple trials (Figure 3.5), with UTILIZED_QUART averaging 78.7 ± 8% for elderly women and 89.3 ± 3.8% for young women (mean difference = 10.6%; S.E. of the difference = 2.1%; d.f. = 34; p = <0.001). However, there was correlation between UTILIZE_FIRST and UTILIZE_QUART for both elderly (r = 0.66; p = 0.003) and young women (r = 0.68; p = 0.002).

We also found that maximum attainable reach did not correlate with tendency to approach this limit during voluntary reaching trials (Figure 3.4). There was a trend, but not a significant correlation, between MAX_DYNAMIC and UTILIZED_FIRST for elderly women (r = 0.462; p = 0.053), and no significant correlation between these variables in young women (r = 0.304; p = 0.221). Furthermore, there was not a significant correlation between MAX_DYNAMIC and UTILIZED_QUART for both elderly (r = 0.332; p = 0.179) and young women (r = 0.047; p = 0.854).

We found no association among elderly women between UTILIZED_FIRST and measures of balance confidence, cognitive status, and functional status. There was no significant correlation between UTILIZED_FIRST and the test scores on the ABC scale (r = -0.035; p = 0.890), Functional Reach (r = 0.425; p = 0.089), Get-Up-and-Go (r = -0.026; p = 0.917), Sit-to-Stand (r = -0.197; p = 0.433), and sway during quiet stance (r = 0.221; p = 0.411). Functional Reach (when expressed as a percent body height) correlated with VOL_FIRST (r = 0.526; p = 0.030), VOL_QUART (r = 0.620; p = 0.008),

MAX_DYNAMIC ($r = 0.495$; $p = 0.043$), and MAX_STATIC ($r = 0.520$; $p = 0.032$), but not with scores in the ABC, Get-Up-and-Go, Sit-to-Stand, and postural sway tests.

For both young and elderly women, there was no difference between MAX_DYNAMIC and MAX_STATIC. Average values of MAX_DYNAMIC and MAX_STATIC were 26.9 ± 3.4% and 27.0 ± 2.9% respectively for young (mean difference = 0.1%; S.E. of the difference = 1.11%; d.f. = 17; $p = 0.711$) and 23.9 ± 5.1% and 23.9 ± 4.5% respectively for elderly (mean difference = 0.05%; S.E. of the difference = 1.30%; d.f. = 17; $p = 0.885$). This indicates that, for the range of speeds associated with our tests, reaching speed had little effect on maximum reach distance.

The difference in UTILIZED_FIRST between young and elderly women was due to differences in utilization of attainable centre of pressure excursion and ankle torque (Figure 3.6). In the first voluntary reaching trial, elderly women used 88.5 ± 7.9% of their maximum COP displacement and 82.1 ± 13.5% of their maximum ankle torque observed in MAX_DYNAMIC trials. In contrast, young women used 96.2 ± 7.5% of their maximum COP displacement and 92.6 ± 11.3% of their maximum ankle torque observed in MAX_DYNAMIC trials. Interestingly, elderly and young were similar in their utilization of available hip torque (which averaged 96.9 ± 33.5% in elderly and 92.9 ± 28.1% in young).

3.4 Discussion

We found that during the forward voluntary reaching task we examined, community-dwelling elderly women were less likely than young to approach their maximum attainable reach (and imbalance). Average values of UTILIZED_FIRST were 23% smaller in elderly than young (64.6 ± 13.6% versus 84.2 ± 11.3%).

We also found that physical capacity and capacity utilization contributed independently to maximum voluntary reach distances. For both young and elderly women, there was no association between UTILIZED_FIRST (or UTILIZED_QUART) and MAX_DYNAMIC. This agrees with others regarding the relatively modest ability of pure motor capacities to predict task performance under daily conditions. For example, in examining the effect of exercise therapy on gait speed, Buchner et al. (1996) found that strength accounted for 23% of the variation in gait speed at study onset. However, changes in gait speed occurring over the intervention period, while correlated with changes in health status and depression, were unrelated to changes in leg strength. In addition, a study examining factors that influence gait adjustment in older adults showed that sedentary older adults adopted a more cautious walking style than active ones, exhibiting shorter step lengths and slower step velocity (Rosengren et al., 1998). Similarly, in the cross- sectional Women's Health and Aging Study (WHAS), Ferrucci et al. (1997) found relatively modest correlations (partial R^2 values of less than 0.20) between lower extremity strength and walking speed, sit-to-stand time, and balance. These results support the notion that, while motor capacities limit our movement possibilities, motor planning and intent dictate the portion of such capacities utilized when performing daily movements.

As the trials progressed during the testing session, subjects became less cautious and more closely approached their maximum attainable reach. This trend was particularly striking among elderly women. However, we observed correlation between UTILIZED_FIRST and UTILIZED_QUART for both groups, indicating that (despite the

decrease in caution over the testing session) reach utilization on the first trial was a good indicator of overall behaviour.

For both young and elderly women, we observed no difference between MAX_DYNAMIC and MAX_STATIC. This indicates that the neuromuscular demands of reaching quickly (rapid initiation and halting of COG excursion) had little effect on maximum attainable reach. This agrees with Kozak and colleagues (2003) observation that elderly women reached no farther under ''comfortable'' than ''fast'' reaching conditions. Accordingly, it seems valid to express reach utilization either as a percent of MAX_DYNAMIC or MAX_STATIC (Feldman & Robinovitch, 2004).

Contrary to our expectations, we found that among our relatively healthy subjects, reach utilization did not associate with balance confidence (as measured by the ABC test). This suggests that tendency to approach imbalance during our reaching task may be governed by behavioural or cognitive variables (such as impulsiveness, laziness, competitiveness, or willingness to please the examiner) that are different than balance confidence. Alternatively, there may be a complex relationship between capacity utilization and balance confidence, indicating for example a reluctance to ''give in'' to one's fear. We also found that balance confidence did not associate with Functional Reach or Get-Up and-Go scores, which agrees with some (Hatch et al., 2003; Hotchkiss et al., 2004) but not all (Kressig et al., 2001) community-based studies.

Our study had important limitations. One concerns the potential effect of behavioural factors (such as motivation and learning) on our measures of maximum attainable reach. We tried to minimize such effects by encouraging subjects in these measures to ''reach a little further,'' until imbalance was observed. A second limitation is

the potential effect on voluntary reach distances of the safety harness that all subjects wore, which may have given them security to reach further than they might in real life. A further limitation is that we measured reach utilization in a single (albeit common) reaching scenario. Finally, our subjects were all women, and therefore we cannot be certain about the applicability of our results to elderly men (who may differ in both maximum attainable reach and in tendency to approach maximum reach).

On the other hand, we took several precautions to help ensure that we measured natural reaching behaviour during the trials. Of primary importance was keeping subjects blinded to the study hypotheses, and instructing them that our goal was ''to measure movement speeds during reaching.'' This, along with the instruction to grasp the target ''as soon as they could reach it'' caused subjects to believe that our primary focus was to measure movement speed and not reach distance. We also attempted to minimize the potential effects on reach utilization of both reaching speed and maximum attainable reach, by (a) using a slow target speed, (b) scaling the target path to measures of Functional Reach acquired in a previous visit, and (c) including a grasp component to provide a defined end point to the reach (although this makes it difficult to compare our voluntary reach distances to standard measures of Functional Reach). The lack of correlation between UTIL_FIRST and MAX_DYNAMIC indicates that our design was successful in isolating behavioural and neuromuscular influences on reaching performance.

In conclusion, we found that elderly women in the Reach Utilization Test were more cautious than young in approaching their maximum attainable reach, and imbalance. Tendency to approach imbalance did not associate with maximum attainable

reach distance, or with balance confidence. Accordingly, we are unclear why elderly women are less likely than young to approach imbalance. While such cautiousness may protect against falls, it may also lead to reductions in mobility and hamper the performance of daily activities. The Reach Utilization Test quantifies the influence on movement patterns of true motor capacities versus behavioural variables, and appears to reflect a domain of fall risk (tendency to approach imbalance) that is independent of physical capacity and balance confidence. Future studies are required to evaluate the clinical utility of this measure for predicting risk for falls and declines in mobility in elderly populations.

Table 3-1: Descriptive characteristics of subjects by age category

	Young (n = 18)	*Elderly (n = 18)*
Age (years)	22 ± 4 (19–34)	77 ± 5 (70–87)
Height (cm)	163.7 ± 7.2 (154.0–180.0)	159.1 ± 5.1 (151.0–169.0)
Weight (kg)	59.1 ± 12.4 (43.3–86.4)	66.0 ± 10.3 (54.5–96.2)
Functional Reach (cm)	35.5 ± 5.4 (25.5–47.5)	31.4 ± 6.3 (18.0–40.3)
Get-Up-and-Go test (s)	9.6 ± 1.5 (7.1–11.8)	12.0 ± 2.3 (9.5–15.9)
Sit-to-Stand test (s)	7.3 ± 1.6 (4.3–10.9)	11.8 ± 3.3 (7.3–20.6)
Balance sway test[a] (cm/s)	4.0 ± 1.4 (1.3–7.0)	6.1 ± 2.2 (3.2–11.5)
ABC scale (16–80)	77.9 ± 2.6 (73.0–80.0)	72.7 ± 8.4 (49.0–80.0)
Mini-Mental Status (0–30)	29.6 ± 0.6 (28.0–30.0)	28.2 ± 1.4 (26.0–30.0)

Cell entries show mean ± 1 S.D., with range shown in parentheses.

[a] Average velocity of the COP in the anterior/posterior direction.

Table 3-2: Mean parameter values[a] separated by age

	Young	Elderly	*p*-value
MAX_DYNAMIC (%BH)	26.9 ± 3.4	23.9 ± 5.1	0.022
MAX_STATIC (%BH)	27.0 ± 2.9	23.9 ± 4.5	0.022
VOL_FIRST (%BH)	22.8 ± 4.9	15.8 ± 5.5	<0.001
VOL QUART (%BH)	24.1 ± 3.2	19.0 ± 5.1	<0.001
UTILIZED_FIRST (%)	84.2 ± 11.3	64.6 ± 13.6	<0.001
UTILIZED_QUART (%)	89.3 ± 3.8	78.7 ± 8.0	<0.001

%BH, percent body height.

[a] Cell entries show mean ± 1 S.D.

Figure 3-1: Reach utilization experiment. A computer-controlled stepper motor was used to move the target towards and away from the subject with a saw tooth displacement profile, with a speed of 4 cm/s, amplitude of 18 cm, and a mean value equal to the subject's Functional Reach (FR). The subject was instructed that, upon hearing an aural go cue (a beep of 200 ms duration), she should reach and pull down the target "as soon as she could reach it using a single continuous motion." The path of the target is shown in the inset, with the black circles indicating a typical sequence of target locations at the time of the go cue.

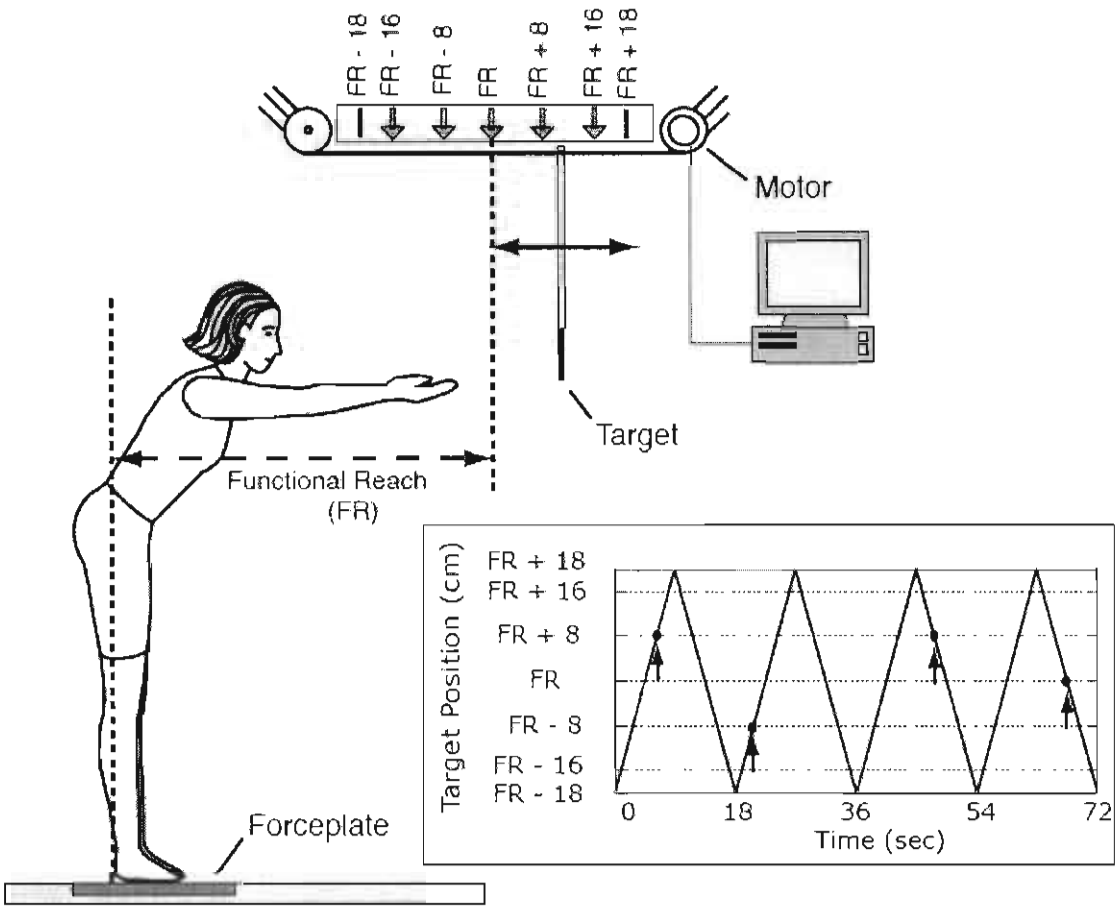


Figure 3-2: Accuracy and repeatability of target presentation. The circles indicate the positions of the target at the time of the go cue for a typical subject. For each of the seven combinations of target distance and target direction, the go cue was presented within ±1 cm of the desired location.

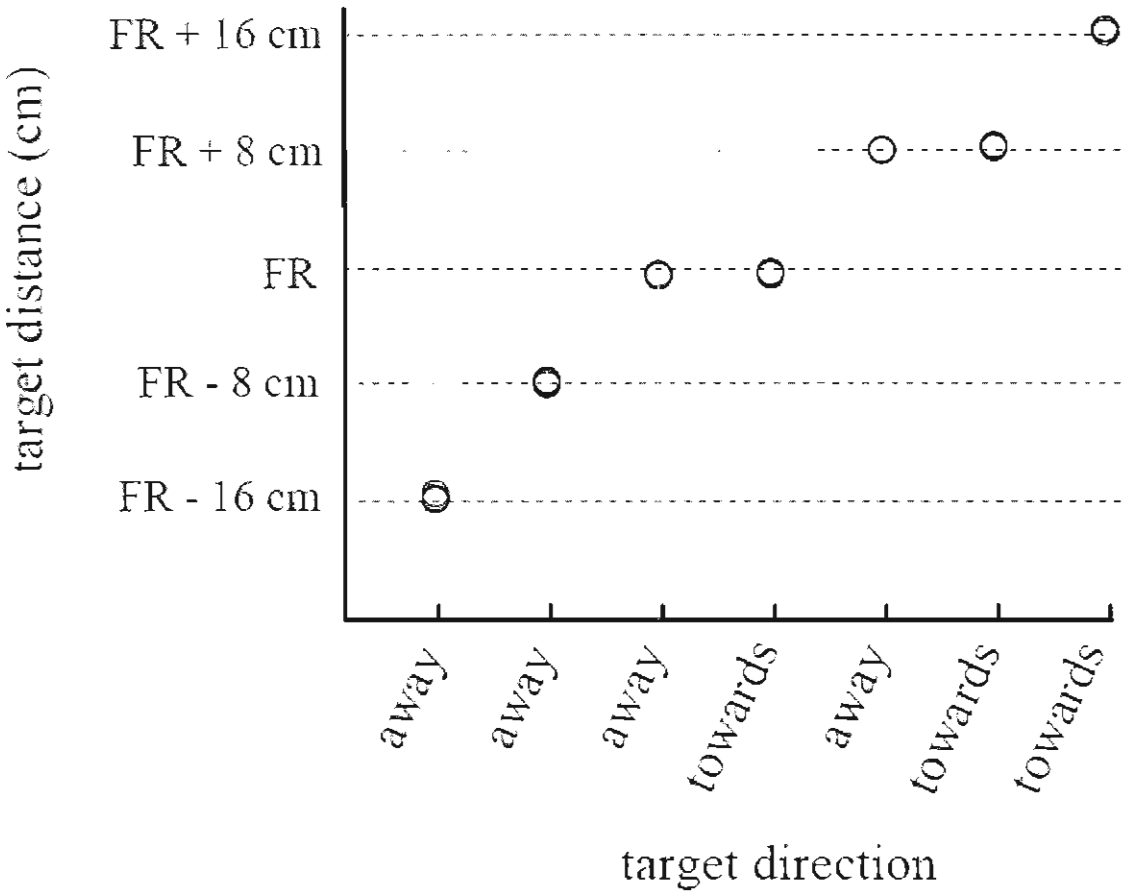


Figure 3-3: Typical temporal variations in hand movement, anterior/posterior displacement of centre of pressure (COP) and whole-body centre of gravity (COG), and sagittal plane joint torques for young and elderly women during reaching trials. The solid line corresponds to the MAX_DYNAMIC trial, and the dashed line shows the VOL_FIRST trial.

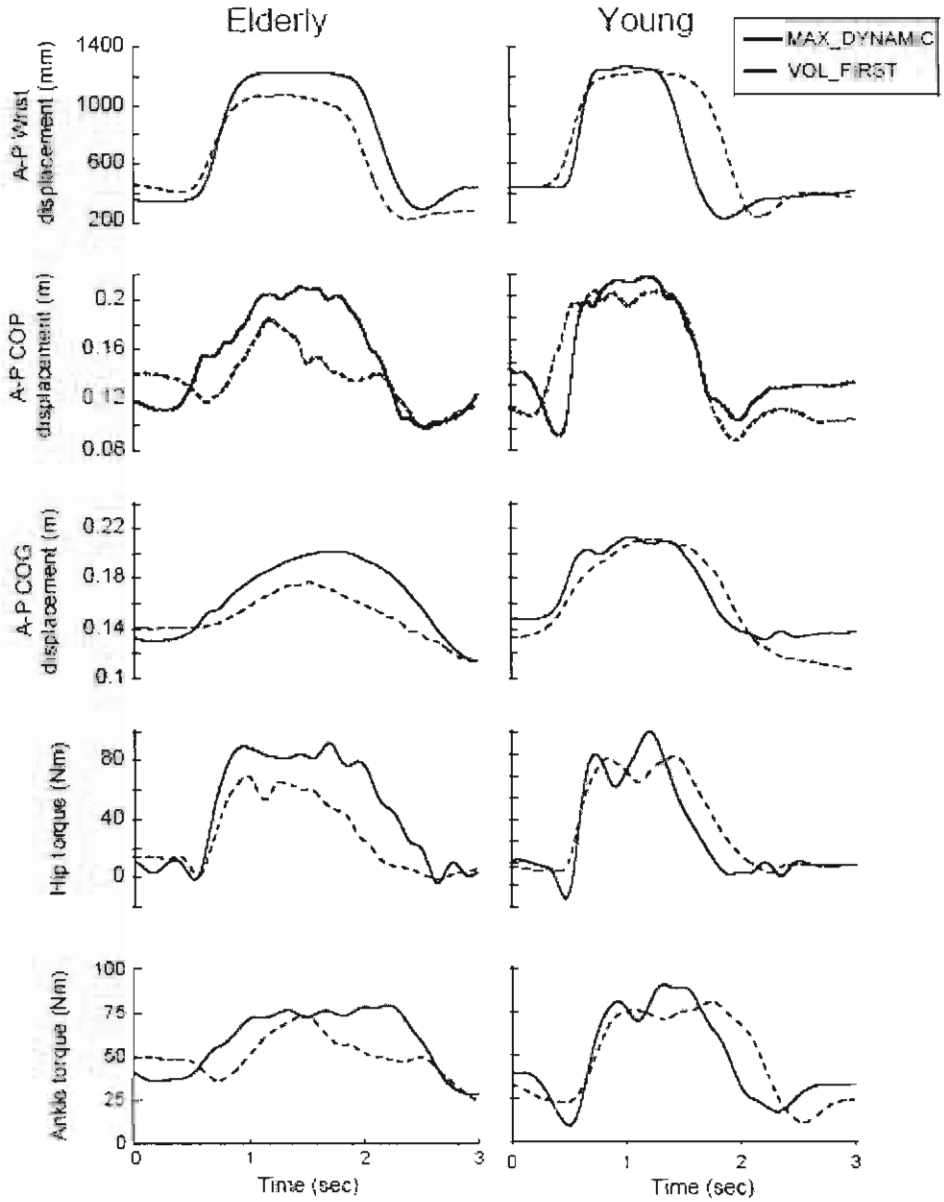


Figure 3-4: Relationship between voluntary reach in the first trial (UTILIZED_FIRST) and maximum attainable reach (MAX_DYNAMIC). Average values of MAX_DYNAMIC were lower in elderly women than in young (24 ± 5%BH vs. 27 ± 3%BH; shown by the vertical dashed lines). Elderly women were also less likely than young to approach their MAX_DYNAMIC during voluntary reaching trials, as reflected by smaller average values of UTILIZED_FIRST (65 ± 14% vs. 84 ± 11%; shown by the horizontal dashed lines). MAX_DYNAMIC did not correlate with UTILIZED_FIRST for elderly ($r = 0.462$; $p = 0.053$) or young women ($r = 0.304$; $p = 0.221$). This suggests that physical capacity and cautiousness *contributed independently* to reaching behaviour in our subjects.

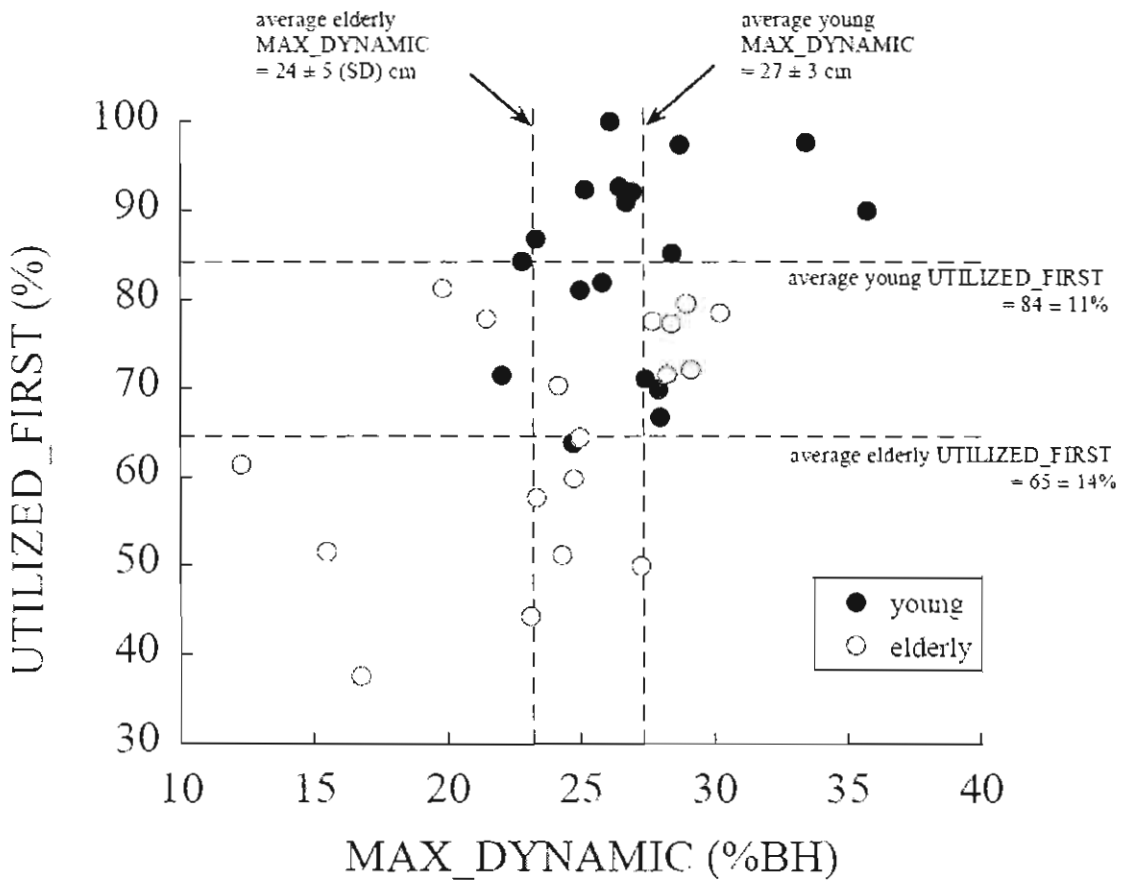


Figure 3-5: Relationship between voluntary reach in the first trial (UTILIZED_FIRST) and over all trials (UTILIZED_QUART). Both groups became more confident over multiple trials. UTILIZED_QUART averaged 79% for elderly women and 89% for young. There was significant correlation between these variables for both elderly (r = 0.66; p = 0.003) and young women (r = 0.68; p = 0.002), indicating that performance in the first voluntary reaching trial was a good indicator of overall behaviour.

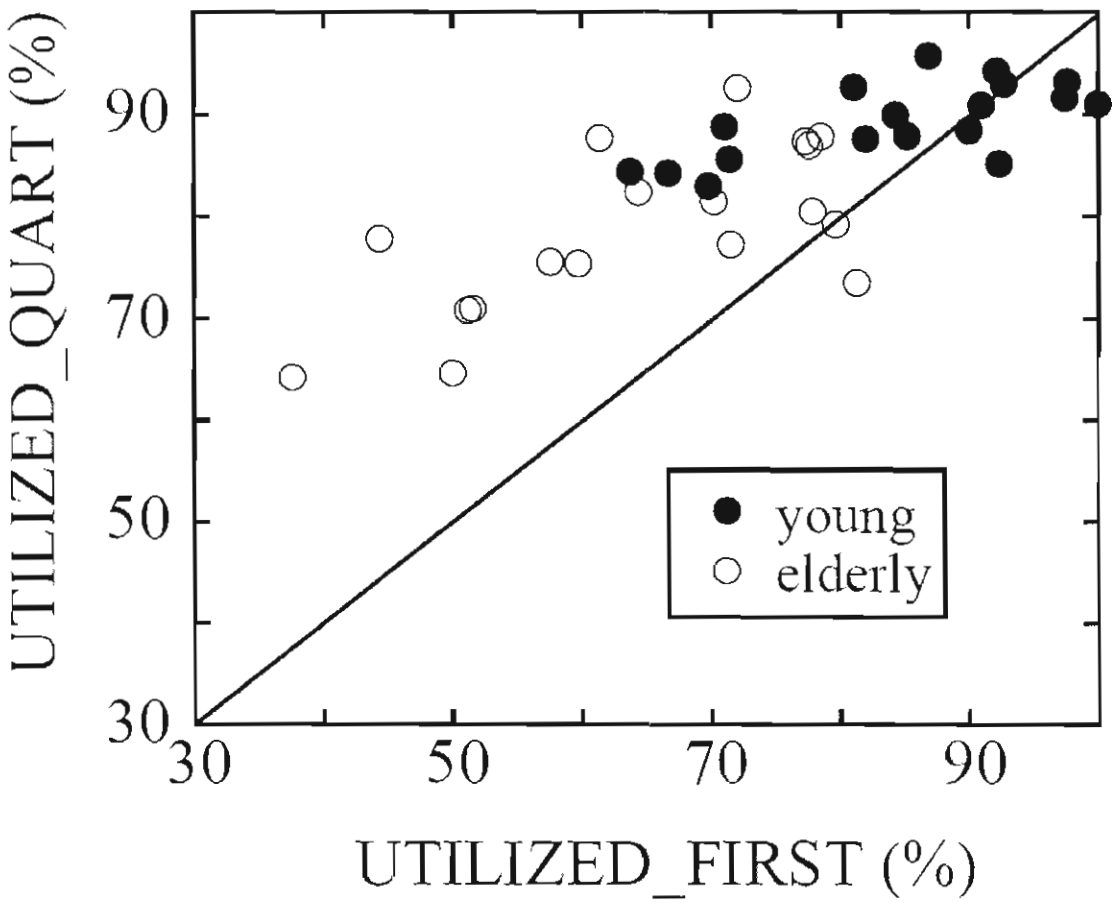
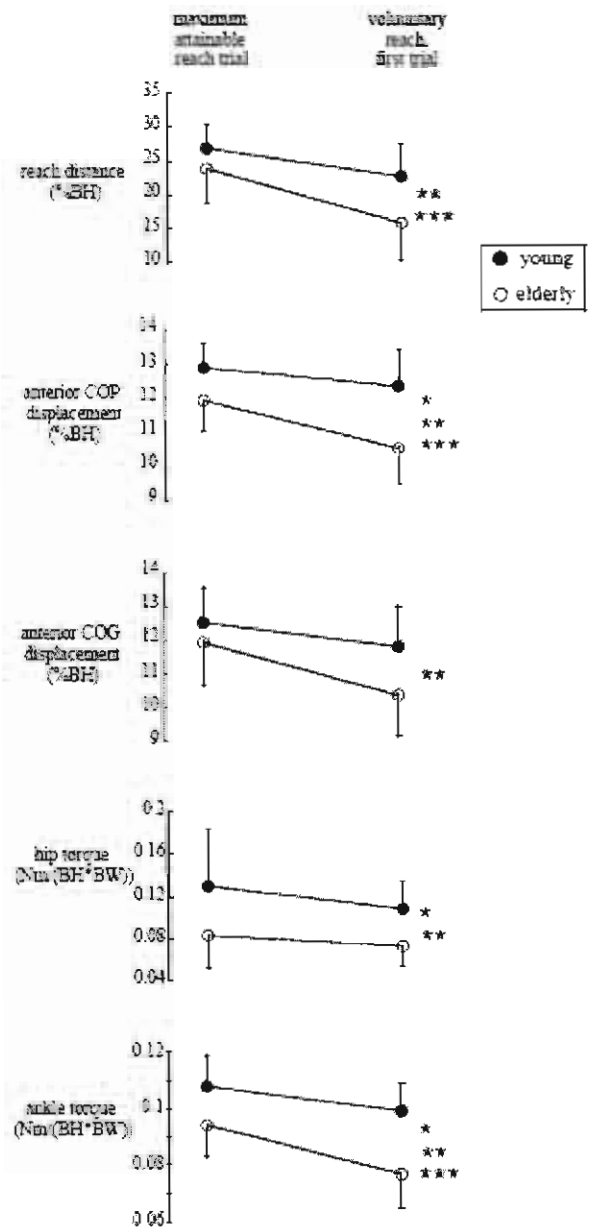


Figure 3-6: Comparison between young and elderly women in reach distances, anterior COP and COG displacements, peak hip torques, and peak ankle torques in MAX_DYNAMIC and VOL_FIRST conditions. Explanation of symbols: (*) indicates that the parameter was significantly different between young and elderly in the MAX_DYNAMIC condition ($p < 0.01$); (**) indicates that the parameter was significantly different between young and elderly in the VOL_FIRST condition ($p < 0.01$); and (***) indicates that the ratio of the parameter in the VOL_FIRST to MAX_DYNAMIC conditions was significantly different between young and elderly ($p < 0.01$).



CHAPTER 4 REDUCING IMPACT SEVERITY DURING SIDEWAYS FALLS: EVIDENCE IN YOUNG ADULTS OF THE PROTECTIVE EFFECTS OF IMPACT TO THE HANDS AND STEPPING

4.1 Introduction

There are over 25,000 annual hip fractures among the elderly in Canada, and over 90% of hip fractures are due to falls (Grisso et al., 1991). However, unlike fall-related wrist fractures (which are common throughout the lifespan), hip fracture is a relatively rare event in young adults, even among athletes who regularly experience sideways falls. This is surprising, since the energy available during a fall ranges from 100 to 300 J (Robinovitch, Brumer, & Maurer, 2004), while only about 25 J are required to fracture the proximal femur of young adults (Courtney, Wachtel, Myers, & Hayes, 1995).

One possible factor to explain the low occurrence of hip fractures in the young relates to the mechanics of the fall. Risk for hip fracture in the elderly is six times greater during sideways than forward or backward falls, and 30 times greater if the fall results in direct impact to the hip region (Nevitt & Cummings, 1993b). Conversely, impacting one or both hands decreases fracture risk 3-fold (Greenspan, Myers, Maitland, Kido, Krasnow, and Hayes, 1994; Schwartz et al., 1998). Therefore, young adults might be protected against hip fracture during a sideways fall by the tendency to avoid impact to the hip, or the tendency to impact the hands and/or knees.

Anecdotal support for this notion includes the high frequency of fall related wrist fractures in young adults (Owen, 1982), which suggests that hand impact is common, and results from several studies showing there is sufficient time during descent for individuals to move their hand(s) into a protective position and activate upper extremity muscles to break a fall (DeGoede et al., 2001b; Dietz and Noth, 1978; Kim & Ashton-Miller, 2003; Robinovitch et al., 2005).

More direct support is provided by the results of Hsiao and Robinovitch (1998), who found that during unexpected sideways falls initiated by sudden translation of the support surface, young adults tended to avoid hip impact by rotating their trunk forward during descent to land on both outstretched hands. Unfortunately, this study involved a small sample size (n=6) and a small number of sideways falls (n=13). Furthermore, since each participant underwent multiple trials, the results may have reflected a learning effect.

Conflicting results were reported by van den Kroonenberg et al. (1996), who found that most of their young adult participants did not impact the ground with the outstretched hand during self-initiated sideways falls, despite being instructed to do so. In their study, hip impact tended to occur before contact of the arm or hand. Sabick et al. (1999) also reported this tendency of the arm to strike the ground after hip impact. A problem with these studies was the self-initiated nature of the falls, which allowed for pre-planning of a landing strategy, unlike unexpected falls in daily life.

Given the conflicting nature of this evidence, our goal in the current study was to determine whether unexpected sideways falls in young adults elicit a common sequence

of protective responses that are known to protect against hip fracture. We focused specifically on the following questions: (1) what is the frequency of impact to the lateral aspect of the hip during unexpected sideways falls in young adults? and (2) what is the frequency of impact to the upper extremity and knees during such falls? We addressed these questions through a novel experimental paradigm, which challenged participants to try to maintain balance after experiencing a sudden unpredictable perturbation, and in the vast majority of cases elicited a sideways fall.

4.2 Materials and methods

4.2.1 Participants

The participants in our falling experiments consisted of 44 young individuals (31 women and 13 men) ranging in age from 19 to 26 years (mean age=21 years (SD=2)), body mass from 43 to 112 kg (mean mass=65 kg (SD=14)), and body height from 149 to 187 cm (mean height=168 cm (SD=9)). All participants were undergraduate students enrolled in a biomechanics course (73 students were enrolled in this course) who volunteered to participate in an experimental "balance competition." Participants were healthy and free of diagnosed neurological disease or debilitating orthopedic conditions, and 87% of them participated at least twice a week in some type of physical activity. All participants provided written informed consent, and the experimental protocol was approved by the Research Ethics Committee of Simon Fraser University.

4.2.2 Experimental protocol

During the balance test, the participant stood barefoot on top of a rubber sheet that, without warning, was made to translate horizontally to the participant's right side (Figure

4.1a) by means of a linear motor (T4D motor, Trilogy System Corporation, Webster, Texas, USA). As shown in Figure 4.1b, the total displacement was set to 1.15 m (actual mean value=1.14 m, SD=0.04), the terminal velocity was set to 2.2 m/s (actual peak value=2.22 m/s, SD=0.07), and the initial acceleration and terminal deceleration was set to 30 m/s^2 (actual mean value=29.2 m/s^2, SD=2.0).

We used several precautions to minimize participants' ability to pre-plan a postural response, in order to mimic a real-life fall. The only instruction we provided to the participant was: "your balance will be perturbed, and your goal is to maintain your balance." Moreover, the room was completely masked to prevent the participant from anticipating what was to come. The participant was guided to stand on a rigid platform mounted flush to the gymnasium mat (30 cm thick) landing surface, and the entire surface was masked with the rubber sheet. Drapes were used to conceal the linear motor and related hardware. No information was provided about the compliance of the ground surface, or the nature of the perturbation. In order to focus on naive responses, each participant performed only one trial, and no practice trials were allowed. Furthermore, all testing was completed in a single afternoon, and participants who were waiting to be tested were prevented from having contact with those who completed the testing.

Immediately following the experiment, we conducted a structured interview with each participant, which included the question "right before the trial, while you were standing on top of the platform, were you aware that the floor was going to move before it did?" Only 10% of participants responded "yes" to this question, and 90% responded "no." Thus, few participants interpreted the statement "your balance will be perturbed" to

mean that the floor would move from under them. While we did not inquire about alternative expectations, these may have included expectations that they would be nudged from behind, or would be asked to perform movements that challenged their balance to gradually increasing degrees (such as standing or hopping on one leg, or reaching as far as possible, etc.).

We used an eight-camera, 240 Hz motion measurement system (Motion Analysis Inc., CA, USA) to acquire three-dimensional positions of 18 reflective, skin-surface markers located at the head, sacrum (L5/S1 junction), and bilaterally at the acromion process (shoulder), lateral epicondyle of the humerus (elbow), distal end of the radius (hand), anterior–superior–iliac spine (ASIS), greater trochanter (hip), lateral epicondyle of the femur (knee), lateral malleolus (ankle), and third metatarsal (toe). As a safety measure, participants wore helmets and wrist guards (over which the head and hand markers were secured). Three markers were also placed on the translating rubber sheet overlying the impact surface.

4.2.3 Data analysis

For each trial, we inspected motion data to determine whether or not a fall occurred, whether a step was executed in an attempt to recover balance, the sequence of impacting body parts, and the orientation of the pelvis at the time of impact. A participant was considered to have fallen if any body part other than the feet contacted the impact surface. We classified a trial as involving a "complete step", if there was lifting and repositioning of the left (loaded) foot in a more lateral position on the ground, or the right (unloaded) foot in a more medial location, before impact to a hand, knee, or the pelvis.

We classified a trial as involving a "partial step", if there was lifting and swinging of the left leg in the lateral direction, or the right leg in the medial direction, but impact to a hand, knee, or the pelvis before that foot was repositioned on the ground. Finally, we classified a trial as "no attempt to step" if there was no apparent attempt to lift and swing the left foot laterally, or the right leg medially. Hand and hip impact times were determined by the frame in which the hand or pelvis marker (ASIS, sacral, or hip) crossed a virtual line located 5 cm above the mean vertical position of the three markers located on the impact surface. The hip proximity angle (α) reflects how close the individual came to directly impacting the lateral aspect of the pelvis (or greater trochanter of the proximal femur), with α=0 deg indicating direct impact to the lateral aspect of the pelvis, and ±90° indicating impact to the buttocks or anterior aspect of the pelvis (Figure 4.2a). Anatomical considerations suggest that values of α greater than 30° will produce direct loading to the ischium (for backward rotation) or ilium (for forward rotation). Therefore, an α value less than 30° was regarded as indicating hip impact (Figure 4.2b). Pelvis impact velocity was calculated as the vertical velocity of the left hip marker at the instant of pelvis impact.

4.2.4 Statistical analysis

Our primary research questions were addressed descriptively by examining the frequency and temporal sequence of impact to the hip, knees, and hands. We conducted secondary analysis using independent sample *t*-tests to determine whether specific protective responses (use of a step, or impacting the hand before the hip) affected pelvis impact velocity (an indication of fall severity). We also used Pearson correlations to examine associations between continuous variables. We regarded $p<0.05$ to indicate

significant effects. All statistical tests were conducted with statistical analysis software (SPSS Inc., version 12.0, Chicago, IL, USA).

4.3 Results

The large majority of trials caused sideways falls that produced direct impact to the lateral aspect of the hip (Table 4.1). Only 5% of participants were able to avoid falling altogether, and another 5% fell but did not impact the pelvis. The remaining 90% of participants fell and impacted the pelvis, and 98% of those cases involved direct impact to the hip region. The average hip proximity angle was 8° (SD=15) to the posterior side (range=63° posterior to 18° anterior; Figure 4.2c). Pelvis impact velocity averaged 3.01 m/s (SD=0.83), or 75% (SD=20) of that predicted from simple free-fall dynamics.

Impact to the hands and knees was common (Table 4.1). Impact occurred to the left hand in 98% of falls, and in 95% of these cases, before pelvis impact. Impact occurred to the right hand in 64% of falls (19% before pelvis impact), to the left elbow in 60% of falls (25% before pelvis impact), and to the left knee in 60% of falls (100% before pelvis impact). Impact to the head was avoided in all cases, primarily by laterally flexing the trunk and impacting the outstretched hand(s). The average inclination of the trunk with respect to the vertical at the instant of pelvis impact was 42° (SD=15).

The average time interval between the onset of the perturbation and impact to the pelvis was 626 ms (SD=40). The most common impact sequence in trials that involved hip impact were (a) initial impact to the left knee, followed by impact to the left hand, and finally left hip, and (b) initial impact to the left hand, then the left hip, and the right hand (Table 4.2). The average interval between hand impact and pelvis impact was 50 ms

(SD=40; range=−12–175 ms), and only two participants impacted the hip before the hand. The impact velocity of the pelvis decreased 3.6% for every 10 ms increase in the interval between hand and pelvis impact (r=−0.6; p<0.001; Figure 4.3).

Sixty-two percent of participants exhibited an obvious attempt to recover balance by stepping (Figure 4.4). Thirty-two percent of participants initiated but did not complete a step (Figure 4.4a). Thirty percent of participants executed one or more complete steps, and in all of these trials the right (perturbation-unloaded) foot was lifted before the left. In 11% of trials, the right foot was swung medial and crossed anterior or forward to the left foot (Figure 4.4b). In 14% of trials, the right foot was swung medial and crossed posterior or behind the left foot (Figure 4.4c). In 5% of trials, the participant executed a two-step skipping or jumping movement, and these were the only participants to avoid falling (Figure 4.4d). This skipping response involved the following chronology of events: (a) lifting of the right foot, (b) lifting of the left foot and a brief aerial phase where both feet were off the ground, (c) landing of the right foot in a location near to its initial position (medial to the left foot), and (d) landing of the left foot in a location lateral to its initial position. Participants who were able to complete a step before their fall (n=9) took longer to impact the pelvis following the perturbation (691 ms (SD=46) versus 613 ms (SD=52), p<0.001), had a longer interval between hand impact and pelvis impact (85 ms (SD=47) versus 39 ms (SD=31), p=0.001), and had a lower pelvis impact velocity (2.46 m/s (SD=0.94) versus 3.16 m/s (SD=0.74), p=0.02; Figure 4.5).

4.4 Discussion

We found that sideways falls in young adults consistently result in impact to the hip region. This is considerably different than Hsiao and Robinovitch's (1998) finding that young individuals tend to avoid hip impact during sideways falls by rotating the trunk to face the impact surface, and landing on the hands and knees. We observed this technique only once. This discrepancy probably relates to differences in study design. Participants in Hsiao and Robinovitch's study were exposed to multiple perturbations of gradually increasing severity, and were provided with a priori knowledge of the direction of the perturbation and the compliance of the ground surface. These factors probably assisted them in pre-planning their balance recovery and falling strategies. By eliminating the ability to practice and predict the nature of the perturbation, our current study design probably elicited falls that more closely resemble those occurring in real life.

We also found that the hand impacted before the hip in the vast majority of falls, and the knee impacted before the hip in most falls. Furthermore, the impact velocity of the pelvis decreased with increases in the time interval between hand and pelvis impact (which presumably influenced the energy absorbed by the upper extremity prior to pelvis impact).

Participants impacted their left hand in 98% of falls and in only two cases after hip impact. This differs from van den Kroonenberg et al.'s (1996) finding that impact to the hip precedes impact to the hand during sideways falls. van den Kroonenberg also reported a more upright trunk inclination at impact (21.7° (SD=13.3)) with respect to the vertical, as opposed to our current value of 42° (SD=15). These different outcomes again

probably relate to differences in study design. In van den Kroonenberg et al.'s study, unlike ours, the sideways falls were self-initiated (and thus pre-planned) from standing, and the participants had knowledge of the low compliance of the impact surface.

Our results suggest that even unsuccessful attempts to recover balance by stepping may reduce risk for hip fracture during a fall. In particular, we found that taking a step (a) increases the time to pelvis impact, (b) increases the interval between hand impact and pelvis impact, and (c) decreases the impact velocity of the pelvis. Furthermore, the two participants who avoided a fall did so by using a two-step skipping or jumping movement that yielded a wide final stance. This sequence is similar to the "side-step" response to lateral platform translation described by Maki et al. (2000), which was especially common among elderly participants, and for perturbations applied while "walking in place." However, while we observed a distinct aerial phase in this step recovery pattern, they did not, perhaps due to the smaller perturbation strengths involved in their study.

It is difficult to elicit realistic falls in a laboratory environment, and our study had important limitations. First, although our participants were unable to predict the magnitude or direction of the perturbation, they did know that their balance would be perturbed in some manner. Thus, their preparedness (or "intentional set") was different from that which typically accompanies a real-life fall, and this may have altered their response. This is an inevitable limitation (given ethical considerations) on any laboratory experiments involving balance perturbations. Second, our participants performed a relatively basic motor task (quiet standing) before the onset of the perturbation, and additional studies are required to determine how fall mechanics are affected by the

execution of more complex tasks, such as walking or turning (Smeesters, Hayes, & McMahon, 2001a), or holding an object (Bateni, Zecevic, McIlroy, & Maki, 2004). Third, we examined only sideways perturbations to balance, and future studies are required to determine how falling patterns are influenced by the strength and direction of the perturbation (Allum, Carpenter, Honegger, Adkin, & Bloem, 2002; Hsiao & Robinovitch, 1998). Fourth, due to the compliant nature of the landing surface, we did not directly measure impact forces, but instead we estimated impact severity from impact sequence and impact velocity.

In summary, results from this laboratory study suggest that sideways falls in young adults commonly produce direct impact to the hip region, and that the severity of hip impact is reduced by initial impact to the hand and by stepping, both of which were common. These protective responses may help to explain why, in contrast to the elderly, fall-related hip fractures are relatively rare (but not unknown;(Kannus, Leiponen, Parkkari, Palvanen, & Jarvinen, 2006)) in young adults—even in sports such as soccer and basketball, where unpadded sideways falls are relatively common. An important goal for future studies is to determine how fall movements are affected by aging and specific neurological or musculoskeletal impairments. Are the common protective responses we observed preserved with aging, or replaced with alternative strategies? How do age-related declines in strength and reaction time influence the efficacy of these responses (DeGoede et al., 2001b; Robinovitch et al., 2005)? A related but unexplored question concerns the practical benefit of exercise programs designed to "re-train" elderly men and women to fall in a way that is more protective to the hip. To address these questions, we will need to design experiments that are realistic enough to elicit natural falling

responses, but safe enough for older adults to participate. This might be achieved through the use of hip protectors and additional protective gear, harnesses and elastic tethers to slow the rate of fall descent, or by having participants fall into a swimming pool or extremely soft mattress (although attention must also be directed to the possibility of injury, such as strains or sprains, during the balance recovery and descent phase of falling). Parallel efforts should seek improved evidence of movement strategies during real-life falls from video recordings in high-risk environments (Holliday, Fernie, Gryfe, & Griggs, 1990) and wearable sensor systems (Bourke, O'Brien, & Lyons, 2006; Holliday et al., 1990).

Table 4-1: Frequency of impact to various body parts in trials that resulted in falls

Body part	*Frequency (n=42) (%)*
Head	0
Left hip	93
Left hand	98
Right hand	64
Left elbow	60
Right elbow	0
Left knee	60
Right knee	2
Buttocks	2

Table 4-2: Frequency of various impact sequences during sideways falls that resulted in hip impact

Impact sequence	*Frequency (n=39) (%)*
Left knee; left hand; left hip	31
Left hand; left hip; right hand	31
Left knee; left hand; left hip; right hand	23
Left hand; left hip	10
Left knee; left hip; left hand	5
Left hip; left hand	0

Figure 4-1: Experimental setup and motion profile for falling experiments. (a) During the experiment, the participant stood upon a rigid platform mounted flush to surrounding gymnasium mats, which were concealed under a rubber sheet. Drapes were placed on both sides of the room to prevent the participant from guessing the nature of the perturbation. (b) A linear motor was used to translate the rubber sheet suddenly to the right, initiating imbalance (dashed lines=reference command profile; solid lines=actual movement profile from a typical trial, measured from reflective markers on the rubber sheet).

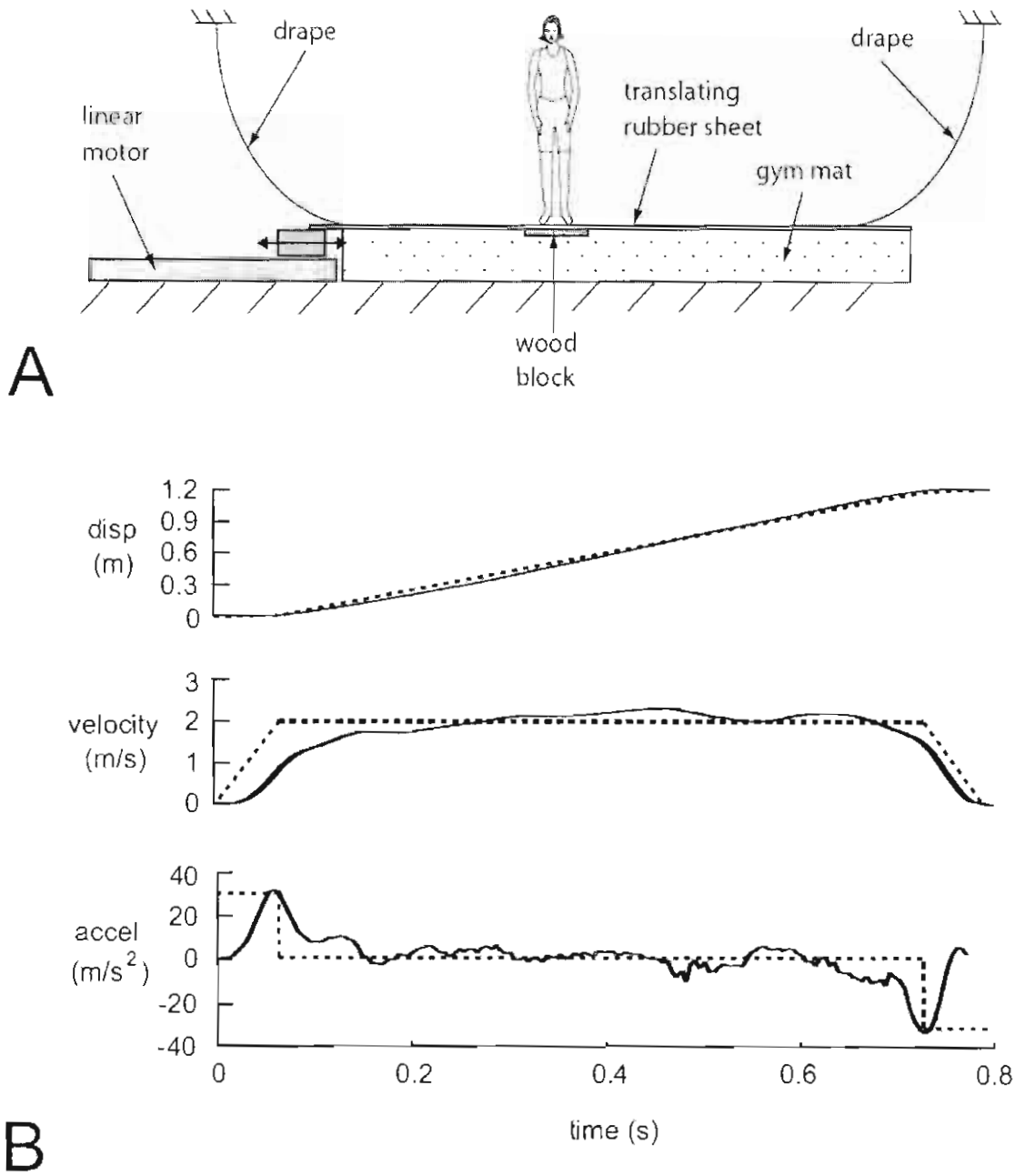
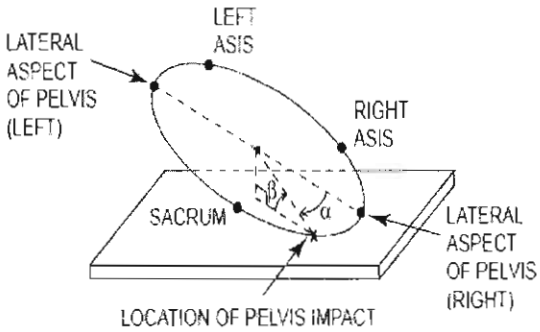
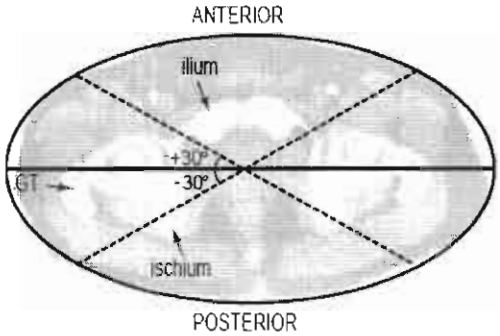


Figure 4-2: (a) The hip proximity angle (α) reflected how near the point of impact was to the lateral aspect of the pelvis. To calculate α, we first identified an ellipse whose circumference passed through the sacrum, right ASIS, and left ASIS markers. The lateral aspects of the pelvis were assumed to coincide with the endpoints of the major axis of this ellipse. We then identified the site of pelvis impact as the lowest point on the circumference of this ellipse, at the time of impact. Finally, we defined α as the angle between the site of pelvis impact and the nearest lateral aspect of the pelvis, measured within the plane of the ellipse. (b) Cross-section of the pelvis showing anatomical landmarks. Values of α greater than 30° should result in significant loading of the ilium, and values of α<−30° should load the ischium. (c) Histogram showing the observed distribution of α.



A



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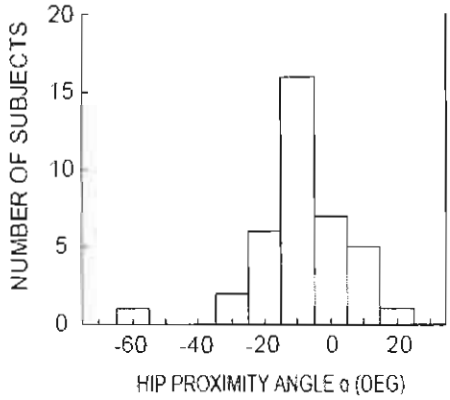


C

Figure 4-3: Interval between hand and pelvis impact versus pelvis impact velocity. Pelvis impact velocity was normalized by the estimated free-fall impact velocity ($v=\sqrt{2gh}$, where g=9.81 m/s2 and h=height of the hip marker). The average interval between hand impact and pelvis impact was 50 ms (SD=40), and the longer the interval the lower the pelvis impact velocity ($r=-0.6$; $p<0.001$). On average, participants who were able to complete a step before their fall (n=9) had a longer interval between hand impact and pelvis impact ($p=0.001$), and a lower pelvis impact velocity ($p=0.02$). Two trials involving no step were not included due to hand marker drop-out.

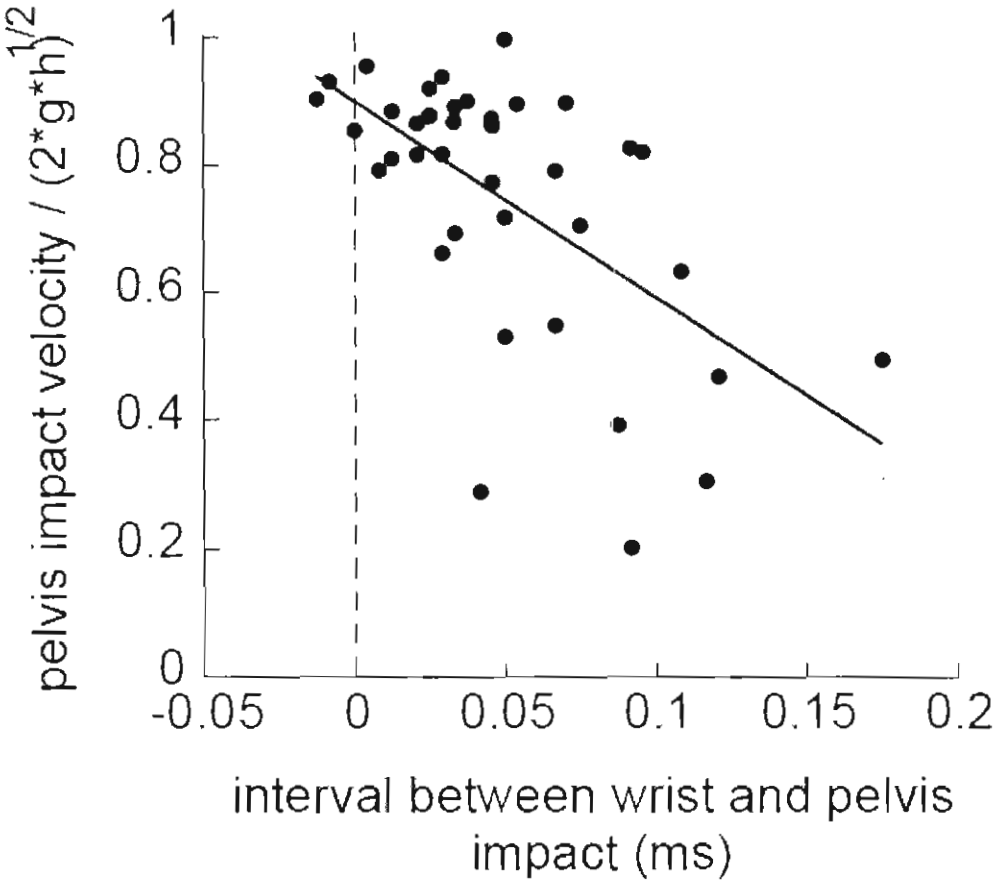


Figure 4-4: Typical landing configurations during sideways falls. Individual panels illustrate (a) a cross-forward step followed by a fall, (b) a cross-behind step followed by a fall, (c) a partial step followed by a fall, and (d) a skipping movement resulting in balance recovery.

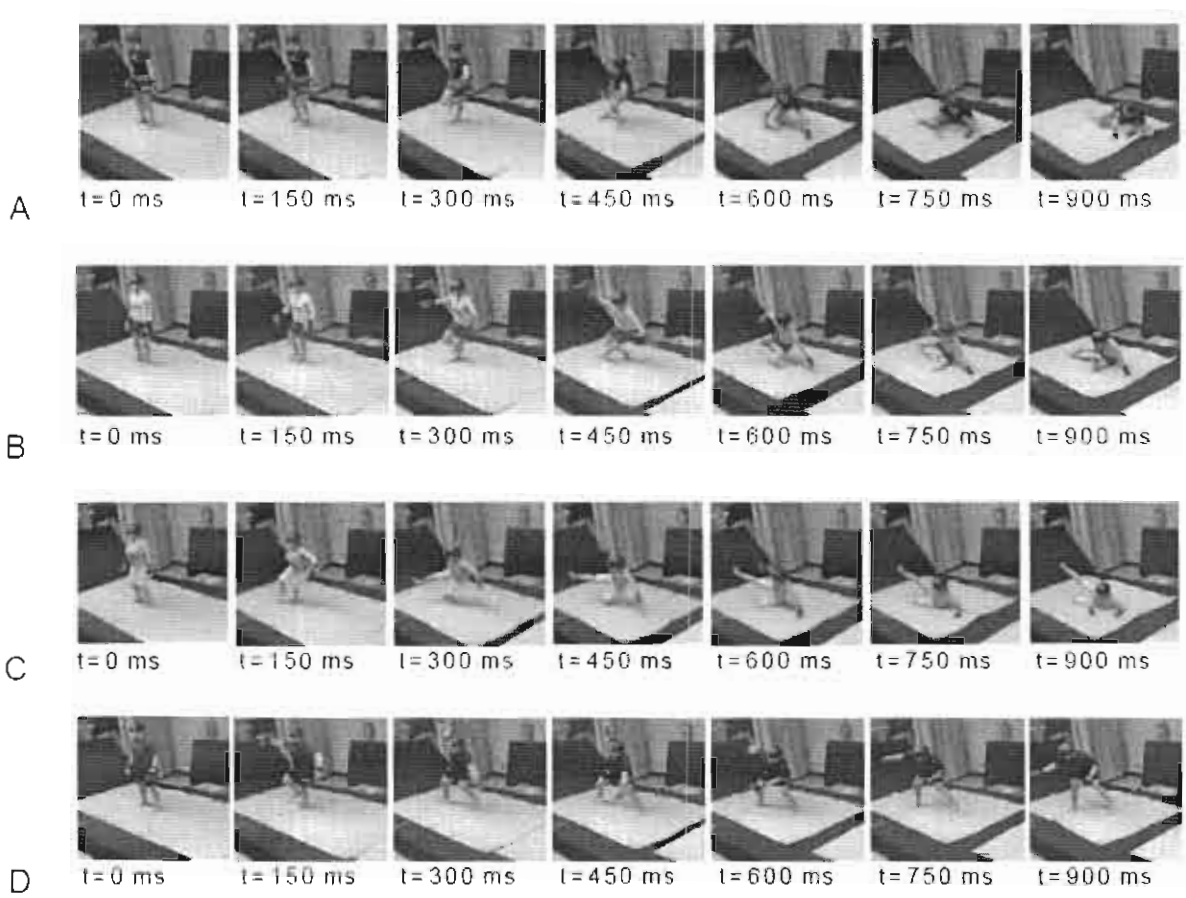
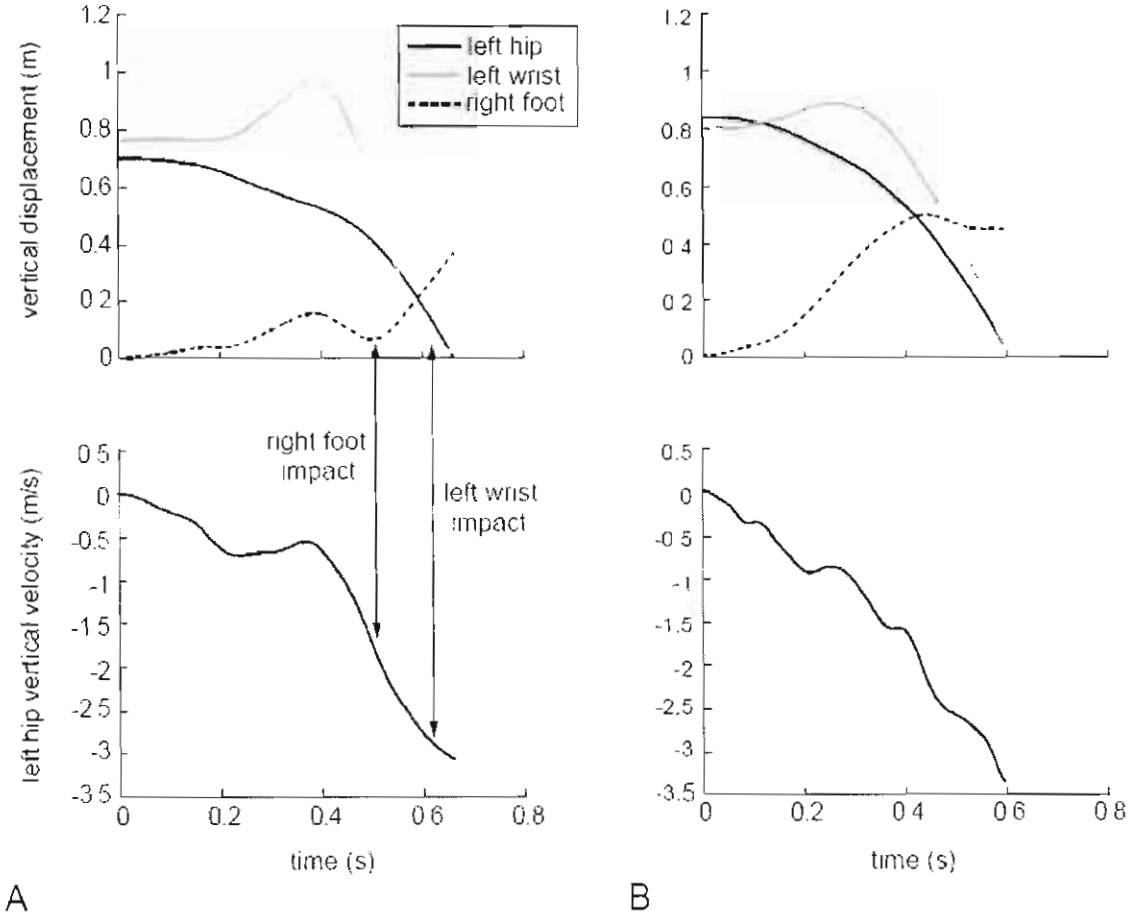


Figure 4-5: Temporal variation in the vertical position of the left hip, left hand, and right foot markers (top) and vertical velocity of the left hip marker (bottom; negative velocity implies downward movement) for (a) trial involving a full step and

(b) trial involving no step. The beginning of the traces (t=0) represents the onset of perturbation and the end of the trace represents the instant of pelvis impact. In (a), the slope of the velocity trajectory is decreased by initial impact to the right foot and left hand.



CHAPTER 5 SAFE LANDING DURING A FALL: THE TIME TO INITIATE POSTURAL RESPONSES AFFECTS THE ABILITY OF YOUNG ADULTS TO AVOID HIP IMPACT

5.1 Introduction

Accidental falls are a major cause of injury. However, most falls do not result in significant injury, despite the large available energy. This is due in part to humans being able to arrange the body segments during descent into a safe landing configuration. For example, laboratory studies have shown that the duration of a fall from standing ranges from 500 to 1000 ms, which is sufficient for young adults to execute a range of triggered or voluntary "safe landing" responses, ranging from breaking the fall with the outstretched hand, to rotating forwards or backwards during a sideways fall to avoid hip impact (Feldman & Robinovitch, 2007b; Hsiao & Robinovitch, 1998; Robinovitch et al., 2003). The muscle activations that govern these responses appear to be timed carefully to allow the energy of the fall to be absorbed safely, based on estimation of the time to contact (DeGoede, Ashton-Miller, Schultz, & Alexander, 2002; Dietz & Noth, 1978; Dietz, Noth, & Schmidtbleicher, 1981; Dyhre-Poulsen & Laursen, 1984; Feldman & Robinovitch, 2007b; Greenwood & Hopkins, 1977; Hsiao & Robinovitch, 1998; Laursen, Dyhre-Poulsen, Djorup, & Jahnsen, 1978; Lee, 1976; Melvill Jones & Watt, 1971; Santello, 2005). Such responses are probably triggered by vestibulo-ocular and vestibulospinal signals (Bisdorff et al., 1999; Greenwood & Hopkins, 1977; Melvill Jones & Watt, 1971; Muller & Weed, 1916), and shaped by higher cortical centres, based on

factors such as the direction and height of the fall, the stiffness of the landing surface, or the presence of nearby obstacles (Horak & Nashner, 1986; Maki, McIlroy, & Fernie, 2003).

However, we have limited understanding of human's ability to modify the landing position of the body during a truly unexpected fall, since previous laboratory studies of falling had participants plan their falling strategy prior to the onset of the fall, or involved self-initiated falls (Sabick et al., 1999; van den Kroonenberg et al., 1996), or instructed participants in the desired landing technique before the fall was initiated (Robinovitch et al., 2003). This is substantially different from a real-life fall, where one rarely has the ability to plan a descent strategy before imbalance occurs, and time delays in selecting and initiating the response may render a specific safe landing strategy ineffective.

In the current study, we addressed this issue by testing the hypothesis that the ability of young women to avoid hip impact (by rotating forward or backward during descent) is influenced by the time, relative to the onset of the fall, when this response is initiated. We found that the decision to rotate needed to occur within 200 ms of the onset of the fall in order to rotate at least 30 deg, and thereby avoid hip impact. This was due primarily to an invariable reaction time in initiating rotation after realizing the intent to do so, and limits on the maximum rate of rotation and duration of the fall. This study provides important evidence of the strict temporal demands that govern safe landing responses, to help guide the development of exercise interventions to prevent fall-related injuries in older adults.

5.2 Methods

5.2.1 Participants

Fifteen young women participated in the experiment, ranging in age between 20 and 32 yrs (mean = 23 ± 4(SD) yrs), in body mass from 44 to 80 kg (mean = 62 ± 12 kg), and body height from 158 to 183 cm (mean = 166 ± 8 cm). We studied women because they are approximately three times more likely than men to suffer hip fracture over the lifespan (Papadimitropoulos, Coyte, Josse, & Greenwood, 1997). However, safety precautions prevented us from including elderly women in our study. All participants were free of diagnosed neurological disease or debilitating orthopedic conditions, and were able to follow simple instructions in English. All participants provided written informed consent, and the experimental protocol was approved by the Research Ethics Committee of Simon Fraser University.

5.2.2 Experimental Protocol

During the experiment, sideways falls onto a gymnasium mat were initiated by suddenly releasing a tether (by means of an electromagnet (Warner Electric model PB500, South Beloit, IL; break release time ~15 ms)) which supported the participant at a 10-deg sideways lean (Figure 5.1). Participants were instructed prior to release to try to achieve a landing configuration that matched the image projected on a 110 x160 cm screen in front of them, which appeared starting just before, during, or after the initiation of the fall. In particular, during the trials we randomly varied the time that the image first appeared relative to the time of tether release. The earliest cue delivery was 300 ms before release (cue time = -300 ms), the latest was 300 ms after release (cue time = +300 ms), and intermediate values were -200, -200, 0, +100, and +200 ms. At the start of the

session, we explained that, if the image appeared of a person resting on her buttocks and hands (insert A to Figure 5.1), the participant was to rotate backward during descent to land in that position (cue type = BACKWARD). If the image appeared of a person resting on her hands and knees (insert B to Figure 5.1), the participant was to rotate forward during descent to land forward-facing on her hands and knees (cue type = FORWARD). If no image was displayed, the participant was to fall sideways with no rotation to land on her hip (cue type = "DO NOT ROTATE"). In all trials, participants were instructed to avoid attempts at balance recovery (e.g., by stepping). Participants first performed one practice trial for each combination of cue type and cue time. Data were then acquired for two additional trials for each combination of cue type (3 conditions) and cue time (7 conditions).

For each trial, we used a seven-camera, 60-Hz motion measurement system (ProReflex; Qualysis Inc., Glastonbury, CT, USA) to acquire 3-dimensional positions of 18 reflective markers placed at the top of the head, sacrum (L5/S1 junction), and bilaterally at the acromion process, lateral epicondyle of the humerus, distal end of the radius, anterior-superior-iliac spine (ASIS), greater trochanter, lateral epicondyle of the femur, lateral malleolus, and third metatarsal. To detect the instant of tether release, we acquired force data at 960 Hz from a load cell located in-line with the tether (model 31; Sensotec, Columbus, OH, USA). The participant's initial leaning position before release was standardized using a large goniometer.

5.2.3 Data Analysis

We determined the instant that the pelvis first impacted the ground using the marker position technique previously described (Robinovitch et al., 2003). Briefly, this

technique involved acquiring a series of "control trials" with each participant of body marker positions as the participant lay still on a 0.25 in thick plywood board placed over the mat. We acquired measures in six different configurations: lying sideways on each side, sitting, lying prone, lying supine, and resting in a quadruped position on the hands and knees. We then used the data from these control trials to determine participant-specific look-up tables to indicate the occurrence of impact to the pelvis. In particular, the instant of pelvis impact was taken as the time frame where the vertical coordinate of the right or left trochanter marker (whichever was first to impact) descended below that of the corresponding marker in the appropriate control trial (e.g., the prone control trial if the participant landed prone).

Furthermore, for each trial, we estimated the initial point of impact on the pelvis by first creating a virtual ellipse whose circumference passed through the sacrum, right ASIS and left ASIS markers, and minor axis passed through the sacrum marker (Figure 5.2). We then determine the lowest point on the circumference of this ellipse at the instant of pelvis impact (Robinovitch et al., 2003; Smeesters, Hayes, & McMahon, 2001b). We defined the pelvis impact angle (α) as the angle from the major axis of the ellipse to the line joining the centre of the ellipse and the point of impact. The angle α reflects how close the individual came to directly impacting the lateral aspect of the pelvis, with a value of zero (or 180 deg) reflecting direct impact to the lateral aspect of the pelvis. We also calculated the total axial rotation of the pelvis (β) from tether release to pelvis impact (Figure 5.2), the maximum axial rotational velocity of the pelvis during descent (β_DOT_MAX), the time interval between tether release and onset of rotation (T_ROTATE_RELEASE), the time interval between cue delivery and onset of rotation

(T_ROTATE_CUE), and the time interval between tether release and pelvis impact (T_FALL). In calculating T_ROTATE_RELEASE and T_ROTATE_CUE, we defined the onset of rotation as the instant when axial rotation of the pelvis (β) exceeded 5 deg in either (the correct or incorrect) direction.

5.2.4 Statistical Analysis

We used 2-way repeated-measures ANOVA to test whether the cue type (BACKWARD versus FORWARD) and cue time influenced α, β_DOT_MAX, T_ROTATE_RELEASE, T_ROTATE_CUE, and T_FALL. When significant effects were observed, post-hoc comparisons were examined using a Bonferonni adjustment to the assumed significance level. We used absolute values of α and β_DOT_MAX in the ANOVA analysis, to assess whether cue type affected the magnitudes of total rotation and peak rotational velocity. ANOVA analyses for β_DOT_MAX, T_ROTATE_RELEASE, and T_ROTATE_CUE did not include trials involving a cue time of +300 ms, since these rarely involved an observable onset of pelvis rotation. We conducted all statistical tests with SPSS analysis software (Version 12.0, Chicago, IL, USA), and regarded $p < 0.05$ to indicate significant effects.

5.3 Results

Mean values of all outcome variables are shown in Table 5.1 for trials involving the BACKWARD cue type and in Table 2 for trials involving the FORWARD cue type.

We found that the absolute value of the pelvis impact angle α was strongly influenced by cue time ($F_{6,84} = 77.4$; $p < 0.001$), but did not depend on cue type ($F_{1,14} = 0.26$; $p = 0.62$). Although pair-wise comparisons revealed significant differences in α

between all cue times ($p \leq 0.004$), the reduction in α was rather moderate between cue times of -300 ms and 0 ms, and increased dramatically between cue times of 0 to +300 ms (Figure 5.3). For example, the average value of α over the FORWARD and BACKWARD conditions was 68.3 (SE = 3.5) deg for cue time = -300 ms, 50.0 (SE = 4.9) deg for cue time = 0 ms, and 8.2 (SE = 2.5) deg when cue time = +300 ms.

T_ROTATE_RELEASE was significantly influenced by cue time ($F_{5,70} = 50.6$; $p < 0.001$), but did not depend on cue type ($F_{1,14} = 0.056$; $p = 0.816$). T_ROTATE_RELEASE averaged 184 (SE = 11) ms for cue time = -300 ms, and increased in a nearly linearly fashion to 567 (SE = 30) ms for cue time = +200 ms. Pair-wise comparisons revealed significant differences ($p \leq 0.005$) between all cue times except -200 versus -100 ms ($p = 0.61$), 0 versus +100 ms ($p = 0.09$), and +100 versus +200 ms ($p = 0.30$). T_ROTATE_CUE also associated with cue time ($F_{5,70} = 4.4$; $p = 0.002$), and did not associate with cue type ($F_{1,14} = 0.056$; $p = 0.816$). However, pair-wise comparisons revealed that there was no difference ($p \geq 0.06$) in T_ROTATE_CUE between cue times of -200, -100, 0, +100, and +200 ms; only the -300 ms cue time had larger (slower) values when compared to the -100 ms ($p = 0.017$), 0 ms ($p = 0.004$), and +100 ms ($p = 0.02$) cue times. Thus, for cue times between -200 and +200 ms, the time required for participants to initiate rotation with respect to the onset of the cue was nearly constant, averaging 422 (SE = 14) ms. However, when considered with respect to the onset of the fall, delays in cue delivery caused nearly equal delays in the onset of rotation.

The maximum rotational velocity β_DOT_MAX did not associate with either cue time ($F_{5,70} = 2.219$; $p = 0.062$), or cue type ($F_{1,14} = 0.034$; $p = 0.857$). The total duration of the fall T_FALL did not associate with cue time ($F_{6,84} = 1.350$; $p = 0.244$), but did

associate with cue type ($F_{1,14} = 16.929$; $p = 0.001$), averaging 942 (SE = 17) ms in the FORWARD condition and 917 (SE = 14) ms in the BACKWARD condition.

Figure 5.4 illustrates for a typical participant how cue time affected the temporal variation in pelvis angle during descent. All traces are synchronized to the time of tether release (t = 0), and end at the instant of pelvis impact (when the angular position is equal to α). It is evident that the onset of rotation relative to the time of tether release (T_ROTATE_RELEASE) increases with increasing values of cue time. However, cue time does not influence the subsequent peak slope or velocity of rotation (β_DOT_MAX). There is a consistent slowing (deceleration or braking) or even a reversal in angular velocity commencing approximately 150 ms before impact, a phenomenon observed in all participants. Finally, near symmetrical behaviour is observed in the FORWARD condition (top traces, with positive values of angular position) and BACKWARD condition (bottom traces, with negative values). For this participant, α was similar for cue times of -300, -200, and -100 ms, and thereafter declined rapidly to a negligible value at cue time = +300 ms.

5.4 Discussion

In the current study, we examined whether the ability of young women to change the direction of a sideways fall – and avoid hip impact - by rotating forward or backward during descent depended on the time following the onset of the fall when the decision was made to rotate. We found that both forward and backward rotation were effective for avoiding hip impact, but only if the intent to rotate was realized within a short time window (200 ms) after fall initiation.

We found that the absolute value of pelvis impact angle (α) was influenced strongly by the time of cue delivery. This was due primarily to the effect of cue time on the time interval between fall initiation and commencement of rotation (T_ROTATE_RELEASE), since cue time had no effect on the maximum rotational velocity (β_DOT_MAX) or the total fall duration (T_FALL), and had only a minimal effect on the time interval between cue delivery and commencement of rotation (T_ROTATE_CUE).

We also found that the absolute value of α did not depend on cue type, indicating that participants were just as efficient at avoiding hip impact by rotating forward as backward (an observation similar to that reported in a previous study, where participants were instructed on the direction of rotation before the fall was initiated; (Robinovitch et al., 2003). We also observed similar values in FORWARD and BACKWARD conditions of T_ROTATE_RELEASE, T_ROTATE_CUE, and β_DOT_MAX, suggesting that similar biomechanical and neural constraints govern these two motor programmes.

The need for an early commitment to rotate during descent can be understood by partitioning the motor programme into three distinct stages, and considering the time intervals and kinematic characteristics of each: (1) an initial reaction time defined by the interval between fall initiation and commencement of rotation, (2) a forceful response involving rapid axial rotation of the trunk and pelvis, and (3) a final halting of rotation where body segments are decelerated and maintained in a final position in preparation for impact. During our falling trials, the total descent time (T_FALL) averaged 929 (SE = 15) ms, and did not depend on cue time or cue type. However, the final approximately 150 ms of the fall was consistently marked by a reversal in the angular acceleration of the

pelvis, and thus was unavailable to contribute additional rotation. (A possible mechanism for this late-stage braking was rotation of the upper extremities into an optimal landing position, and the corresponding rotation of the trunk and pelvis in the opposite direction, due to conservation of angular momentum). Therefore, meaningful rotation needed to be completed in the first 780 ms of the fall. However, following the onset of the cue (signifying the decision to rotate), an average of 422 (SE = 14) ms was required to initiate pelvis rotation; this interval (T_ROTATE_CUE) was also unaffected by cue type and only slightly influenced by cue time. Thus, a delay in realizing the intent to rotate of 358 ms or more after the onset of the fall would typically leave no time for rotation. Furthermore, the maximum speed of pelvis rotation (β_DOT_MAX) averaged 209 (SE = 7) deg/s, and again did not associate with cue type and cue time. Therefore, if we consider that at least 30 deg of pelvis rotation is required to avoid direct impact between the ground and the greater trochanter of the proximal femur (a reasonable assumption based on anatomical considerations; Figure 5.3c), this required an additional interval of at least 143 ms (if we instead regard 45 deg of rotation as essential, the required rotation interval increases to 215 ms). Thus, in our experiments, participants were unable to avoid hip impact if they were required (due to the cue timing) to delay more than 200 ms in realizing the intent to rotate forward or backward following the onset of the fall.

There are several strengths to this study. We focused on the high-risk situation of a sideways fall, and examined how a specific protective response (axial rotation during descent) influences an important measure of hip fracture risk – the orientation of the pelvis at impact. By structuring our study design on a performance measure – the ability to rotate forward or backward and thereby achieve a large value of α – and manipulating

the time when participants were cued to rotate, we were able to determine how time delays in realizing the intent to rotate influence the body's landing configuration, and the threshold delay, beyond which participants were unable to avoid hip impact. We used a visual image projected on a wall to instruct participants to rotate forward or backward. This mimics the real-life situation of relying on visual information of the ground surface location and texture to shape the landing response. Furthermore, we used a choice reaction time paradigm to minimize the potential for participants to pre-plan their responses, and simulate the real-life scenario of tailoring the landing response based on visual input of the environment. However, the two-choice response paradigm may have introduced delays in measured reaction time (approximately 100 ms longer when compared to a simple reaction time). An alternative paradigm would have involved a single response direction with embedded catch trials (where no response was required) to reduce the risk for pre-planning.

There are also important limitations to this study. As is the case with any laboratory study of falling, our conclusions may be specific to our sample population and to the type of fall elicited in our experiments. During most real life loss-of-balance episodes, one likely directs initial attention towards balance recovery (e.g., through stepping or grasping) before considering options for safe landing. Although we randomized the timing of tether release and the nature and delay in cue delivery, our participants were aware that each trial would involve a fall, and focused their initial attention on safe landing, rather than balance recovery.

Furthermore, while the young women who participated in this study (when given sufficient preparation time) could successfully utilize the safe landing responses we

examined, we are unable to confirm whether the same protective responses could be successfully utilized by elderly women. Nevertheless, it is interesting to consider how isolated declines with age in response times (Spirduso, 1980; Welford, 1988) may impair one's ability to effectively use the forward and backward rotation strategies. Among the studies that are specific to falling, both Robinovitch et al. (2005) and DeGoede et al. (2001a) found that elderly women averaged about 110 ms slower than young women in moving their hands into a protective position for breaking a fall. Our current results indicate that a similar increase in the delay to initiate pelvis rotation would certainly increase the challenge, but not eliminate the feasibility, of converting a sideways fall to a forward or backward fall, and thereby greatly reduce the risk for hip fractures.

Table 5-1: Mean values of outcome variables (with standard deviations shown in parentheses) for trials involving the FORWARD rotation cue type and various times of cue delivery.

Outcome variable	Time of cue delivery						
	-300 ms	-200 ms	-100 ms	0 ms	+100 ms	+200 ms	+300 ms
α (deg)	75.3 (4.0)	66.8 (3.3)	62.2 (5.1)	50 8 (4.9)	37.4 (5 7)	13 9 (3 1)	5.3 (2.8)
T_ROTATE_RELEASE (ms)	181 (5)	270 (9)	304 (8)	368 (13)	471 (23)	615 (27)	--
T_ROTATE_CUE (ms)	481 (5)	470 (09)	404 (8)	368 (13)	371 (23)	415 (27)	--
β_DOT_MAX (deg/s)	211 (41)	225 (32)	215 (54)	206 (38)	212 (68)	179 (65)	--
T_FALL (ms)	959 (70)	960 (70)	956 (80)	944 (80)	931 (80)	922 (80)	922 (60)

Table 5-2: Mean values of outcome variables (with standard deviations shown in parentheses) for trials involving the BACKWARD rotation cue type and various times of cue delivery.

Outcome variable	Time of cue delivery						
	-300 ms	-200 ms	-100 ms	0 ms	+100 ms	+200 ms	+300 ms
α (deg)	-61.2 (4.7)	-61.9 (5.8)	-59.3 (6 1)	-48 5 (6.8)	-38.0 (5 8)	-21 9 (3.8	-11.1 (4.0)
T_ROTATE_RELEASE (ms)	184 (50)	261 (90)	329 (80)	443 (110)	470 (140)	566 (110)	--
T_ROTATE_CUE (ms)	484 (50)	461 (90)	429 (80)	443 (110)	370 (140)	366 (110)	--
β_DOT_MAX (deg/s)	-201 (52)	-217 (49)	-216 (47)	-220 (26)	-213 (64)	-190 (69)	--
T_FALL (ms)	904 (60)	903 (70)	906 (60)	933 (60)	934 (60)	924 (60)	912 (50)

Figure 5-1: Experimental schematic for sideways falling experiments. A sideways fall onto a gymnasium mat was initiated by suddenly releasing a tether, which supported the participant at a 10 deg lean angle. The participant was instructed that, if image (A) was displayed, she should rotate backward during descent to land on the buttocks (converting the sideways fall to a backward fall); if image (B) was displayed, she should rotate forward during descent to land on the hands and knees (converting the sideways fall to a forward fall); if no image was displayed, she should fall sideways with no axial rotation during descent.

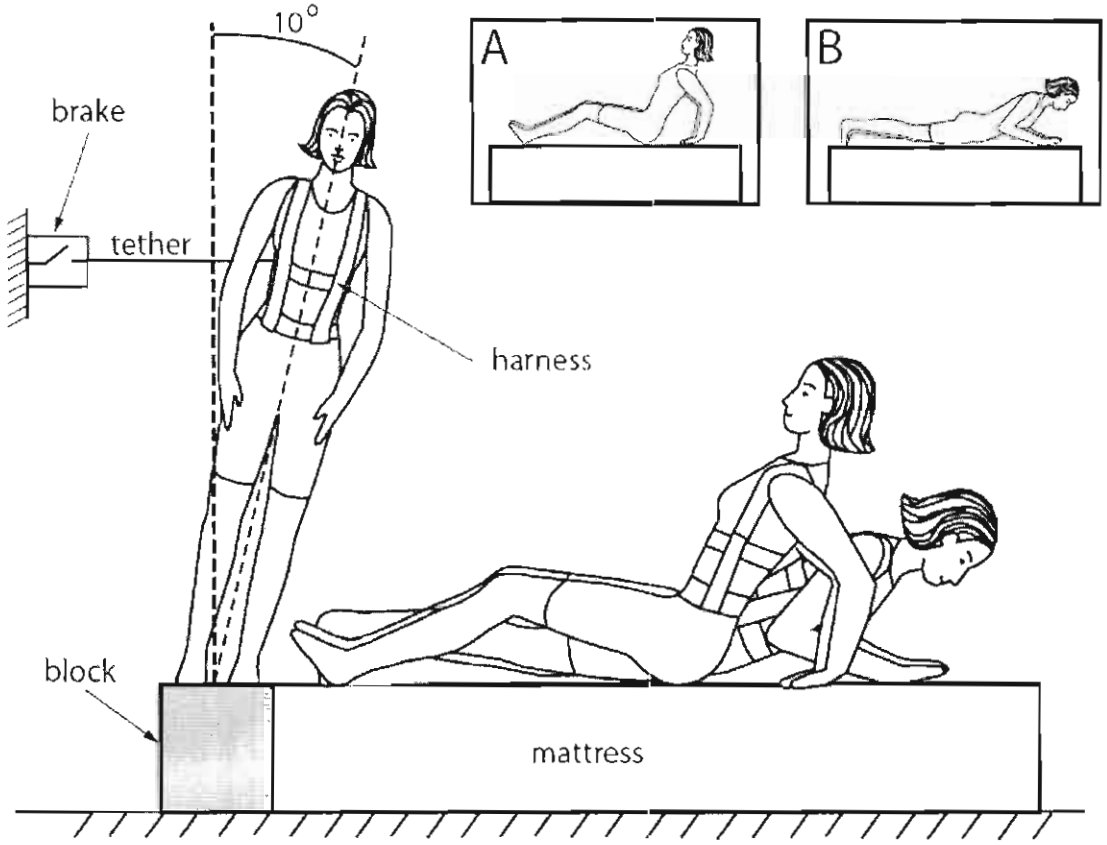


Figure 5-2: Definition of the pelvis impact angle. We estimated the location of pelvis impact as the lowest point on the circumference of the ellipse passing through the sacrum, right ASIS, and left ASIS at the time of impact. The lateral aspects of the pelvis were estimated to coincide with the endpoints of the major axis of the ellipse. The pelvis impact angle (α) indicated how near the site of pelvis impact was to the lateral aspect of the pelvis. A value of $\alpha = 0$ deg indicated direct impact to the lateral aspect of the pelvis, $\alpha = 90$ deg indicated impact to the anterior aspect of the pelvis, and $\alpha = -90$ deg indicated impact to the posterior aspect of the pelvis. The angle β indicates the time-varying axial rotation of the pelvis during descent.

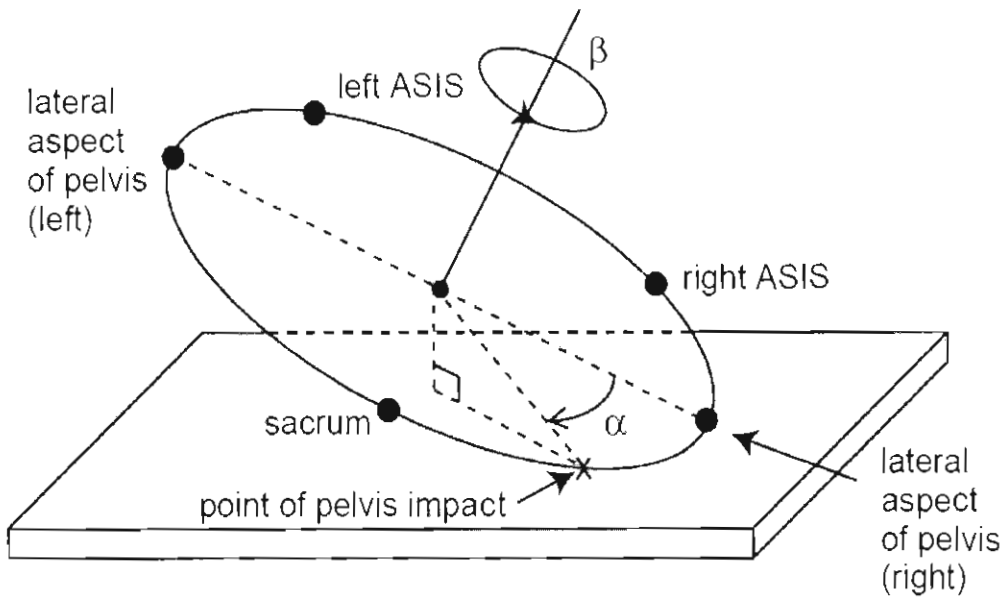


Figure 5-3: Absolute values of the pelvis impact angle (α) for FORWARD and BACKWARD rotation trials. (A) The absolute value of α was influenced by the time of cue delivery but not the cue type (FORWARD versus BACKWARD). Error bars represent ± 1 SE. Note the sharp decline in α for cue times of +200 ms and +300 ms. (B) Typical fall kinematics in the BACKWARD and FORWARD conditions when the visual cue is presented 200 ms before or 200 ms after tether release. Images display body configurations every 150 ms, up to 900 ms. (C). The same results as in part (A) are shown in the cross-section of the pelvis. Each band shows the mean value ± 1 SE in α. A = cue presented 100-300 ms before release (data combined from all three conditions); B = cue presented at instant of release; C = cue presented 100 ms after release; D = cue presented 200 ms after release; E = cue presented 300 ms after release. The upper quadrant represents forward rotation and the lower quadrant represents backward rotation. The dashed lines set the limits of the defined hip impact area (± 30 deg).

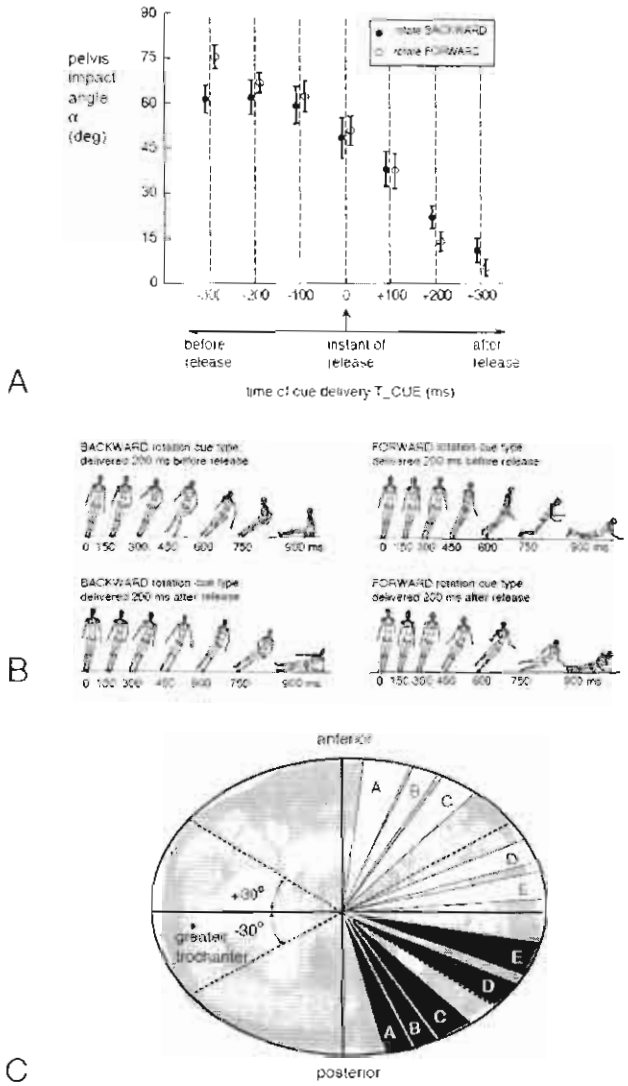
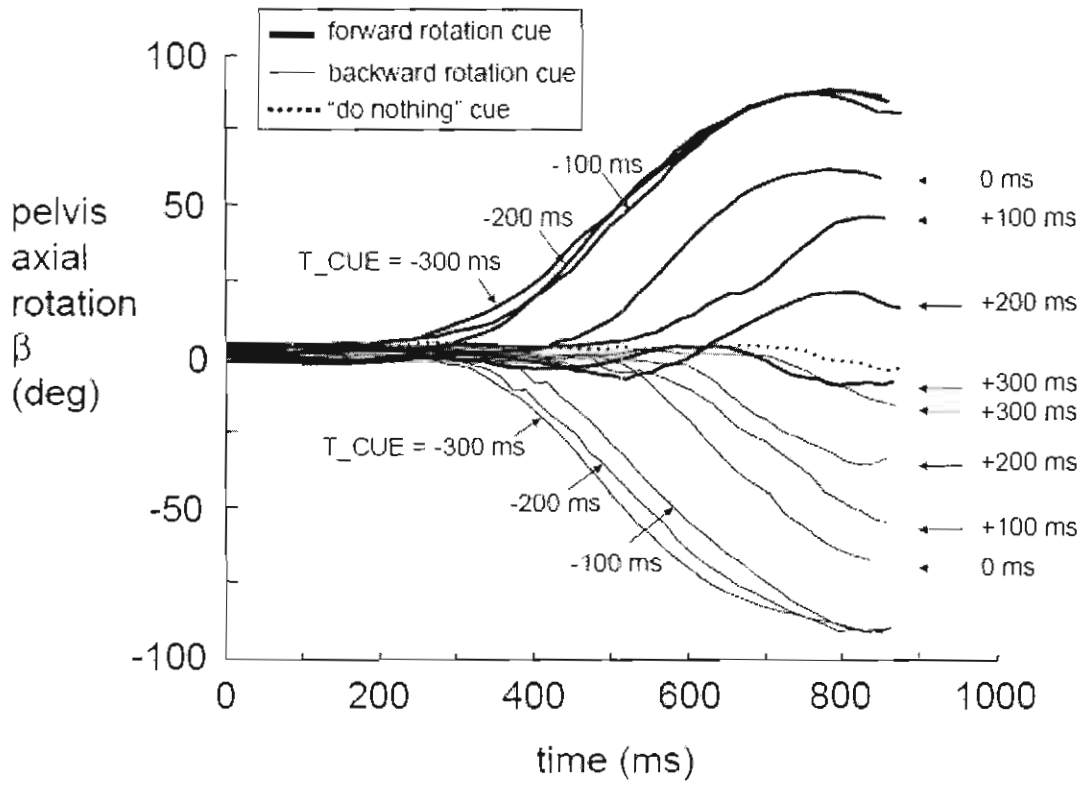


Figure 5-4: Temporal variations in axial rotation of the pelvis (β) during descent as a function of delay in cue delivery, for a typical study participant (all traces are from the same participant). The tether is released at time t = 0, and pelvis impact occurs at the end of the trace (in this participant, the duration of each fall was between 850 to 890 ms). The dashed line indicates a control trial, and the time interval between tether release and cue delivery is labelled for all other traces. The earlier the cue delivery, the greater the axial rotation. See text for further explanation.



CHAPTER 6 CAN YOUNG ADULTS ACCURATELY RECALL THE MECHANICS OF THE FALL IMMEDIATELY AFTER A SIDEWAYS FALL OCCURS?

6.1 Introduction

Falls and hip fractures in the elderly population create enormous human and financial costs. In the U.S. and in Canada there were over 320,000 and 28,000 admissions to hospitals for hip fractures respectively. More than 50% of those who fall and sustain a hip fracture never recover the mobility they enjoyed before the fall, leading to a loss of independence and social interaction, and 25% die within the year following the hip fracture (Tinetti et al., 1997; Zuckerman, 1996). It is estimated that the annual cost related to falls to the Canadian and U.S. health care systems is about $3.2 billion and $19 billion respectively (Stevens, Corso, Finkelstein, & Miller, 2006).

An important goal for researchers is to understand the relative importance of fall mechanics versus bone strength in the ethiology of hip fracture. To address this, investigators have utilized interviewed-administered questionnaires to acquire self-reported information on fall characteristics in individuals who fall and fracture, versus those who fall and did not fracture (Greenspan et al., 1998; Greenspan et al., 1994; Keegan et al., 2004; Nevitt & Cummings, 1993b; Parkkari et al., 1999; Schwartz et al., 1998; Wei et al., 2001). Such questionnaires have enquired about the circumstances leading to the fall, the direction of the fall (e.g., forward, sideways, backwards, or straight down), whether attempts were made to recover balance (e.g., by taking a step), which body parts received the impact of the fall or contacted the ground, and whether attempts

were made to break the force of the fall (e.g., through the use of an outstretched hand). Some studies have also acquired information on bone strength, based primarily on bone mineral density (BMD) measures. These studies have consistently concluded that fracture risk depends as strongly on fall mechanics as it does on bone strength. For example, direct impact to the hip increases fracture risk by an average of 30-fold, and impacting the hand or knee decreases fracture risk by an average of 3-fold, in contrast, a one standard deviation decline in hip bone mineral density increases fracture risk by an average of 3-fold (Greenspan et al., 1998; Greenspan et al., 1994; Nevitt & Cummings, 1993b; Wei et al., 2001).

Since most falls are unwitnessed (Hayes et al., 1993; Nurmi et al., 1996; Wagner et al., 2005), the validity of the studies relying on self-reported information depends on the assumption that study participants can accurately recall the mechanics of their falls days or even weeks after the incident had occurred. For example, in a study by Nevitt and Cummings (1993), the average interval between a hip or a wrist fracture and interview was 2.1 months (SD = 1.4). For falls without fracture, the average interval between fall and interview was 1.4 months (SD = 1.2). In a study by Schwartz and colleagues (1998), the median duration of time from a fall to interview was 1.4 months for individuals who suffered a hip fracture and 2.1 months for individuals who sustained a fall with no hip fracture (control group). Greenspan and colleagues (1994) obtained information about the characteristics of the fall within 5 days for participants with hip fractures and 10 days for controls. Moreover, Wei and colleagues (2001) interviewed patients with hip fracture within 3 days of hospital admission, and individuals who fell without hip fractures were interviewed within 2 weeks after confirmation of the fall.

Understanding the accuracy of self-reported falls data is critical, since inaccurate or biased responses can lead to erroneous conclusions and can fail to detect important associations (and effective preventive measures). Self-reported fall narratives are strongly called upon to guide intervention efforts by health professionals such as environmental modification, medication reviews, exercise prescription, or provision of safety equipment such as hip protectors. Also, it is important to understand the percent of hip fractures caused by direct impact to the hip region in order to understand the potential benefit of hip protectors, which are designed to protect only the lateral side of the thigh. Furthermore, it may be possible to train individuals to fall in a manner that is more protective to the hip, wrist, or shoulder, but only if we can identify the characteristics of safe versus injurious falls.

The goal of this study was to determine whether young individuals are able to accurately recall specific characteristics of their fall during a structured questionnaire conducted immediately after the individual experienced an unexpected fall during their participation in the "balance competition" described in chapter 4. We focused specifically on the following questions: (1) how accurate are individuals in recalling the direction of fall and the attempts they made to recover balance? and (2) how accurate are individuals in recalling the occurrence of impact to specific body parts?

6.2 Materials and Methods

6.2.1 Participants

The participants in our falling experiments consisted of 44 young individuals (31 women and 13 men) ranging in age from 19 to 26 years (mean = 21 ± 2 (SD) years), body mass from 43 to 112 kg (mean = 65 ± 14 kg), and body height from 149 to 187 cm (mean

= 168 ± 9 cm). All participants were undergraduate students enrolled in a biomechanics course, who volunteered to participate in the experimental "balance competition" described in Chapter 4. Participants were free of diagnosed neurological disease or debilitating orthopaedic conditions and 87% of them participated at least twice a week in some type of exercise for fitness. All participants provided written informed consent, and the experimental protocol was approved by the Research Ethics Committee of Simon Fraser University.

6.2.2 Experimental Protocol

During the balance test, the participant stood barefoot on top of a rubber sheet that, without warning, was made to translate horizontally to the participant's right (Figure 6.1) by means of a linear motor (T4D motor, Trilogy System Corporation, Webster, Texas, USA). The total displacement of the linear motor was set to 1.15m (actual mean value = 1.14 m, SD = 0.04), the terminal velocity was set to 2.2 m/s (actual peak value = 2.22 m/s, SD = 0.07), and the initial acceleration and terminal deceleration was set to 30 m/s^2 (actual mean value = 29.2 m/s^2, SD = 2.0). The perturbation was initiated at a random interval between 1 and 10 seconds after the participant had adopted a stable posture on the platform. By using such a large perturbation in an unexpected direction, "the balance test" was designed to elicit an unexpected fall, and succeeded in doing so in 42 of the 44 participants. Thus, although the participant was unaware of this, the experiment was designed to elicit an unexpected fall, as opposed to testing the participant's "balance" per se.

We used several precautions to minimize participants' ability to preplan a postural response, in order to mimic a real-life fall. The only instruction we provided to the

participant was: ''your balance will be perturbed, and your goal is to maintain your balance.'' Moreover, the equipment was masked to prevent the participant from anticipating what was to come. The participant was guided to stand on a rigid platform mounted flush to the gymnasium mat landing surface, and the entire landing surface was covered with a rubber sheet. Drapes were used to conceal the linear motor and related hardware. No information was provided about the compliance of the ground surface, or the nature of the perturbation. In order to focus on naive responses, each participant performed only one trial, and no practice trials were allowed. Furthermore, all testing was completed in a single afternoon, and participants who were waiting to be tested were kept on a separate room and therefore prevented from having contact with those who completed the testing.

We used an eight-camera, 240Hz motion measurement system (Motion Analysis Inc., CA, USA) to acquire three-dimensional positions of 18 reflective, skin-surface markers located at the head, sacrum (L5/S1 junction), and bilaterally at the acromion process (shoulder), lateral epicondyle of the humerus (elbow), distal end of the radius (hand), anterior–superior–iliac spine (ASIS), greater trochanter (hip), lateral epicondyle of the femur (knee), lateral malleolus (ankle), and third metatarsal (toe). Three markers were also placed on the translating rubber sheet overlying the impact surface. We also used a digital camcorder to record video images of the falls. As a safety measure, participants wore helmets and wrist guards (over which the head and hand markers were secured).

As soon as possible after the participant had recovered from the fall, and typically within three minutes, a short questionnaire was administered which surveyed the participants

knowledge of what had happened during the fall.. In particular, we asked "How well do you remember the details of your fall?", with potential answers of (a) very well, (b) well (some details hazy), (c) not much, and (d) not at all. This question was used as a measure of recall confidence. We also asked "What was the direction of the fall (at the onset of descent)?", with potential answers of (a) forwards, (b) backwards, (c) sideways, (d) straight down, and (e) cannot recall, "Which body parts impacted the ground?" with potential answers of (a) head, (b) back/trunk, (c) buttocks, (d) left or right knee, (e) hip, (f) hand, (g) elbow, and (h)shoulder, and "After losing your balance, what attempts did you make to prevent falling?", with potential answers of (a) one step, (b) multiple steps, (c) jumped, (d) shifted weight, (e) other, (f) and cannot recall.In order to measure participant's prior knowledge or perception of the study protocol, we asked "Were you aware that the floor was going to move before it did?", with potential answers of (a) yes and (b) no.

6.2.3 Data Analysis

Data analysis focused on 41 falls (two participants avoided falling, and one did not complete the questionnaire). For each fall, we inspected motion data from the motion measurement system and digital camcorder to determine whether a step was successfully executed in an attempt to recover balance (defined as lifting and repositioning the foot on the ground prior to impacting a hand, knee, or pelvis) and whether a specific body part impacted the floor (defined as a body part's marker crossing a virtual line located 5 cm above the mean vertical position of three reflective markers located on top of the impact surface). This information was then compared to the corresponding self-reported information provided in the questionnaires.

6.2.4 Statistical Analysis

Our first question was addressed descriptively by examining the frequency of participants' correct recall regarding direction of the fall, balance recovery attempts, and impact to specific body parts. Our second and third questions were addressed by using a generalized linear mixed model (GLIMMIX) for binomial distribution procedure to determine if there were differences in proportion of correct recall among different body parts and if the frequency of correct recall correlated with recall confidence. Tukey post-hoc analyses were performed when significant differences were found. We regarded $p < 0.05$ to indicate significant effects. All statistical tests were conducted with statistical analysis software (SAS for Windows version 9.1, SAS Institute Inc., USA).

6.3 Results

Ninety-three percent of falls involved direct impact to the left hip, 98% involved impact to an outstretched hand, and 61% involved left knee impact (Table 6.1). A more extensive biomechanical analysis of the falls is reported elsewhere (Feldman & Robinovitch, 2007).

The majority of participants (98%) were able to accurately describe the direction of the fall (only one participant reported falling straight down instead of sideways). This result was not due to participants' prior knowledge of the protocol since only 6 (14%) participants believed the floor was going to move and 5 out of these 6 participants thought the floor was going to tilt or shake in an unpredictable direction. Furthermore, 51% of the participants were able to accurately describe if they took a step before impacting the floor. Of the remaining 49%, 27% reported not remembering and 22% demonstrated incorrect recall.

In regards to body parts, 71% of participants answered correctly on whether the left hip impacted the ground or not, 51% were correct on whether the left hand impacted the ground or not, and 39% were correct on whether the left knee impacted the ground or not (Table 6.1). Furthermore, 9% of participants reported impacting the head when in fact none of the participants experienced a head impact during the trials.

With respect to the left knee, left hip, and left hand, only 6 participants (15%) were correct in their recall of impact to all three body parts, 20 participants (49%) were correct for only two body parts, 8 participants (20%) were correct for only one body part, and 7 participants (17%) were incorrect for all 3 body parts (Figure 6.2).

When asked "How well do you remember the details of your fall", 26% of participants claimed to remember the details of their fall "very well," 56% "well (some details hazy)," and 17% "not much". Percent of correct recall based on participants' confidence in recall accuracy is given in Table 6-2. For the left hip, left hand, and right hand, recall accuracy was higher for the "medium" and "low" confidence than for "high" confidence. However, for the left knee and left shoulder, recall accuracy was higher for the "high" confidence.

6.4 Discussion

Our results indicate that while 98% of participants were able to accurately describe the direction of the fall, only 15% were accurate in recalling impact to hip, hand, and knee, and 51% were accurate in reporting taking a step to try to recover balance. We also found that recall accuracy from participants claiming to remember the details of their fall "very well" appeared similar in accuracy than those who claimed to remember "well" or "not much".

Results from previous studies relying on self-reported data suggest that falling sideways as opposed to backwards or forwards increases fracture risk by 6-fold, falls that result in direct impact to the hip region increases fracture risk 30-fold, and impacting the hand or knee decreases fracture risk 2 to 3-fold (Greenspan et al., 1994; Nevitt & Cummings, 1993b; Schwartz et al., 1998; Wei et al., 2001). Even though the results from these studies are biomechanically sensible (avoidance of hip impact and sharing the impact energy with the hands and knees should decrease the risk of hip fracture; Feldman & Robinovitch, 2007), the true effect of these variables on hip fracture risk may be difficult to measure due to the limited accuracy of self-reported fall characteristics.

How would the results of previous studies change if the recall accuracy were assumed to be the same as the young participants in this study? In the case of Schwartz et al. (1998) for example, the percentage of men that reported hitting the knee at the time of their fall was 14% for the case group (sustained a hip fracture due to a fall) and 36% for the control group (no hip fracture due to a fall). If we assume that the recall accuracy for impact to the knee is only 39% (Table 6.1), the adjusted percentage of men hitting the knee at the time of the fall would be 58% for the cases group and 52% for the control group. Keep in mind that the assumption about recall accuracy made here is based on young individuals' recall about the details of their fall minutes after the event; however, because of the much longer intervals between the fall and the interview as well as higher levels of memory impairment, we expect the recall to be even less accurate for elder individuals.

Even though most studies recognize the limitations of relying in self-reported data, they have underrated their importance. For example, Wei and colleagues (2001) claim to

have decreased recall bias or inaccurate statements by excluding participants with mental dysfunction, very old age (older than 85 year), and a time lapse of more than 3 months since the fall. Furthermore, Nevitt and Cummings (1993) claimed to have enhanced recall of fall circumstances because of the generally high level of education and mental status of the study's participants. Hayes et al. (1993) attempted to decrease recall bias by obtaining information within days of the fall and to record as unknown all the information about which the faller was uncertain, assuming that faller that were confident about their response must be correct.

There are important limitations to this study. Because of safety concerns, we only recruited young participants. Also, participants were falling onto gym mats and, as opposed to what may happen on a real situation, the falls resulted in no injury or pain. This prevented participants from using this information to help recall the body parts that may have impacted the ground. However, the occurrence of any pain or injury may also bias the accuracy of recalls. For example, a person that sustained a hip fracture may be more likely to report an impact to the hip region than a person that had a fall without a hip fracture, even if the hip region impacted the floor on both cases. Therefore, special attention should be paid to hematomas and bruises on fallers that sustained a hip fracture as well as the ones that did not sustained a hip fracture.

Recalling even a fall event seems to be a challenge for older adults, especially if they have cognitive problems and/or if the fall results in no injury (Cummings, Nevitt, & Kidd, 1988; Fujimoto et al., 2000; Haga et al., 1996; Hale, Delaney, & Cable, 1993; Lachenbruch, Reinsch, MacRae, & Tobis, 1991; Mackenzie, Byles, & D'Este, 2006; Peel, 2000) . For example, Cummungs, Nevitt, and Kidd (1998) showed that when participants

were interviewed at the end of a 12-month follow up, depending on the time period of recall (3,6,or 12 months), 13% to 32% of those with falls confirmed by weekly follow-up and home visits did not recall falling. Recall was better for the preceding 12 months than for 3 or 6 months and those with lower scores on the Mini-Mental State Examination were more likely to forget falls. Similarly, Hale, Delaney, and Cable (1993) showed that for the 3-, 6-, and 12-month periods, respectively, 31 percent, 44 percent, and 89 percent of participants who reported falls weekly by postcard recalled at least one fall at the end of one year. Peel (2000) reported that falling was more likely to be remembered if an injury had occurred (87% of those who had an injurious fall subsequently recalled falling, compared with 62% recall in the non-injured group of fallers).

Research is currently underway in our laboratory to seek improved evidence of accuracy in self-reporting mechanisms of falls by comparing real-life falls from video recording in high-risk environments with testimonials from the corresponding fallers. We are also focusing on the development of wearable sensor that can acquire three-dimensional data on body segment movement during a fall without the need for the elder individual to be in the view of a video camera.

In conclusion, the results of this study indicate that while fallers were able to recall the direction of their fall, the task of visualizing and recalling balance recovery attempts and body parts that impacted the floor is difficult and prone to error. Furthermore, we found that recall accuracy from participants claiming to remember the details of their fall "very well" appeared similar in accuracy than those who claimed to remember "well" or "not much".. Researchers and healthcare professionals should be aware of the limitations of self-reported characteristics of falls specially related to impact at different body parts.

Therefore, physical exam findings (regarding bruising, abrasion, or tenderness) should be favoured over self reports as an indicator of fall impact sites.

Table 6-1: Actual and self-reported frequency of impact to specific body parts.

Body Part		Recall		**Total**
		Correct	Incorrect	
Left Shoulder	IMPACT	1	1	2
	NO IMPACT	29	10	39
	Total	30	11	41
Left Elbow	IMPACT	16	10	26
	NO IMPACT	2	13	15
	Total	18	23	41
Left Hand	IMPACT	21	19	40
	NO IMPACT	0	1	1
	Total	21	20	41
Left Hip	IMPACT	28	10	38
	NO IMPACT	1	2	3
	Total	29	12	41
Left Knee	IMPACT	15	10	25
	NO IMPACT	1	15	16
	Total	16	25	41
Head Impact	IMPACT	0	0	0
	NO IMPACT	37	4	41
	Total	37	4	41
Right Hand	IMPACT	11	13	24
	NO IMPACT	10	7	17
	Total	21	20	41

Table 6-2: Percent of correct recall based on participants' confidence in recall accuracy.

Body Part	Confidence in recall accuracy		
	High (N = 11)	Medium (N = 23)	Low (N = 7)
left elbow	36%	52%	27%
left hand	26%	61%	58%
left hip	55%	79%	73%
left knee	55%	34%	27%
left shoulder	83%	70%	73%
right hand	45%	48%	58%

Figure 6-1: Experimental setup and motion profile for the perturbation platform.

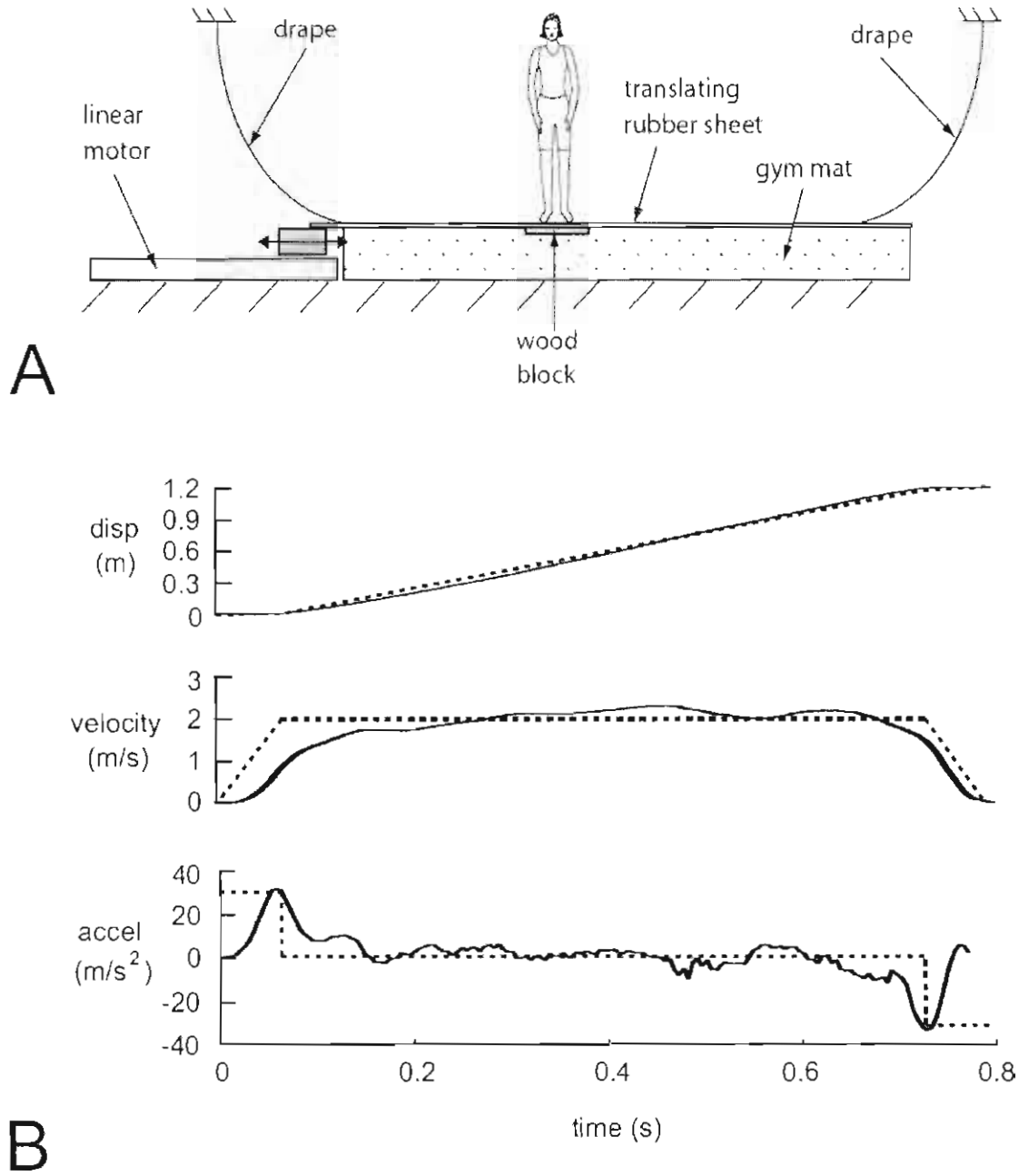
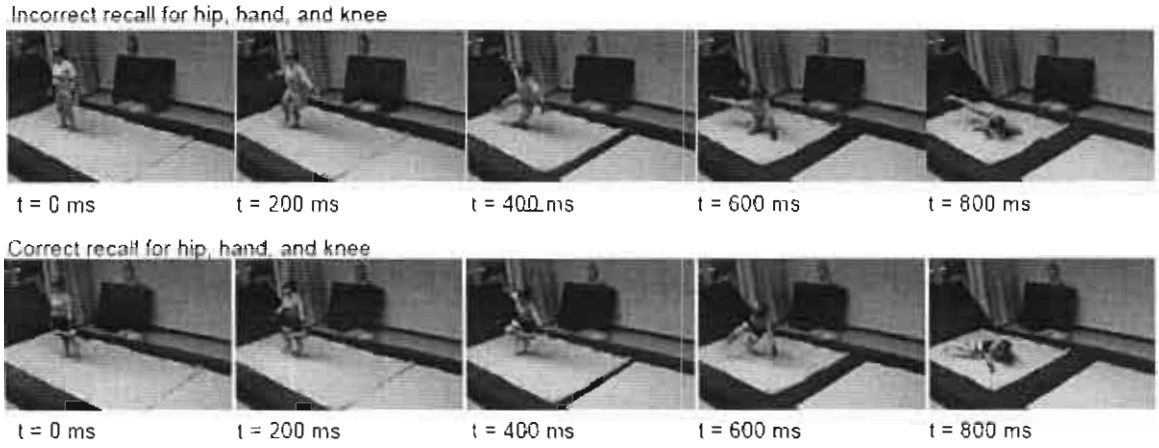


Figure 6-2: Landing configurations for participants with incorrect (top) and correct (bottom) recall for all three body parts, left hip, left hand, and left knee.



CHAPTER 7 AN ANALYSIS OF THE EFFECT OF STRENGTH AND REACTION TIME ON IMPACT SEVERITY DURING SIMULATED SLIPS AND TRIPS

7.1 Introduction

Falls and fall related injuries are a major health problem in the elderly population. Over 30 percent of community dwelling elderly and 50% of nursing home residents over age 65 fall each year (Speechley & Tinetti, 1991). Consequences of falls include fracture of the hip, spine, arm, pelvis, and wrist, head concussions, bruises, and lacerations. The injuries associated with falls in the elderly account for more than $1 billion in annual health care costs in Canada and $13 billion in the U.S. (National Advisory Council on Aging, 1999).

Slips and trips are the most common self-reported causes of falls in the elderly, collectively accounting for at least 50% of cases (Berg, Alessio, Mills, & Tong, 1997). Slipping can be defined as a sudden sliding of the foot over the ground surface due to insufficient frictional forces. Tripping can be defined as a loss of balance due to sudden collision of the foot with the floor and an obstacle (or surface irregularity) during the swing phase of walking.

Two factors often used to determine impact severity are impact velocity and time to impact. Previous investigators have demonstrated that impact velocity is a strong determinant of impact force and fracture risk during a fall (Robinovitch et al., 1991; Robinovitch, Hayes, & McMahon, 1997; van den Kroonenberg et al., 1995). As discussed in Chapter 5, another factor that determines injury risk is the time available

during descent to arrange the body segments into a safe landing configuration (e.g., break the fall with the outstretched hand).

The main goal of the current study was to examine the theoretical influence of fall type (slip versus trip) on these two measures of impact severity. A second goal was to determine how, in each of these cases, impact severity can be affected by the timing and magnitude of lower extremity torque generation during descent. To address these questions, I developed mathematical models of falls due to slips and trips from standing height. I then conducted a systematic sensitivity study to examine how impact velocity and fall duration depend on fall type (slip vs. trip), ankle torque magnitude, and reaction time in generating ankle torque during descent when perturbation of different magnitudes were applied.

7.2 Methods

Inverted pendulum models were developed to simulate a trip (Figure7.1A) and a slip (Figure 7.1B). The trip is a simplified model from the point where the foot strikes the floor or another object, moving the body's center of gravity (COG) with an initial angular velocity. Therefore, the trip model has only one degree of freedom (the lean angle θ). The slip is a simplified model from the point where the foot contact a sliding surface, moving the foot in one direction with an initial linear velocity and the body's center of gravity (COG) in an opposite direction with an initial angular velocity. The slip model has two degrees of freedom (the cart position x and the lean angle θ). Both models simulate falls where the knees and hips remain extended throughout descent, while rotation and energy absorption occurs at the ankles.

The equation of motion for the trip model is:

$$(mr^2 + ml^2)\,\ddot{\theta} - mgl\,\sin\theta = -T \qquad (1)$$

The equations of motion for the slip model are:

$$m\ddot{x} + ml\cos\theta\,\ddot{\theta} - ml\,\dot{\theta}^2\,\sin\theta = 0 \qquad (2)$$

and

$$(mr^2 + ml^2)\,\ddot{\theta} + ml\,\cos\theta\,\ddot{x} - mgl\,\sin\theta = -T\,, \qquad (3)$$

where m is the mass of the pendulum, l is the distance between the ankle joint and the pendulum center of gravity (COG), θ is the angle between the link and the vertical, x is the horizontal position of the cart, $\ddot{\theta}$ is the angular acceleration, $\ddot{x}$ is the horizontal acceleration of the cart, g is the gravitational constant (9.81 m/s^2), and T is torque at the ankles, positive in the counter clockwise direction.

Among the assumptions inherent in these models were: (1) that movement is restricted to the sagittal plane, (2) that the feet remain in contact with the ground throughout descent, and (3) that contraction of muscles spanning the ankles generates a net joint torque, which can instantly change in magnitude. Anthropometric parameters were set to match a typical adult female. In each model, the lengths, masses, and moments of inertia were representative of an adult female (height = 1.6 m, mass = 53.7 kg, moment of inertia = 11.5 kg m^2 (Winter, 1990).

For the purpose of comparing the slip and trip models, the initial kinetic energy for a given "perturbation strength" were the same for each model. This was achieved by varying $\dot{\theta}_i$ and $\dot{x}_i$ to simulate different perturbation strength, and corresponding values of kinetic energy at the onset of the fall (table 1). $\dot{\theta}_i$ and $\dot{x}_i$ were selected to simulate the range of gait velocities at the onset of a trip, or foot slip velocities that may be encountered in daily life (Brady, Pavol, Owings, & Grabiner, 2000). In all simulations, the initial configuration of the model was defined by θ_i = 5 deg, and x_i = 0, causing the initial gravitational potential energy of the body to be 481 J.

In all slipping simulations, the horizontal velocity of the body is zero throughout descent and COG falls straight down (Figure 7.1C). In contrast, the COG follows an arc trajectory during tripping simulations (Figure 7.1D). In both cases, the horizontal velocity at impact is zero. The following constraint equation was used in the slip model to simulate a slip from standing, where the initial horizontal velocity of the COG is zero. This was achieved by the following equation:

$$\dot{\theta}_i = \frac{\dot{x}_i}{\ell \cos\theta_i} \quad , \qquad (4)$$

where the subscript "i" refers to parameter time t=0. Finally, each model incorporated an ideal torque generator to simulate the net effect of bilateral (equal right and left side) contraction of muscles spanning the ankles. Torque magnitude varied between 0 and 200 Nm, and torque onset after fall initiation (reaction time) varied between 0 and 300 ms.

We used MATLAB to numerically integrate (using the Runge-Kutta 4th order method) the equations of motion. Each simulation proceeded until the occurrence of

pelvis impact, signified by the segment crossing the horizontal or until balance recovery, signified by the segment crossing the vertical.

The work performed at a given joint during descent was determined by numerically integrating the area under the torque-rotation curve. Joint work (W) was defined negative if the direction of torque was opposite to the direction of joint rotation. Checks were made to ensure that conservation of energy was maintained throughout all simulations, as defined by PE + KE – W = constant, where PE is the potential energy and KE is the kinetic energy of the body at any given time.

Impact severity was represented by the vertical (downward) component of the velocity of the pendulum COG at the instant the pendulum passed through the horizontal.

7.3 Results

The vertical velocity and acceleration of the pendulum COG during the early stages of descent was consistently higher in slips than trips, and the difference increased as perturbation strength increased (Figure 7.2).

Times to impact were longer for trips than slips for each combination of perturbation strength and torque magnitude (Figure 7.3). However, the difference decreased as perturbation strength increased. For example, in the zero torque condition, time to impact was 66% longer in the trip than slip in the perturbation strength = 0 condition (1.246 vs. 0.751 ms) and 41% longer in the perturbation strength = 3 condition (0.587 vs. 0.415 ms).

Increases in ankle torque caused delays in time to impact. This was especially striking when ankle torque reached a value close to that required for balance recovery and therefore more evident in the low than high perturbation strengths. In contrast, delays in

the onset of ankle torque (slower reaction times), caused a decrease in time to impact, and this was more striking for low perturbation strengths than high.

Impact velocities were higher for slips than trips for all but the zero torque condition (Figure 7.4). Increases in ankle torque also caused decreases in impact velocity and the effect was stronger in trips than slips. For example, in the perturbation strength = 2 and reaction time = 300 ms condition, there was 25% decrease in impact velocity for the trip (and 14% for the slip) when the ankle strength increased from 0 to 200 Nm. Finally, delays in ankle torque onset caused an increase in impact velocity, and this was more striking for high perturbation strength than low, and for slips than trips. For example, in the perturbation strength = 2 at ankle torque = 200 Nm condition, impact velocity increased by 33% for the slip (and 16% for the trip) when reaction time increased from 0 to 300 ms.

7.4 Discussion

Our simulation results suggest that, for given initial values of potential energy and kinetic energy, slips create greater likelihood for injury than trips for at least two reasons. First, impact velocities were higher for slips than trips. Higher impact velocities are associated with higher impact forces and fracture risk (Robinovitch et al., 1991; Robinovitch et al., 1997; van den Kroonenberg et al., 1995). Second, slips involve shorter fall durations than trips, due to the faster increase in vertical acceleration. In real life falls, a longer fall duration should improve the ability of individuals to utilize protective responses such as stepping and stumbling, and impacting the ground with the outstretched hand (Feldman & Robinovitch, 2007; Kim & Ashton-Miller, 2003).

Our results also indicate that the development of ankle torques during descent causes an increase in time to impact and decreases in impact velocities. However, this effect is much stronger for low than for high perturbation strengths. Significant delays in time to impact and decreases in impact velocities are restricted by the limited ability of the ankle muscles to generate torque, especially at high perturbation. Average peak attainable plantarflexor torques (associated with forward falls) are reported to range from 200 Nm (combining left and right ankle joints) for young females to 260 Nm for young males, while dorsiflexor torques (associated with backwards falls) range from 90 Nm for young females to 160 Nm for young males (Sepic, Murray, Mollinger, Spurr, & Gardner, 1986). Inversion and eversion torques (associated with sideways falls) range from 20 Nm for young females to 40 Nm for young males (Aydog, Aydog, Cakci, & Doral, 2004; Sepic et al., 1986).

Our results are supported by the experimental results of Nagata and Ohno (2007), who used crash test dummies to determine fall severity in backwards falls (caused by applying different accelerations to a moving cart) involving different frictional characteristics between the feet and the ground. They found that for trials where all the joints were fixed (similar to the one-link model with no torque generation presented here), duration of falling was, on average, 15% longer for the non-slippery surface than the slippery surface (0.98 ms vs. 0.83 ms), and that the duration of falling decreased as the maximum floor velocity increased. They also found that there was no clear difference in head impact velocity between the slippery and non-slippery cases (average 6.19 vs. 6.47 m/s). This is similar to the findings from the zero torque conditions of the current study. However, contrary to our findings, impact velocities remained almost constant as

the velocity of the floor movements was increased. This may be due to the fact the foot of the dummy lost contact with the ground prior to impact (in the case of slippery surface, the dummy rotated in the air).

A previous study by Robinovitch and colleagues (2005) have reported that the average time required to move the hands to break the fall was 530 ms for healthy young women and 615 ms for healthy elderly women when instructed to contact a shoulder height target as quickly as possible while standing. Based on our simulations with no ankle torque, most healthy elderly women would have enough time to break a fall caused by slip when perturbation strength = 0 (time to impact = 0.751 s) but not when perturbation strength = 1 (time to impact = 0.573 s) or 2 (time to impact = 0.481 s) or 3 (time to impact = 0.415 s). Conversely, other than a perturbation strength = 3 (time to impact = 0.587 s), there is enough time to break a fall caused by a trip (time to impact > 700 s).

Time to impact and impact velocities seen during the simulations can be compared to values found in chapters 4 and 5. In chapter 4, participants stood barefoot on top of a rubber sheet that, without warning, was made to translate horizontally at a velocity of 2 m/s to simulate a slip. Time to pelvis impact ranged from 550 ms to 776 ms while the simulations using the slip model at perturbation 2 (velocity = 2 m/s) ranged from 481 ms to 521 ms. Impact velocities ranged from 0.81 to 4.10 m/s for the experiments using young subjects and from 3.65 to 3.89 m/s for the simulations. In chapter 5, a sideways fall onto a gymnasium mat was initiated by suddenly releasing a tether, which supported the participant at a 10 deg lean angle. This is comparable to the trip model with perturbation strength = 0 (initial angular velocity of the COG = 0). Impact times averaged

929 ± 67 ms for the experiments but were longer than 1245 ms for the simulations. Pelvis impact velocities averaged 2.69 ± 0.30 m/s for the experiments and higher than 3.76 m/s for the simulations. The longer times to impact and lower impact velocities seen during the experiments may be due to the use of fall protective responses such as breaking the fall with the hands and knees. While we regard our simulations as appropriate representations of real life falls, in future studies would be useful to examine the effect of more realistic rates of torque development and torque maintenance. We also assumed that the feet remain fixed on the ground and that moments can be generated at the ankles throughout descent. However, in real life falls, both feet may rise off the ground. Lastly, I restricted my efforts in this preliminary study to a one-link model but future studies should investigate the effect of multiple links on fall severity of slips versus trips.

Our study provides important information on biomechanical differences between falls caused by slips and trips. We found that falls from slips poses a higher impact severity than trips and this may translate into higher rates of injury. Slips were associated with shorter fall durations and higher impact velocities.

Table 7-1: Initial KE based on initial slip velocity and a corresponding initial angular velocity for the slip model and initial angular velocity for the trip model.

Perturbation strength	Slip model		Trip model	Initial KE (J)
	Initial slip velocity $\dot{x}_i$ (m/s)	Initial angular velocity $\dot{\theta}_i$ (rad/s)	Initial angular velocity $\dot{\theta}_i$ (rad/s)	
0	0	0	0	0
1	1	1.095	0.500	7.1
2	2	2.190	1.002	28.4
3	3	3.285	1.502	63.9

Figure 7-1: Representation of the slip (A) and trip (B) models. Illustration of slipping (c) and tripping (D) pattern to the ground. The small filled circles represent the path of the center of gravity (COG).

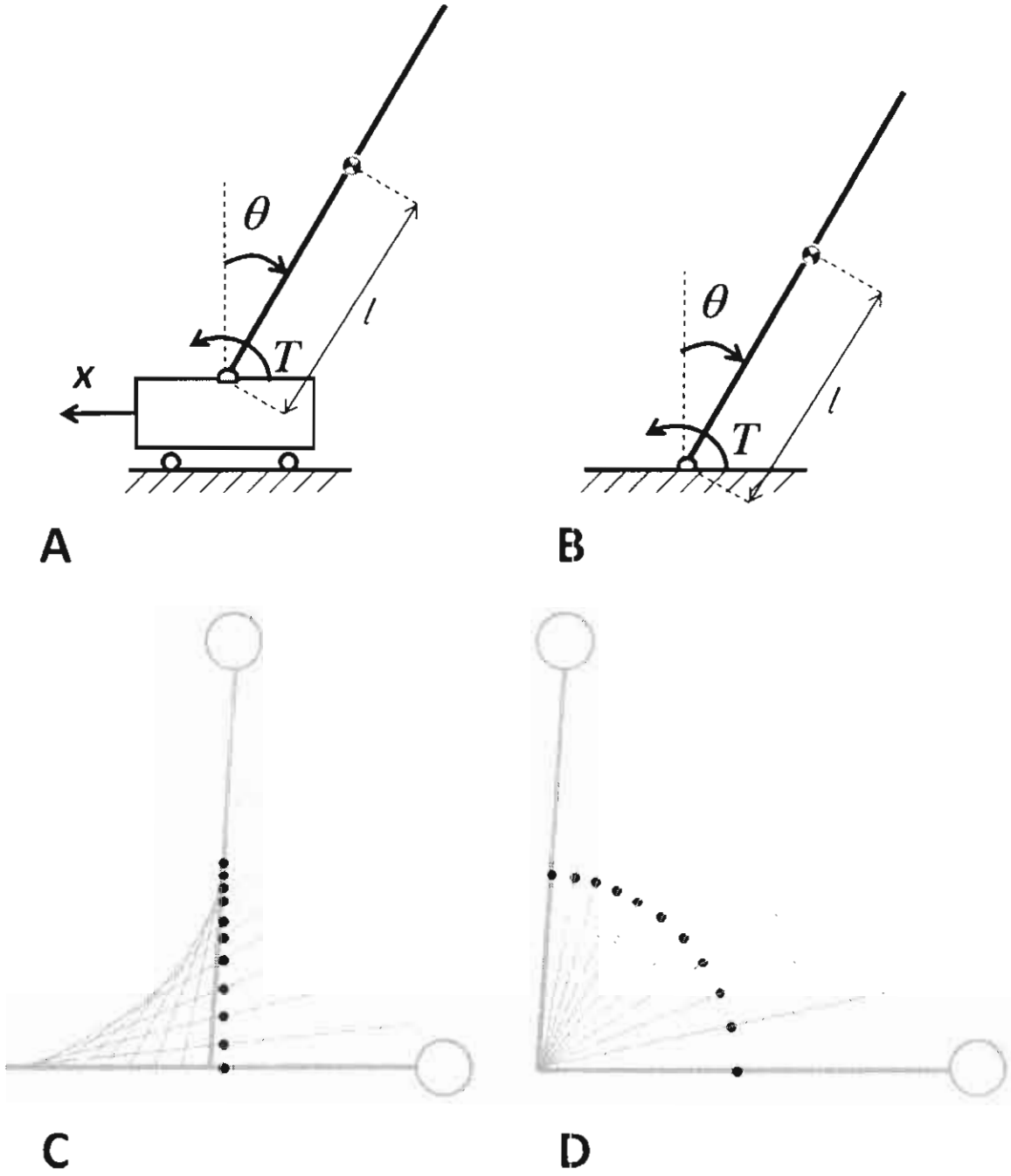


Figure 7-2: Traces of displacement, velocity, and acceleration over time for simulations with different strength perturbation and 0 ankle torque.

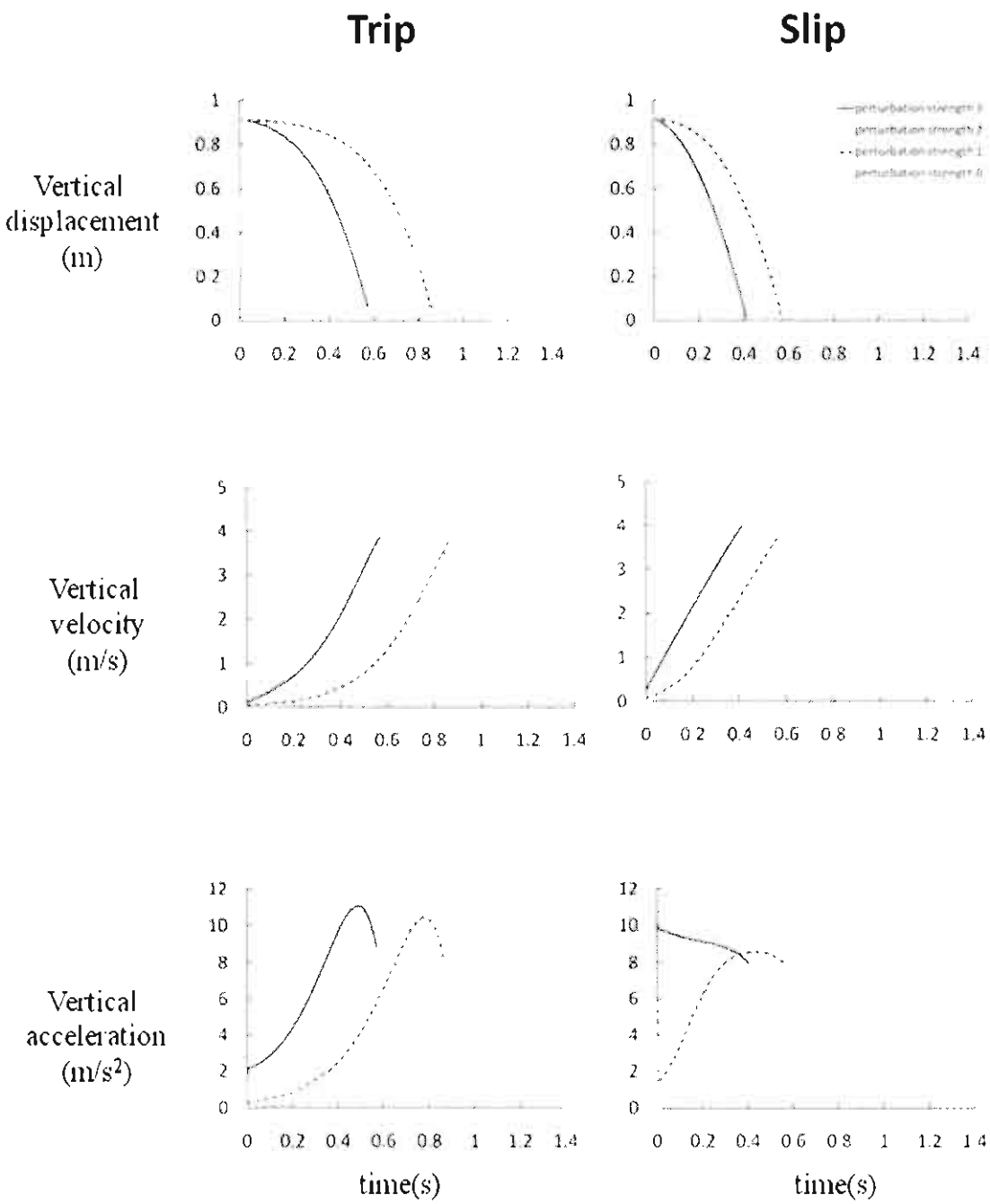


Figure 7-3: Sensitivity traces displaying the effect of ankle torque generation, reaction time, and perturbation strength on time to impact.

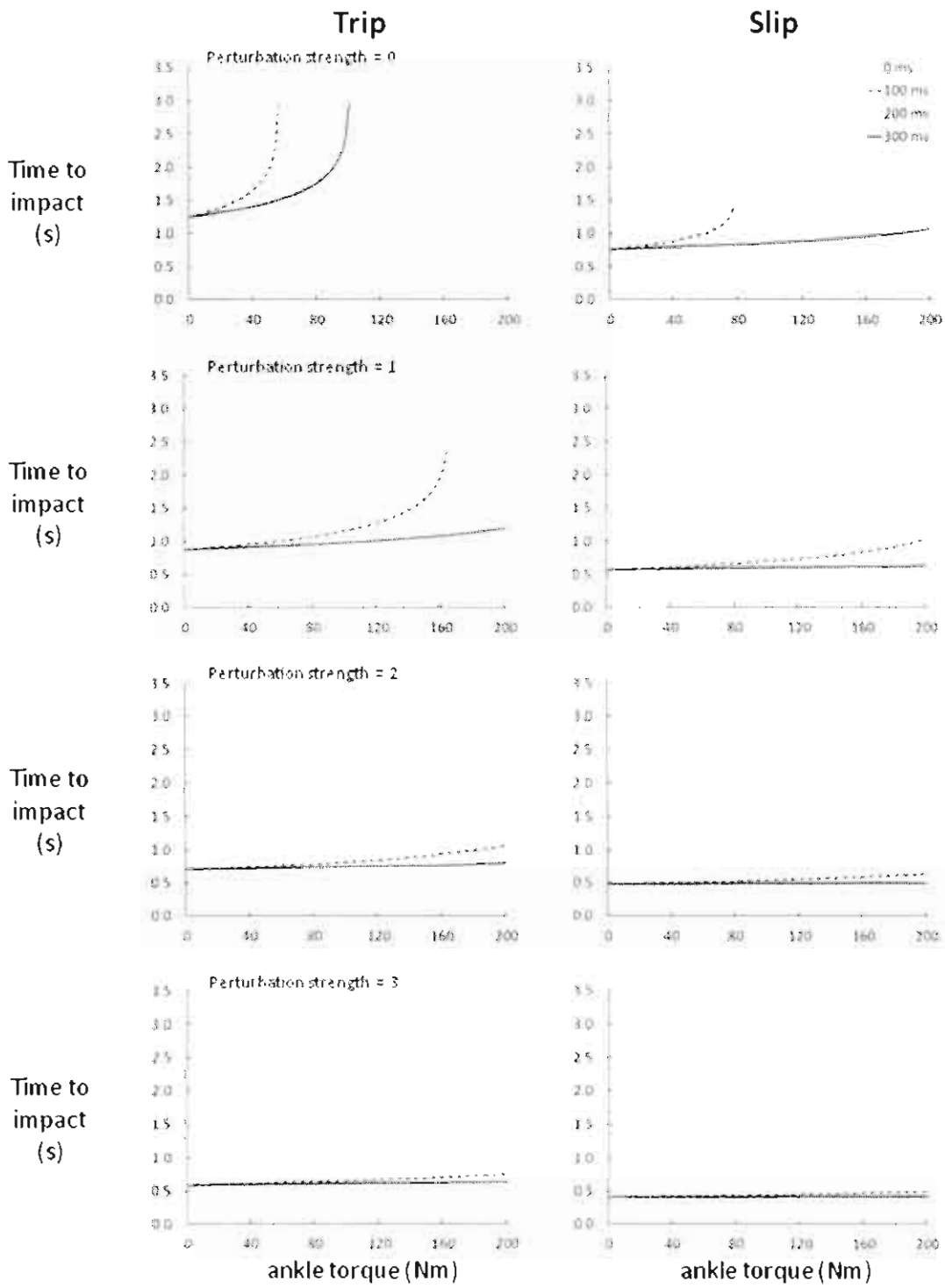
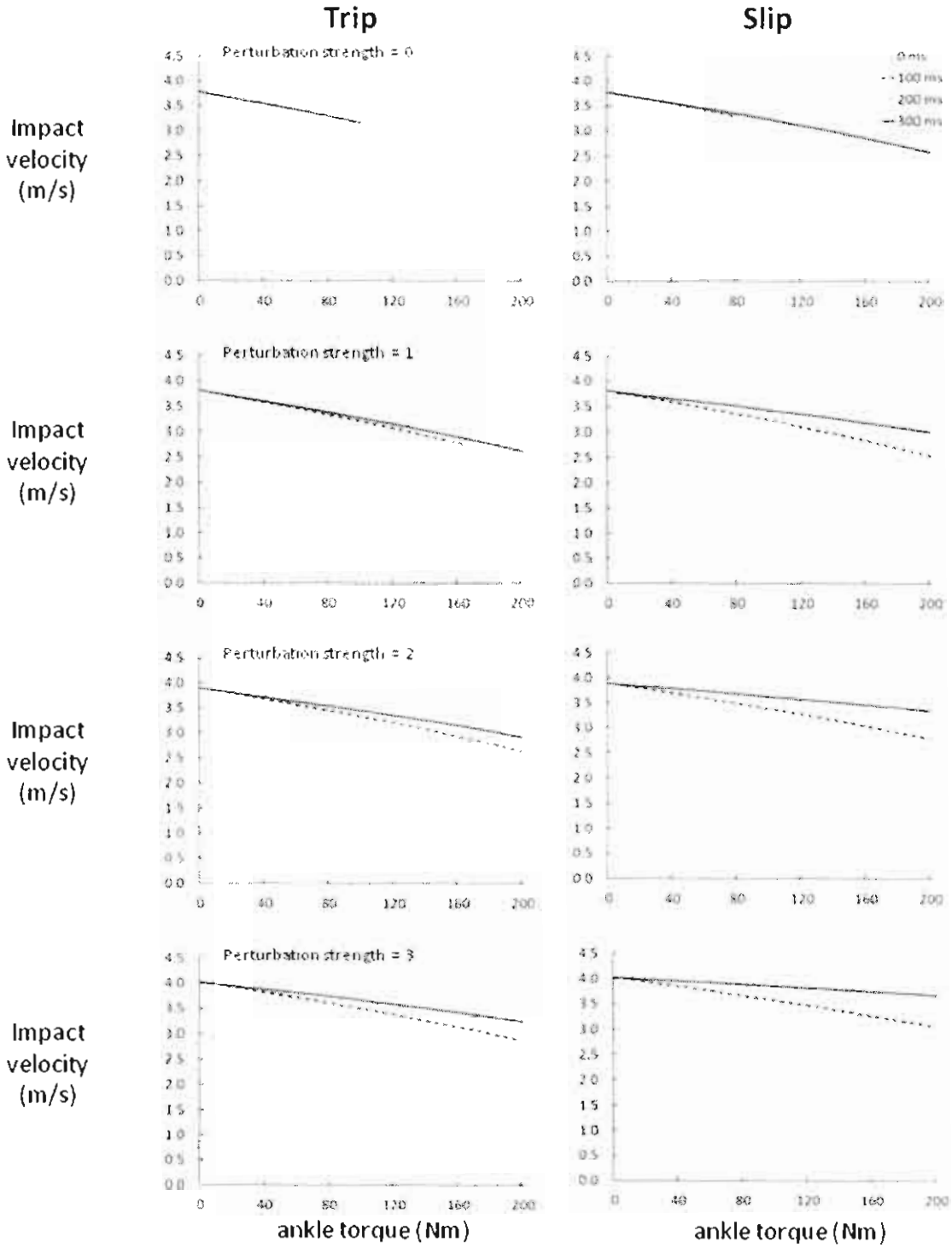


Figure 7-4: Sensitivity traces displaying the effect of ankle torque generation, reaction time on, and perturbation strength on impact velocity.



CHAPTER 8 THESIS SYNTHESIS AND CONCLUSION

The main purpose of my doctoral research was to improve our understanding of the cause and circumstances of falls and fall-related injuries. I focused specifically on the measurement and analysis of strategies for balance maintenance and safe landing, which have traditionally received relatively little attention, when compared to the balance recovery and impact stage of falls.

In my first two studies (Chapters 2 and 3), I developed a novel technique (the "Reach Utilization Test") to test whether differences existed between young and elderly groups in cautiousness or tendency to approach imbalance during a forward reaching task. This technique provides insight on how movement patterns are influenced by true motor capacities versus behavioural variables. My results demonstrate that elderly individuals are less likely than young to approach their maximum attainable reach (and imbalance) during a voluntary reaching task. My results also indicate that physical capacity (as measured by maximum attainable reach) and capacity utilization (as measured by voluntary reach) are independent determinants of reaching behaviour among the elderly. This leads us to hypothesize that each may be an independent predictor of mobility and risk for falls.

In my third study (Chapter 4), I developed a novel experimental paradigm to elicit realistic falls in a laboratory environment. The goal of this study was to determine whether unexpected sideways falls in young adults elicit a common sequence of protective responses that may protect against hip fracture. My results demonstrate that sideways falls in young adults commonly produce direct impact to the hip region, and

that the severity of hip impact is reduced by initial impact to the hand and by stepping, both of which were common. Results from this study help to explain why, in contrast to the elderly, fall-related hip fractures are relatively rare in young adults—even in sports such as soccer and basketball, where unpadded sideways falls are relatively common.

In my fourth study (Chapter 5), I tested the hypothesis that the ability of young women to avoid hip impact during a sideways fall (by rotating forward or backward during descent) is influenced by the time, relative to the onset of the fall, when this response is initiated. I found that forward and backward rotation were effective for avoiding hip impact, but only if the intent to rotate was realized within a short time window (200 ms) after fall initiation. This study provides important evidence of the strict time demands that govern safe landing responses.

In my fifth study (Chapter 6), I interviewed the young individuals immediately after they participated in the "balance competition" (Chapter 4), to test whether they were able to accurately recall the characteristics of their fall. My results indicate that, while individuals were able to correctly recall the direction of their fall, the task of recalling balance recovery attempts and impact configurations was prone to error. Furthermore, I found no association between participants' confidence in recalling their fall, and their actual recall accuracy. These results cast doubt on the accuracy of self-reported falls data, which are relied upon to guide intervention efforts.

Finally, in my final study (Chapter 7), I developed mathematical models to examine the theoretical influence of fall type (slip versus trip) on measures of impact severity. My

results suggest that falls from slips create shorter fall durations and higher impact velocities than trips, which may translate into higher risk for injury.

8.1 Future Directions

This thesis improves our understanding of behavioural and neuromuscular factors affecting balance maintenance and safe landing responses during falls. It introduces tools such as the "Reach Utilization Test" (Chapters 2 and 3), which may have a valuable role as screening or assessment tool in clinical environments. Future studies are required to evaluate the clinical utility of such measures for predicting risk for falls and mobility impairments in elderly populations. This thesis also describes innovative protocols for eliciting realistic falls (Chapter 4) and investigating safe landing responses (Chapter 5) in the laboratory environment. An important goal for future studies is to determine how fall mechanics are affected by aging and specific neurological or musculoskeletal impairments. Are the common protective responses we observed preserved with aging, or replaced with alternative strategies? To address these questions, we will need to design experiments that are realistic enough to elicit natural falling responses, but safe enough for older adults to participate. This might be achieved through the use of hip protectors and additional protective gear, harnesses and elastic tethers to slow the rate of fall descent, or by having participants fall into a swimming pool or extremely soft mattress (although attention must also be directed to the possibility of injury, such as strains or sprains, during the balance recovery and descent phase of falling). Another approach is to develop more sophisticated mathematical models, capable of predicting the effect on fall severity of age and disease-related declines in muscle strength and response speed. However, the most promising approach probably involves the development of methods to

measure fall movements in elderly individuals in real life, as a fall actually occurs, either through the use of networks of digital video cameras in high risk environments (e.g. nursing homes), or through wearable sensors in high-risk individuals. These approaches can also be used to examine (in "real-time") how mobility, balance, and risk for falls is influenced by interventions such as strength and balance training, modification of assistive devices, medication changes, and elimination of environmental hazards. Moreover, it should allow us to focus on more cognitive impaired or physically frail populations, who are especially vulnerable to falls, but have been traditionally omitted from laboratory-based studies.

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