CLINICAL AND LABORATORY BALANCE ASSESSMENT IN THE ELDERLY

by

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A DISSERTATION

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DISSERTATION ABSTRACT

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Falls can have severe consequences for elderly adults. In 2000, nearly 10,300 people aged 65 years or older died as a result of falls, and 2.6 million individuals were treated for non-fatal fall-related injuries. In order to reduce fall incidences, it is important to identify possible causes of falls, such as muscle weakness and imbalance. In this study, we examined balance control in the elderly during task transitions while performing the Timed Up and Go test (TUG). The TUG is a commonly used clinical balance test that includes transition phases between three daily activity tasks: sit-to-stand, walking and turning.

Our findings suggested that elderly adults, especially fallers, have reduced balance control ability while making transitions during TUG. During sit-to-walk (STW), when compared to young adults, elderly adults demonstrated a smaller forward center of mass (COM) velocity, a smaller anterior-posterior (A-P) COM-Ankle angle, and a larger upward kinetic energy ratio at seat-off. Additionally, the medial-lateral COM control in elderly fallers was also perturbed due to their significant reduction in forward COM velocity. The reduced initial hip extensor moment and increased ankle plantarflexor moment in elderly fallers was associated with their reduced generation of horizontal momentum during STW. Smaller A-P COM-Ankle angles and taking more steps when

making a turn demonstrated a reduction in balance control ability in elderly adults. Our analyses suggest that balance control is an important factor contributing to longer STW and turning durations of TUG. Furthermore, lower extremity muscle strength at hip and knee joints demonstrated a stronger association with STW than turning duration.

To enhance the early detection of fall risk, we also assessed the ability of balance tests to predict future risk of falling in elderly adults. Our results indicated that biomechanical balance parameters measured during TUG were associated with future fall status. Among all biomechanical parameters investigated, frontal plane balance control parameters appear to be the most significant predictors for future falls.

This dissertation includes unpublished co-authored material.

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TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

Table Page

CHAPTER I

INTRODUCTION

Falls are a major public health problem affecting elderly adults. Approximately a third of elderly adults over the age of 65 fall annually [1, 2]. The risk of falls increases significantly with age as 32-42% of adults over the age of 75 fall at least once over a one-year period [3, 4]. According to The International Classification of Diseases (ICD 9), a fall is defined as ‘an unexpected event where a person falls to the ground from an upper level or the same level’ [5]. Falling can occur at any time or place with most falls taking place in the afternoon and at home [6]. The majority of falls happen while walking on level or uneven surfaces due to a trip or slip, misplaced step, or loss of balance [6]. The health consequences of falls range from mild to severe. Ten to twenty percent of falls cause serious injuries such as fractures or head traumas [7]. Wrist fractures are more common between ages 65 and 75, with hip fractures occurring more often after the age of 75 [8]. Fall-related injuries are associated with mortality and considerable morbidity [7].

In 2000, nearly 10,300 people aged 65 years or older died as a result of falls and 2.6 million elderly adults were treated for non-fatal fall-related injuries [9]. Forty-three percent of patients injured in falls were discharged to a nursing facility [7]. Almost one-third of elderly adults experiencing a fall injury needed assistance with activities of daily living [10]. The medical costs for fatal and non-fatal fall-related injuries are high [7, 11, 12]. Direct medical costs of fatal and non-fatal fall-related injuries for people aged 65 years and older totaled $19.2 billion in 2000 [9]. The above summaries indicate the high incidences and costs of falls among the elderly population. With increased life expectancy nowadays, it is clear that this problem deserves clinical attention.

To reduce fall incidences through the development of effective screening and rehabilitation programs, it is important to identify the causes of falls, especially the risk factors for falling that are modifiable by physical therapy or other health care providers. Studies have documented several extrinsic and intrinsic factors that increase the risk of falling. Extrinsic risk factors are those related to the environment such as hazardous environment or footwear [13-15]. Intrinsic risk factors are those related to the individual such as balance impairment, gait deficit, muscle weakness [16], impaired vision [17], impaired cognition, use of medication, depression, urinary incontinence, dizziness, and fall history [18-20]. Among the intrinsic factors, muscle weakness and poor balance control are modifiable for falls in the elderly [21, 22]. Exercises comprising balance and strength training have been shown to effectively reduce fall incidents [23]. Therefore, it is important to examine balance control in the elderly during activities of daily living and its relation to falls. The following literature review will first discuss general age-related changes, following by discussions of commonly used clinical and laboratory balance assessments and their relation to falls.

Age-related Changes

The major age-related change in the musculoskeletal system is significant reductions in muscle mass and strength. A 12-year longitudinal study reported that isokinetic muscle strength declined at a rate ranging from 1.4 to 2.5% per year [24]. Furthermore, compared to young adults, older people in their seventies and eighties scored, on average, about 20-40% lower on isometric strength tests [25]. Age-related loss of muscle tissues is attributed to reduced numbers of type I and II fibers [26, 27]. Also, the architecture of the human gluteus medius muscle has been shown to be significantly altered in elderly individuals compared to young adults. Anatomic cross-sectional area, volume, fascicle length and pennation angle are all found to be smaller in elderly adults than young adults [28]. Additionally, denervation of muscle fibers, caused by death of motor neurons, could be one of the reasons for muscle strength reduction in the elderly [29]. Muscle strength reductions due to aging were reported to be greatest in the lower extremities [30]. Whole muscle analysis via cadaver or radiological observations has shown that thigh and leg muscles in older adults demonstrated significant size reductions over time compared to those in young adults [26, 31].

Muscle strength of the lower extremities is essential for balance maintenance and mobility. Reduced muscular strength is one of the physical changes that can significantly impact an older adult's functional ability [29, 32]. A systematic review and meta-analysis study indicated that lower extremity weakness is a clinically important and statistically significant risk factor for falls in the elderly [16]. It has been demonstrated that a loss in lower extremity strength is directly correlated with the ability of elderly adults to cross obstacles [33]. Improvement in lower extremity muscle strength could increase the speed of crossing stride and functional independence of older adults [34].

In the musculoskeletal system, not only is muscle mass reduced during aging, bone mineral density starts to decrease after reaching a peak during the first decade (19-29 years of age) of adult life [35, 36]. The rate of bone density loss in the hip and calcaneus increases with age in women who are 65 years or older [37]. Although lower bone density does not directly relate to impaired balance control in the elderly, bone density loss may put elderly adults at a higher risk of hip fracture. Hip fractures commonly result in permanent disability or death and are one of the most damaging fractures among elderly people [38].

Reduction in range of motion (ROM) is another age-related change in the musculoskeletal system. Declines in ROM are associated with decreases in mobility and activities of daily living (ADL) performance [39]. Due to decreased flexibility and reduced ROM, many older adults develop a stooped posture (lumbothoracic kyphosis with posterior pelvic tilt) [40]. The stooped posture forces a posterior shift of center of mass (COM). Backward COM shift may induce a backward fall and restrict posture adjustment ability during movement. A recent eight-week active-assisted stretching training study showed that the stretching exercises reduce ROM losses due to aging [41]. The stretching training program also improved performance of functional outcome measures such as gait velocity, Timed Up and Go test, and chair stand.

The changes in musculoskeletal system have effects on the neuromuscular system, which integrates sensory information and regulates movement control. Sensory information, which plays an important role in balance control, mainly is received from three systems: vision, vestibular, and somatosensory systems. The function of these three sensory systems declines with age [42-44]. Visual acuity starts to decline after age 50 [44]. Research has found that elderly adults depend more on vision to maintain balance than the young [45]. Moreover, poor vision has been documented as a risk factor for falling [44]. Age-related degeneration of vestibular function could lead to dizziness, and poor detection of angular and linear changes of the head during balance perturbations, which is also associated with a high risk of falling. Somatosensory information from the feet provides crucial information for posture adjustments when walking on uneven surfaces. Peripheral neuropathy associated with diabetes, which is common in elderly populations, is also a risk factor for falling. In addition to the decline of function in each sensory system, age-related changes in the neuromuscular system are also significant. The following paragraphs discuss the age-related changes in the neuromuscular system according to the three levels of neural systems for postural control: (1) Spinal stretch reflex; (2) Automatic control; (3) Voluntary control [40].

The amplitude of the spinal stretch reflex declines with normal aging. The latency of the Achilles and patella tendons' stretch reflex increases slightly with age [46, 47]. Decreased ability to modulate the stretch reflex may affect older adults' ability to maintain a static posture [48]. The ability to react to a sudden balance threat is also important. When subjects were perturbed by a movable platform, they activated the gastrocnemius muscle with onset latency at about 80 to 90 ms. This is longer than a spinal stretch reflex latency (45 ms). Nashner (1976) concluded that the control of reactive balance is more the result of automatic neural postural subsystems than a spinal reflex. [49].

To understand age-related changes associated with the automatic neural postural subsystems, Woollacott et al. (1986) examined the posture responses of healthy older and young adults using a movable platform. Compared to young adults, older adults showed either delayed onset of muscle activations or a reversed temporal distal to proximal muscle activation pattern. This demonstrated that the deterioration in postural control due to aging is related to the control of the automatic muscle response. The older adults showed significant increases in the amount of body sway under conditions in which both ankle joint inputs and visual inputs were distorted or absent than the young adults. The authors concluded that the higher sensory integrative center is also affected with age [50].

The highest level of the three level posture control hierarchy is voluntary control, including anticipatory postural abilities. Prior to a voluntary movement, postural adjustments are needed to stabilize the body. Inglin and Woollacott (1988) recruited 15 young and 15 older adults to test the effects of age on the anticipatory activation of postural muscles prior to arm movements. The subjects were asked to push or pull on a handle in response to a visual stimulus. The investigators found that the older subjects had longer onset latencies of postural muscles as well as longer reaction times for the arm muscles [51]. Bleuse (2006) also studied anticipatory postural adjustments in elderly subjects using electromyography and force plate measurements. Postural muscles were monitored when the subject raised one arm under three paces: self-selected slow, medium, and maximum. The researchers found that the elderly adopted various muscle strategies (more hip strategy) in order to perform the same movement with less stability [52]. In summary, older adults have difficulty making proper, efficient reactive and anticipatory responses with their neuromuscular system when facing an external (moving force platform perturbation) or internal posture threat (voluntary arm movement).

Other age-related changes in the neuromuscular system are found during gait. Kinematics analysis done by Winter (1990) during level ground walking found significant differences between elderly and young subjects in the following parameters: a shorter step length, an increased double-support stance period, a decreased push-off power, and a more flat-footed landing [53]. These differences indicate adaptation by the elderly toward a safer, more stable gait pattern.

While kinematic analysis provides information regarding temporal-distance characteristics, kinetic analysis aids in understanding forces and muscle responses during locomotion. DeVita and Hortobagyi (2000) reported that aging causes a redistribution of joint torques and powers in the lower extremities [54]. The older adults were reported to generate greater hip extensor moments and less knee extensor and ankle plantar flexor moments than the young adults when walking at the same speed.

Clinical Balance Assessments

Several clinical tests are currently used to evaluate age-related changes in balance control and fall risks. Examples of these clinical tests include: the Berg Balance Scale (BBS), the Timed Get-up and Go test (TUG), and the Fullerton Advanced Balance scale (FAB). Each clinical test will be discussed separately in the following paragraphs.

Berg Balance Scale

BBS is a performance-orientated measurement. Fourteen functional tasks are included in BBS [55]. Each item is scored on a scale of zero to four. BBS is designed to challenge subjects to maintain balance with a decreased base of support (from sitting to one-leg standing). Alternate stepping, 360 degree turning, and reaching are also measured. The 14 tasks within the BBS measure both steady state and anticipatory balance abilities within different skills, but do not measure reactive balance control.

Berg et. al. (1992) tested three groups of subjects (stroke patients in acute hospital, older adults from a home for the elderly and the community) with BBS. The authors concluded that the BBS score was a significant predictor of falls. However, the subjects included in the study demonstrated lower balance ability (BBS mean for the three groups are 31, 47, 38) and may not be representative of the general population [56]. Bogle, Thorbahn and Newton (1996) examined the ability of BBS to predict an elderly person's risk of falling [57]. In the 66 subjects analyzed, a total of 48 subjects scored 45 and above. The sensitivity for BBS at a six-month follow-up was 53% (8/15) and the specificity was 92% (36/39). Shumway-Cook (1997) reported that BBS was an excellent clinical balance test in predicting retrospective falls [58]. With a cut-off score at 49, BBS correctly classified 77% of fallers (sensitivity) and 86% of non-fallers (specificity). The average BBS scores for the fallers and non-fallers in this study were 52.6 and 39.6, respectively.

The above studies have shown that BBS score can distinguish fallers from non-fallers [56-58]. However, negative results on the ability of BBS to identify fallers were also reported by several studies. In a six month prospective study, BBS, combined with other clinical measures, showed poor ability to identify community-dwelling fallers [59]. Only 12% of fallers and 95% of non-fallers were correctly predicted. Additionally, no differences were found in BBS scores between fallers and non-fallers. The average BBS for fallers and non-fallers in this study were 53.4 and 53.9, respectively. Sensitivities of 25% and 45% for one fall and multiple falls detections, respectively, were reported by Muir (2008) [60]. The results suggested that BBS with a cut-off score at 45 was inadequate to identify most people at risk of falling. Both studies involved prospective design, which is the strength of the studies. However, the high BBS average score found in this particular subject group could be a limitation when applying the results to a population with a lower score.

In summary, BBS shows lower sensitivity and high specificity scores. Previous findings suggested that BBS is better at identifying individuals in the non-faller category but is not suitable for identifying people at high risk for falling. A limitation of the BBS is its poor prospective predictability of fallers, especially when applied to high scoring fallers. In addition, the subjects' balance characteristics have great influence on the results; thus caution is required when interpreting the results from different studies.

Timed Up and Go Test

The Timed Up and Go test (TUG) was first developed by Mathias in 1986 and it measures the time required for a person to stand up from a chair, walk 3 meters, turn around, return to the chair and sit down [61]. The TUG measures several basic essential functional tasks and their transition phases. Overall, TUG scores increase with age and use of assistive devices [62]. The TUG has been able to identify individuals who have cognitive impairment [63] and a higher risk of falling [64]. The time taken to complete the TUG moderately correlates to gait speed ($r = -0.61$), the Berg Balance Scale ($r = -0.81$), and the Barthel Index ($r = 0.78$) [65].

In addition, the TUG is correlated with several other balance measures. Studies have also suggested that TUG is sensitive enough to identify older adults at risk for falls [59, 66]. In a study directed by Shumway-Cook (2000), fifteen older adults with no history of falling and fifteen fallers with two or more falls in the past 6 months participated. The cut-off score was placed at 14 seconds by discriminate analysis. A sensitivity of 87% and specificity of 87% were reported. Another study by Bischoff

(2003) included a larger sample (413 community-dwelling and 78 institutionalized older adults). However, only female subjects were included in this study. The Receiver Operating Characteristics (ROC) analysis was performed. The TUG had a high diagnostic validity of discrimination of community-dwelling and institutionalized status indicated by an area under the curve of 0.969.

In summary, the TUG is a validated screening tool to identify fallers. However, similar to other clinical balance scales, it is not without its limitations. Although the TUG has clinical utility as a fall risk screening tool, it cannot provide detailed information regarding impairments in physiological domains that contribute to the risk of falling. It provides little in the way of information about how to target intervention strategies.

Fullerton Advanced Balance Scale

The Fullerton Advanced Balance Scale (FAB) was developed to identify more subtle changes in the multiple dimensions of balance (e.g., Motor, sensory, musculoskeletal) among independently functioning older adults [67]. This scale can be used to identify functional limitations associated with impairments in the visual, somatosensory, and vestibular sensory systems as well as the neuromuscular and musculoskeletal systems. It includes 10 individual testing items such as tandem walking, one leg standing, and two-footed jumping with a maximum score of 40. FAB has proven high test-retest reliability (0.96) as well as intra and inter-rater-rater reliability (0.91, 0.95, respectively). FAB was also found to be correlated (0.75) with the Berg Balance Scale [67]. Furthermore, one study has retrospectively investigated the fall predictive validity of the scale in the elderly [68]. A score of 25 or lower out of 40 was associated with a heightened risk of falling. The sensitivity of the scale was shown to be 74.6% while the specificity reported was 52.6% [68].

In summary, FAB has the advantage of measuring subtle changes in the multiple dimensions of balance, especially in functional independent elderly adults. However, more prospective studies are needed to test the ability of FAB to predict fall status.

Biomechanical Balance Parameters

Although clinical balance tests can indicate balance control abilities, they are inadequate for assessing underlying mechanisms of mobility impairment. In contrast, biomechanical analyses are able to provide insight into movement and postural control. The most common parameter used to describe balance control is the motion of the whole body center of mass (COM) and its relation to the base of support [27, 57, 59]. The inverted pendulum model has been developed to describe postural control in quiet standing [57, 59]. In this model, the center of pressure (COP) oscillates on either side of the COM with COP movement always exceeding the COM movement [57]. The COP and COM separation has been shown to relate directly to the horizontal acceleration of the COM. Thus, it can be considered an error signal that the balance control system is sensing [57]. Moreover, the instantaneous velocity of the COM and location of the COM with respect to the base of support were identified as important factors in maintaining balance during standing [59]. In addition, the horizontal distance between the COM and COP of the stance foot is related to the external joint movements of the supporting limb. A larger horizontal COM-COP distance will result in larger external joint movements in the supporting limb. Therefore, control of the COM motion and its relation to the COP of the stance foot are important to demonstrate the resultant kinetic demands at the joints of the supporting limb when stepping over obstacles [60].

Studies have tested the use of COM parameters in balance assessment and fall risk prediction. Elderly people who are at a higher risk of falling have larger medio-lateral motion of the COM than young adults during gait [69-72]. Further research suggested that the motion of the COM and its relationship to the center of pressure (COP) might be a better parameter to determine dynamic balance control [69, 73, 74]. A recent parameter known as the COM-COP inclination angle provides information on the interaction between the COM and COP and also accounts for an individual's height [74]. It was shown that the instantaneous inclination angles formed by the COM and COP in the frontal plane were sufficiently sensitive to quantify balance during walking and to identify high risk fallers. When the COP is not available, the COM-Ankle inclination angle during single stance phase was demonstrated to be an alternative assessment for clinical populations [75]. Elderly people with balance impairment demonstrated a greater peak medial and smaller peak anterior inclination angles than healthy elderly adults during gait. Many other biomechanics studies have investigated other functional tasks using laboratory techniques such as sit to stand (STS), gait initiation, gait termination, steady state level ground gait, obstacle crossing and turning [76-81]. This research has provided useful insights regarding each individual task. However, in real life, human locomotion occurs in a continuous series. The investigations of a transition between the two tasks or the descriptions of a series of motions are, therefore, important and needed. For example, studies have demonstrated that the biomechanical characteristics of sit-to-walk (STW) are different from those of sit to stand or gait initiation [82-84]. STW was reported to impose greater challenges to balance control than STS alone due to its requirement of a greater horizontal momentum for gait initiation and a simultaneously narrowing of the base of support [85].

Few biomechanical studies have investigated balance control during STW. When compared to young adults, elderly adults were found to generate less horizontal COM momentum at seat-off in order to maintain a more stable upright posture before walking [86, 87]. Elderly adults who are at risk of falling also showed movement hesitancy during STW with a significant decrease in COM forward velocity after seat-off [88]. It has been further suggested that elderly adults with fear of falling demonstrated a disproportionately increased sideways velocity compared to a reduced forward velocity during STW [89].

Turning during walking is another common daily activity that requires successful transitions. Falling while turning carries a high risk of hip fracture [90]. Turning difficulty has been reported in the elderly population [91, 92], and elderly adults and the elderly adults at risk of falling group took longer to complete the turning component of the TUG [92]. Research suggests that elderly fallers demonstrated unsteadiness, staggering, or stopping during turning compared to elderly non-fallers [93]. Elderly adults and young adults use different turning strategies when making 45^{o} and 90^{o} turns while walking. When compared to young adults, elderly adults used a turning strategy that has a greater biomechanical cost. Choosing the turn strategy that is more biomechanically demanding may contribute to the higher risk of falling in the elderly [94]. Documenting turning strategies and quantitatively investigating COM control during the turning phase of the TUG can shed light on the difficulty the elderly population, especially elderly fallers, encounters during this common daily activity and allow earlier identification of individuals at risk of falling and the crafting of more effective interventions.

Connections between Clinical and Laboratory Balance Assessments

One goal of conducting laboratory research is to apply the results to enhance clinical assessments. Although these biomechanical analyses provide precise information about balance control and fall risk prediction, they have not evolved to the point where the equipment is easily portable or clinically practical. Therefore, information linking biomechanical balance measures with clinical balance tests is needed. With this piece of information, clinicians can understand more about underlying problems of abnormal clinical test results.

Few studies have investigated the correlations between outcomes from clinical balance tests and laboratory biomechanical assessments. Lichtenstein et al. (1990) has reported that knee range of motion and stride length correlated with the performance on the Tinetti mobility index. Additionally, force plate measures (the range of center of pressure trajectory) under quiet double and single leg stance are associated with the results of the mobility index [95]. Another correlation study has reported high correlation between accelerometer measures during quiet standing and two clinical balance tests: BBS and TUG [96]. Furthermore, moderate correlation was found between BBS and standing postural sway measures in the elderly [56].

The above research has shown that biomechanical measures during quiet stance are associated with some clinical balance tests [95, 96]. However, most falls in elderly adults occur under dynamic locomotion [6]. Furthermore, different balance control studies during quiet standing and gait have demonstrated low correlations between clinical stance and locomotor tasks [97]. Studies exploring links between balance performance during dynamic situations and clinical balance exams are still lacking. Information linking biomechanics and clinical balance assessments will provide clinicians more insight about underlying problems of abnormal clinical test results. Yet simply correlating a clinical measure and a biomechanical measure is not enough since this information does not necessarily provide insight for fall prediction. In order to enhance early fall risk detection, the ability of a balance test to predict risk of falling in elderly individuals must be assessed.

Some researchers have studied the use of laboratory measurements in fall prediction. Hillard et al. (2008) investigated the ability of lateral balance factors to predict future falls in community-living elderly adults [98]. The findings suggest that frontal plane balance recovery performance and lateral balance stability are significant predictors of prospective falls. Another study conducted by Maki et al. (1994) reported that lateral spontaneous-sway amplitude under blindfolded conditions was the single best predictor of future falling risk [99]. Hausdorff et al. (2001) demonstrated that stride time variability during gait was able to predict falls in a one-year prospective study [1]. Studies have also examined the ability of both laboratory measures and clinical tests to predict elderly fallers and non-fallers. The results have shown that while the laboratory parameters were correlated with future falls some clinical balance tests were not able to predict fallers [59, 100].

Overall Goal and Specific Aims

Prior summaries have shown that some biomechanical balance parameters during static situations are correlated with clinical balance evaluations scores. Moreover, combining biomechanical analyses and clinical exams may be able to provide more detailed information about balance performance in elderly fallers [83, 95, 96, 101, 102]. In the following proposed studies, we hope to provide knowledge about the relationship between clinical balance tests and biomechanical balance measures that examine dynamic balance control. The results of this study will provide clinicians with additional information necessary to assess underlying impairments within a simple balance test, enhancing early fall risk detection, and helping to create specific intervention plans.

Within the context of this overall objective, three specific aims were proposed in this project:

1. Investigation of COM motion during task transitions. Previous research related to dynamic balance control has provided useful insights regarding each individual task. However, in real life, human locomotion occurs within a continuous series of tasks. Studies have demonstrated that the biomechanical characteristics of STW are different from sit to stand or gait initiation [82-84]. Investigations of a transition between the two tasks or the descriptions of a series of motions are, therefore, important and needed. The TUG is a clinically commonly used mobility test, including transition phases between three daily activity tasks: sit to stand, walking and turning. In addition, a summary of prior research about TUG shows that even if TUG has clinical utility as a fall risk screening tool, it cannot provide detailed information regarding the impairments

in physiological domains that contribute to the risk of falling. Therefore, the purpose of this study is to provide kinematics and kinetics information of TUG, especially the transition phases such as STW and turning. The results of the analysis will be used to examine the differences between three subject groups: young adults, elderly adults, and elderly adults with fall histories. The hypothesis in this study is that elderly adults with fall histories will demonstrate reduced balance control ability when compared to the other two groups. In this study, the balance control ability will be quantified by COM position and its relation to the base of support.

2. <u>Examine correlations between increased TUG components' times and specific functional deficits.</u> Previous studies have shown that elderly fallers need more time in the sit-to-stand and turning components in the modified TUG when compared to young subjects [92]. However, no further data have been collected to demonstrate what factors cause differences between the two groups. Therefore, this study aims to address this issue by examining underlying impairments (muscle strength and balance control ability), which may cause elderly adults to have difficulty in the STW and turning components. We hypothesize that both STW and turn duration will correlate with muscle strength and balance—the STW time will be associated more with muscle strength and the turn time associated more with balance.

3. <u>Assess the feasibility of using dynamic laboratory balance measures to prospectively predict falls risk in the elderly.</u> Most previous biomechanical studies only examined balance control during static postural sway [59, 99, 100,

103] or as part of clinical balance scales, when focusing on falls prediction [57-59]. Few studies have investigated the ability of laboratory parameters, which analyze dynamic balance, to predict elderly fallers [1, 98]. Considering that biomechanical analysis can provide detailed information about balance control and detect subtle changes in body motions, more research is needed to explore the ability of dynamic biomechanical markers to predict falls. Therefore, the purpose of this study is to assess the feasibility of using biomechanical measures of gait imbalance (COM-Ankle angles in the frontal and sagittal planes) to prospectively predict a fall (or falls) in community-dwelling elderly adults and compare their prediction ability to the clinical tests (TUG, BBS, FAB). Hypothetically, the biomechanical parameters would be able to predict future falls in community-dwelling elderly adults. Furthermore, we expected that combining the clinical balance and biomechanical balance measures would demonstrate a better prediction of prospective fall incidents than clinical balance measures alone.

Flow of Dissertation

This dissertation is structured in a journal format. The studies described in Chapters II-V include co-authored materials. Following the general introduction and literature review in Chapter I, Chapters II through V are individual manuscripts prepared for submission to peer-reviewed scientific journals. Chapters II-III described balance control during STW and turning transitions in the TUG (first specific aim). Chapter IV discusses correlations between increased TUG components' times and specific functional deficits (second specific aim). Chapter V investigates the feasibility of using dynamic laboratory balance measures to prospectively predict fall risk in the elderly (third specific aim). Finally, conclusions are provided in Chapter VI, which also includes the dissertation's limitations and suggestions for future research.

CHAPTER II

BIOMECHANICAL ANALYSES OF TASKS AND TRANSITIONS DURING TIMED UP AND GO TEST

This chapter was developed by Dr. Li-Shan Chou and me. Dr. Chou contributed substantially to this work by participating in the development of methodologies and providing critiques and editing advice. I was the primary contributor to the development of the protocol, data collection, data analysis and did all the writing.

Introduction

Most falls in elderly adults occur during daily activities [2, 6]. Many of these activities require successful ambulatory transitions between two different postures, tasks, or directions (e.g., sitting, standing, walking, and turning). However, there is a lack of information about how body movement is controlled during the transition between tasks.

The routinely used clinical fall risk assessment, the Timed Up and Go test (TUG), includes different daily activities: (1) sit-to-stand, (2) walking, (3) turning, and their transitions [59, 61, 66]. Studies have demonstrated that the amount of time required to complete the TUG correlates with the Berg balance scale, gait speed, and the Barthel index [65]. Additionally, elderly adults with a higher risk of falling take longer to complete the TUG [92]. While timing the TUG can serve as an initial screening tool to detect fallers [59], biomechanical analyses of activities included in the TUG and their transitions would allow a better examination of the underlying mechanisms associated with functional declines in older adults, provide clinicians an insightful interpretation of the TUG results and yield focused knowledge for intervention development.

Measurement of instantaneous positioning of the COM with respect to the center of pressure (COP) during gait could detect elderly individuals with impaired balance control [74, 104, 105]. Elderly individuals with impaired balance demonstrated a smaller sagittal plane angle and larger frontal plane angle during gait. When the COP is not available, the COM-Ankle inclination angle during single stance phase has been demonstrated to be an alternative assessment to clinical populations [75]. Elderly adults with balance impairment demonstrated a greater peak medial and smaller peak anterior COM-Ankle angle than healthy elderly adults during walking. Although COM movement has been reported in healthy elderly adults, data on how COM is controlled in relation to the supporting foot in elderly adults with fall histories during STW are limited [89].

STW imposes greater challenges on balance control than STS alone due to its requirement of a greater horizontal momentum for gait initiation and a simultaneously narrowing of the base of support [85]. Moreover, STW is not a sequential arrangement of two individual tasks, but requires a smooth transition from STS to gait initiation at the seat-off instant. However, such smooth transitions are not observed in elderly adults [86-88]. Elderly adults generated less horizontal center of mass (COM) momentum at seat-off in order to maintain a more stable upright posture before walking when compared to young adults [86, 87]. Many factors could contribute to this age-related change, including declines in muscle strength and joint motion, poor balance control, or fear of falling [106-108]. Elderly adults who are at risk of falling also showed a movement hesitancy during STW, with a significant decrease in COM forward velocity after seat-off [88]. Such strategies to reduce COM forward velocity could change the distribution of COM momentum to other movement directions, such as the medio-lateral direction, and, therefore, perturb the momentum control. It has been further suggested that elderly adults with a fear of falling demonstrated a disproportionately increased sideways velocity, as compared to a reduced forward velocity during STW [89]. This indicates that declines in the ability to properly regulate COM momentum in the sagittal and frontal planes could be contributing factors to imbalance during STW. Examining the COM momentum and its distribution across different movement directions during STW would provide further insights on how balance control is inter-related in different motion planes.

Turning during walking is an essential functional activity. Difficulty in turning while walking has been reported in the elderly population [91, 92], and moreover, falls occurring while turning carry a higher risk of hip fracture [90]. Elderly adults at risk for falling were reported to take a longer time to complete the turning component of the TUG. It has been suggested that elderly fallers demonstrate unsteadiness, staggering, or stopping during turning as compared to elderly non-fallers [93]. Documenting turning strategies and quantitatively examining COM control during the turning phase of the TUG may better reveal the biomechanical challenges imposed on elderly adults, especially elderly fallers, and allow earlier detection of individuals at a higher risk of falling and an effective crafting of preventive interventions.

The first objective in this study was to examine the differences in COM control and its relationship to the BOS during the transitional phases of the TUG (STW and turning) among three subject groups: young adults, elderly adults, and elderly adults with fall histories. As age-related hesitancy has been shown in STW, we hypothesized that when compared to young adults, elderly adults would demonstrated significant change in sagittal plane COM control, such as a smaller forward COM velocity, a more upward COM momentum distribution and a smaller anterior-posterior COM-Ankle angle. However, when compared to healthy elderly adults, elderly fallers would also demonstrate an excessive medial-lateral COM momentum that could perturb balance control in the frontal plane. The second objective was to examine the differences in turning strategies among these three groups. We hypothesized that different turning strategies would be identified among the three groups, with more conservative strategies used by healthy elderly and fallers.

Methods

Subjects

Fifteen healthy young adults (YA; 8 women/7 men; mean age: 26.0±3.4 years, mean height: 167.6±6.4 cm, mean mass: 63.1±9.7 kg), 15 healthy elderly adults (EA; 9 women/6 men; mean age: 76.2±4.2 years, mean height: 163.9±10.5 cm, mean mass: 72.0±16.4 kg), and 15 elderly adults with a fall history (EF; 11 women/4 men; mean age: 77.7±7.7 years, mean height: 162.5±9.5 cm, mean mass: 77.2±23.2 kg) over the age of 70 were recruited from the community. Prior to the study, a power analysis was performed using the horizontal COM velocity collected from 4 subjects in each group. The analysis revealed that eleven subjects per group were required to achieve a power of 0.95 with an alpha level of 0.05.

Inclusion criteria for healthy young and elderly participants were individuals who 1) could walk without the use of an assistive device; 2) had no history of neurological or musculoskeletal deficits that might contribute to gait instability or falls, such as amputation, cerebral vascular accident, significant head trauma or Parkinson's disease; 3) had no uncorrectable visual impairment, vestibular dysfunction, or dementia. The EF in this study were elderly individuals who had fallen twice or more in the year previous to the testing date. The definition of fall in this study is based on the definition by The International Classification of Diseases: "fall is an unexpected event where a person falls to the ground from an upper level or the same level." [5] Furthermore, only falls that occurred during activities of daily living were included, so that falls due to syncope or major intrinsic events, such as stroke, were excluded. An average of 3.1 (±1.0) falls was reported by the EF. In order to target fallers with balance impairments, a Fullerton Advanced Balance (FAB) scale score lower than 30 was required for the EF [68]. The FAB scale is a performance-based measure specifically designed for use with independently functioning elderly adults, with a reported good validity and reliability [67]. The FAB scores were 33.6 (±2.7) and 21.4 (±8.4) for the EA and EF, respectively. Prior to testing, all participants agreed to the experimental procedure approved by the Institutional Review Board, and signed consent forms were obtained.

Experimental Protocol

Participants performed the Timed Up and Go test [61] while barefoot. They were asked to stand up from the bench, walk 3 meters, turn around, return to the bench and sit down. Including the practice trial, a total of four trials were performed. Rest periods between each trial were provided to the participants as needed. The following consistent instructions were provided to all subjects: "Please complete the whole task at your comfortable speed, and we will time you." A height-adjustable bench for sitting was set at each participant's knee height.

Experimental Instrument

Twenty-nine markers were placed on selected bony landmarks of the subject [69]. Whole body motion data were captured with a 10-camera motion system (Motion Analysis Corp., Santa Rosa, CA). Marker trajectory data were sampled at a rate of 60 Hz and then smoothed using a low pass fourth-order Butterworth filter with a cutoff frequency of 8 Hz. While seated, participants placed both feet on a force plate (AMTI, Watertown, MA) to allow for the detection of seat-off [106]. Ground reaction forces were sampled at 960 Hz. Anthropometric reference data for both sexes were adapted from Dempster [109]. Whole body COM position was calculated as the weighted sum of a 13-segment model [69]. The 13 segments are: head and neck, trunk, pelvis, and right and left segment of upper arms, forearms, thighs, shanks, and feet.

Data Analysis

The overall time used to complete the TUG was recorded. In order to present the contribution of COM velocity in each direction, the linear COM kinetic energy was calculated as $\frac{1}{2} * m * v^2$ (*m* = body mass, *v* = velocity). Total COM kinetic energy was the sum of the kinetic energy in all three directions, $\frac{1}{2} * m * v_x^2 + \frac{1}{2} * m * v_y^2 + \frac{1}{2} * m * v_z^2$.

The COM kinetic energy in each direction was then normalized by the total COM kinetic energy, to yield the ratio of kinetic energy in three directions. COM-Ankle angles were calculated as the inclination angles of the line formed by the COM and lateral ankle malleolus marker in the sagittal plane for each frame [75].

Data from onset, seat-off, swing leg toe-off (swing-off), and stance leg toe-off (stance-off) during STW were extracted for analysis. Onset was identified as the instant of initial COM forward position change [106]. Seat-off was identified as the time of the peak vertical ground reaction force [84]. Swing-off and stance-off were identified as the time when the leading foot toe marker and trailing foot moved forward, respectively [106]. The duration of STW was calculated from onset to stance-off. Distances between the right and left lateral malleolus ankle markers in the frontal (step width) and sagittal plane (step length) were examined at stance-off (the first step during STW).

Turning duration was calculated from the start-of-turn to end-of-turn. The start-of-turn was defined as the toe-off immediately prior to the instant when the transverse plane pelvic rotation exceeded the range of rotation detected during walking. The end-of-turn was defined as the sagittal plane COM position after the turn reached the same position as the start-of-turn. Transverse plane COM trajectories (top view) during turning were examined. In addition, a parabolic turning curve was mathematically fitted using the COM positions at the start- and end-of-turn as well as a vertex (Figure 2.1). The vertex was defined as the most anterior position traversed by the COM and located at the mid-point in the medio-lateral direction between the COM positions at the start and end of the turn. The length ratio between the actual and parabolic turning curves was then calculated.

Individual turning strategies were determined based on the foot placements in the transverse plane. The number of steps needed to complete the turn was counted. The supporting limb that made the first direction change was defined as the "turning limb," and this stance period was termed as "pivoting." Ranges of the anterior-posterior (A-P)

COM-Ankle angles during pivoting and stances periods of the step prior to and after pivoting were extracted for analysis.

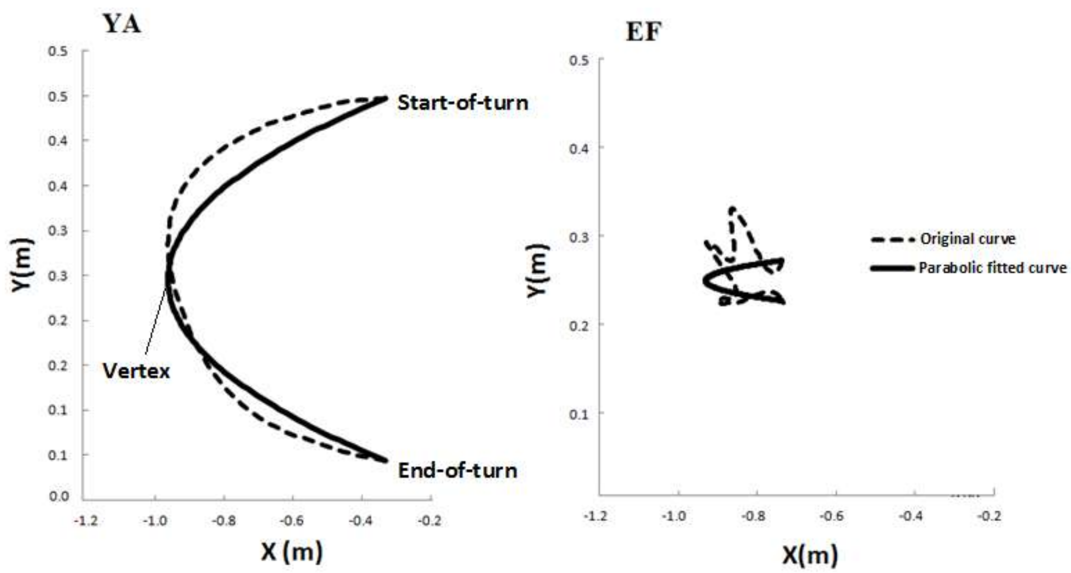

Figure 2.1. Ideal fit curve example for COM trajectory in transverse plane for one young adult and one elderly faller (solid line: ideal fit curve, dashed line: original data).

The differences between the three groups were assessed using multivariate analysis of variance (MANOVA). The dependent variables included in this study were: COM forward and upward velocities, sagittal plane COM-Ankle angle, total COM kinetic energy, and ratios of COM kinetic energy in three directions at seat-off, swing-off and stance-off. In order to control for the possible effect of speed, STW duration was included as a covariate in the analysis for COM-Ankle angles. Follow up analysis was performed, when the results of MANOVA were significant, using the Bonferroni adjustment. The paired t-test was used to examine the differences in velocities, step lengths, and COM-Ankle angles between gait and stance-off as well as between gait and turning. The significance level for all statistical tests was set at $\alpha = 0.05$. SPSS version 19.0 (IBM SPSS Inc., Chicago, IL) and was used for all statistical analyses.

Results

No significant differences were found between the groups in body mass and height ($p \geq 0.092$). No significant age differences were found between the two elderly groups ($p = 0.521$). EA had significant higher FAB score than EF ($p < 0.001$). EF took a significantly longer time to complete the entire TUG than either YA or EA (Table 2.1).

Sit-to-walk

Significant group main effects were detected in STW duration and step length at stance-off (Table 2.1). When compared to YA, STW duration was significantly longer for EF. At stance-off, YA took a significantly larger step than both EA and EF. Additionally, the EA had a larger step length at stance-off than EF. No significant differences were found between groups in the step width during STW.

Young adults demonstrated a greater COM forward velocity throughout the STW phase than both EA and EF (Figure 2.2-a). When compared to EA and EF, YA demonstrated significantly higher forward COM velocities at seat-off, swing-off and stance-off (Figure 2.3-a). EA demonstrated significantly greater forward COM velocities than EF at seat-off and stance-off (Figure 2.3-a). In the vertical direction, YA reached a greater peak velocity than EA and EF (Figure 2.2-b). Furthermore, significant group main effects were found for upward velocity at swing-off and stance-off. YA had a significantly greater upward velocity than EF (Figure 2.3-b). No significant group main effects were detected in upward velocity at seat-off. In the medial-lateral direction, YA

Table 2.1. Temporal distance parameters for STW, gait, and turning phase during the TUG for the three groups [Mean (SD)].

	YA	EA	EF	p-value	p-value (* ǂ #)
TUG duration (s)	7.93 (1.18)	10.18 (1.54)	14.78 (6.41)	< 0.001	0.359* < 0.001ǂ 0.007#
STW duration (s)	1.31 (0.18)	1.60 (0.34)	2.29 (1.45)	0.01	1.00* 0.010ǂ 0.102#
Turn duration (s)	1.58 (0.53)	2.33 (0.77)	3.36 (0.97)	< 0.001	0.048* < 0.001ǂ 0.003#
STW					
Velocity(stance-off)	1.16 (0.17)	0.92 (0.19)	0.65 (0.26)	< 0.001	0.009* < 0.001ǂ 0.003#
Step length (m)	0.59 (0.06)	0.52 (0.08)	0.42 (0.06)	< 0.001	0.045* < 0.001ǂ < 0.001#
Step width (m)	0.21 (0.08)	0.21 (0.07)	0.23 (0.07)	0.714	N/A
Gait					
Velocity(m/s)	1.18 (0.16)	0.95 (0.18)	0.71 (0.27)	< 0.001	0.999* < 0.001ǂ 0.007#
Step Length (m)	0.65 (0.05)	0.54 (0.09)	0.41 (0.12)	0.005	1.000* 0.006ǂ 0.031#
Step Width (m)	0.11 (0.03)	0.13 (0.03)	0.14 (0.06)	0.405	N/A
Turn					
Velocity start(m/s)	1.07 (0.29)	0.89 (0.22)	0.73 (0.30)	0.008	0.286* 0.006ǂ 0.323#
Velocity end(m/s)	1.01 (0.19)	0.85 (0.19)	0.70 (0.25)	0.002	0.199* 0.001ǂ 0.155#

*p**: differences between YA and EA, *p*ǂ differences between YA and EF, *p#* differences between EA and EF.

demonstrated a greater velocity at swing-off than EA and EF (0.114 ± 0.045, 0.077 ± 0.047, 0.072 ± 0.049 m/s respectively, $p = 0.039$).

The magnitude of anterior-posterior (A-P) COM-Ankle angle was significantly greater in YA and EA than EF at seat-off (Table 2.2). YA demonstrated a greater posterior COM-Ankle angle than EA at seat-off. At swing-off, no significant group differences were found in anterior COM-Ankle angle. At stance-off, YA and EA demonstrated a greater posterior COM-Ankle angle than EF. No significant difference was found in A-P COM-Ankle angle at stance-off between YA and EA.

Significant group main effects were found in total COM kinetic energy at seat-off, swing-off and stance-off (Table 2.3). At swing-off and stance-off, YA demonstrated a larger total COM kinetic energy than EA and EF. However, no significant differences were found between EA and EF. When compared to YA, EA and EF distributed their kinetic energy more in the upward than forward direction at seat-off (Table 2.3). At swing-off, significant group main effects were detected in medial-lateral COM kinetic energy ratio with the largest magnitude observed in EF. However, the post-hoc analysis revealed that only the difference between YA and EF approached to a significant level (p=0.067).

Walking

Significant group main effects were detected in the gait velocity and step length during the walking period (Table 2.1). EF walked significantly slower than YA or EA. No significant group differences were detected for the step width. EF demonstrated significantly smaller peak anterior and posterior COM-Ankle angles when compared to YA (Table 2.2).

Turning

YA completed the turning phase significantly faster than EA or EF (Table 2.1). Also, the turning duration was shorter in EA than EF (Table 2.1). The COM forward velocity at the beginning of the turn was significantly slower in EF than YA (p = 0.006). EF took more steps to complete a turn when compared to EA and YA (YA: 4.2 ± 1.1, EA: 5.3 ± 1.0, EF: 6.3 ± 1.0, p = 0.024, p < 0.001, respectively).

Five turning patterns were identified (Pattern A, B, C, D, E: Figure 2.4, 2.5, 2.6, 2.7, 2.8, respectively). Table 2.4 reported the distribution of each turning pattern among the participants.

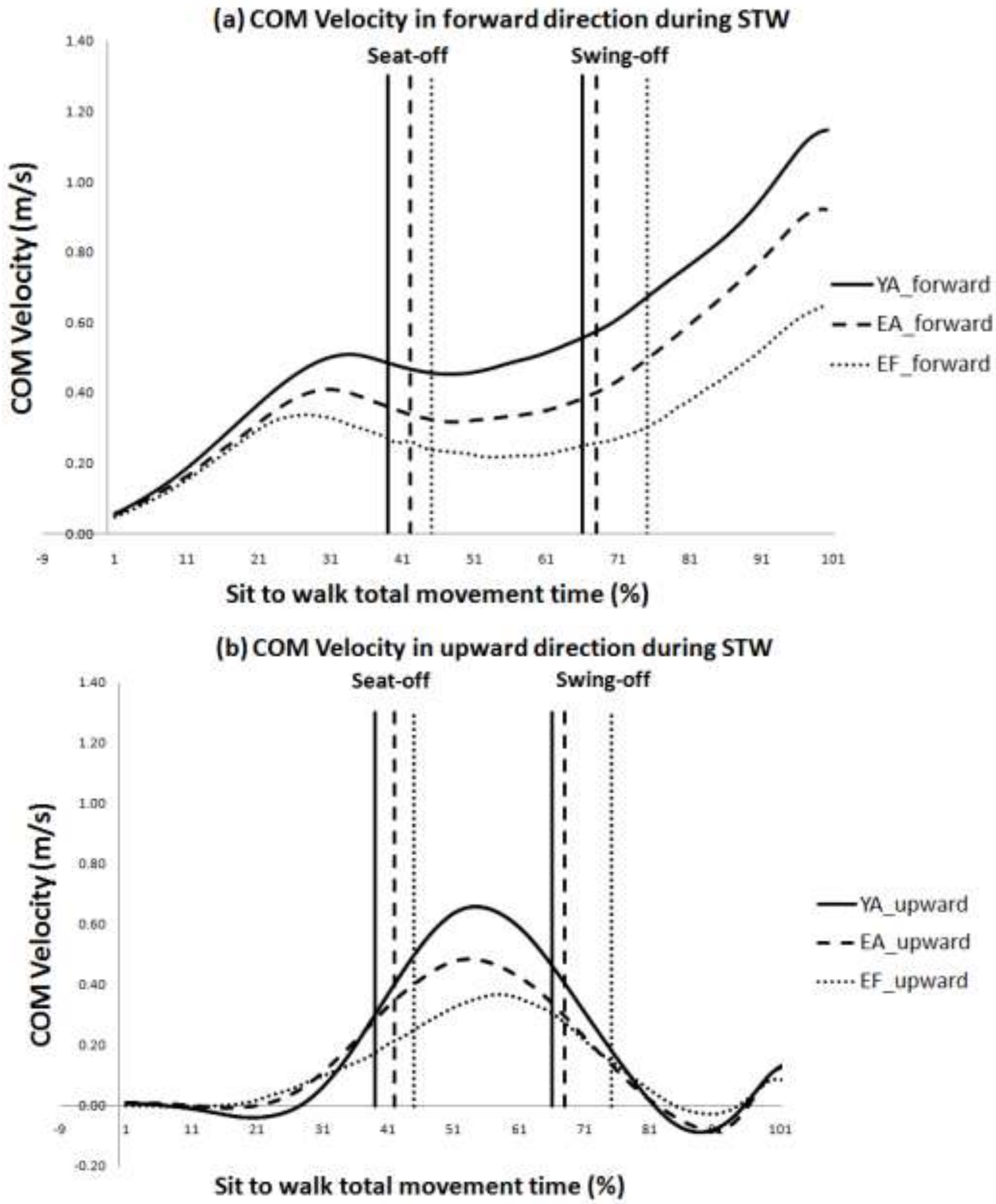

Figure 2.2. Profiles of COM velocities during STW (a) Profiles of COM forward velocities. (b) Profiles of COM upward velocities.

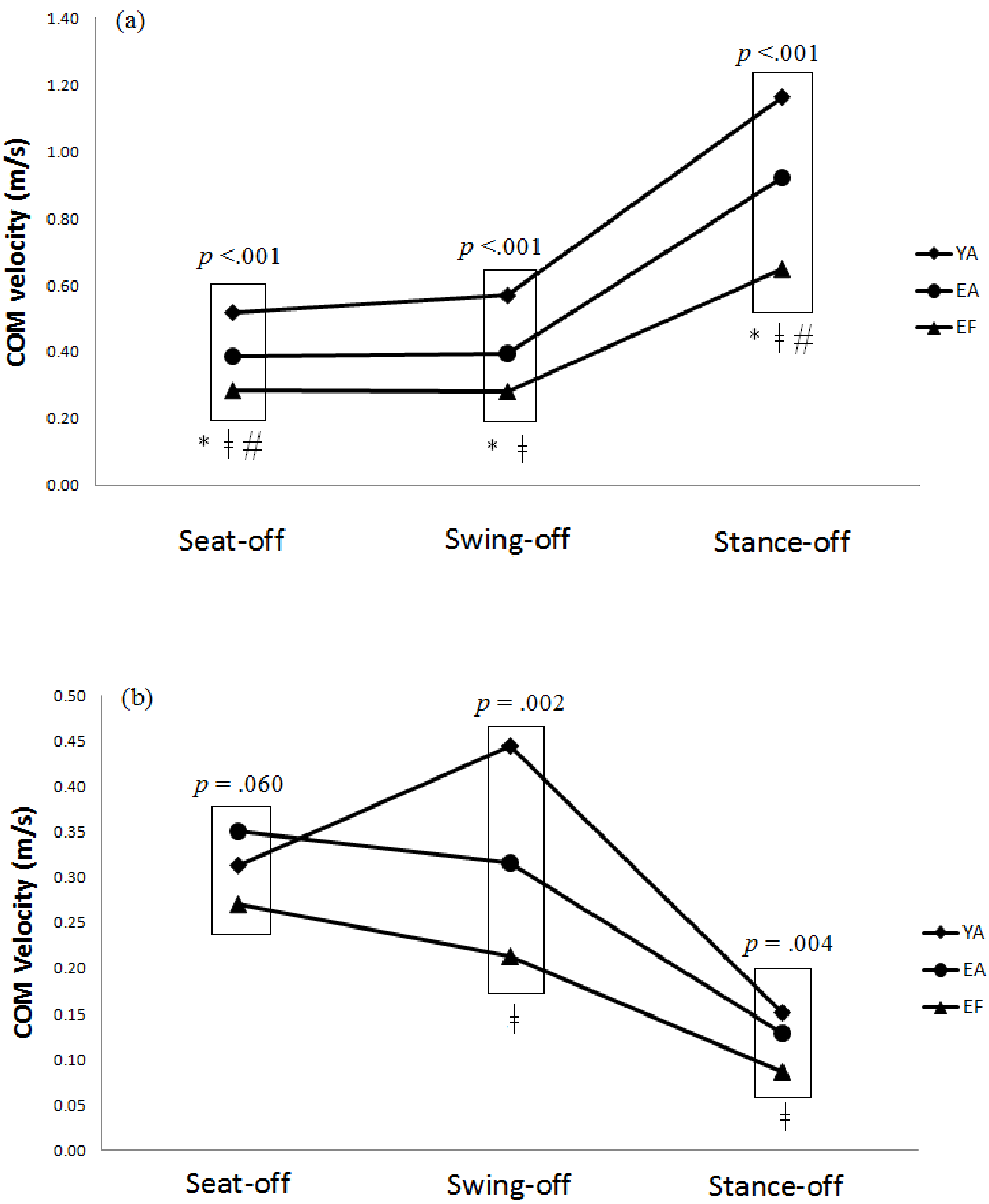

Figure 2.3. Average COM velocities at seat-off, swing-off and stance-off during STW (a) Forward direction, (b) upward direction.

(*: significant differences between YA and EA, ǂ significant differences between YA and EF, # significant differences between EA and EF)

Table 2.2. A-P COM ankle inclination angle at seat-off, swing-off, stance-off during STW, gait and turning for the three groups (a negative value indicates that COM is located posterior to the ankle position).

A-P COM-Ankle inclination angle	YA	EA	EF	p-value	p-value (* ǂ #)
STW					
Seat-off	-4.90 (3.11)	-2.83 (2.57)	0.01 (2.06)	< 0.001	0.054* < 0.001ǂ 0.003#
Swing-off	8.57 (1.92)	6.86 (2.42)	7.72 (2.35)	0.108	N/A
Stance-off	-8.44 (2.64)	-6.80 (2.15)	-2.40 (3.13)	< 0.001	0.494* < 0.001ǂ 0.002#
Gait					
Max. Posterior	13.78 (6.38)	7.12 (1.79)	3.58 (2.45)	< 0.001	< 0.001*< 0.001ǂ 0.022#
Max. Anterior	19.51 (6.03)	18.53 (2.57)	15.40 (3.98)	0.038	0.543* 0.015ǂ 0.060#
Turn (Range of A-P COM-Ankle angle)					
Before pivoting	19.40 (7.44)	16.78 (4.75)	6.36 (3.53)	< 0.001	0.654* < 0.001ǂ < 0.001#
Pivoting	6.24 (2.42)	5.41 (2.52)	2.96 (2.07)	0.002	1.000* 0.003ǂ 0.023#
After pivoting	19.07 (6.04)	10.15 (6.16)	3.74 (5.12)	< 0.001	0.001* < 0.001ǂ 0.014#

*p**: differences between YA and EA, *p*ǂ differences between YA and EF, *p*# differences between EA and EF

Significant group main effects were detected in the range of A-P COM-Ankle angles during the pivoting period and stance periods of the step immediately before and after. EF demonstrated significantly smaller angles than YA or EA during pivoting and its prior step. For the angles for the step after pivoting, YA presented the largest angles followed by EA and then EF (Table 2.2).

The length ratio between the actual transverse plane COM trajectories and fitted parabolic curves was found to be the largest in EF when compared to EA and EF (YA: 1.02 ± 0.02, EA: 1.03 ± 0.02, EF: 1.12 ± 0.14, $p = 0.007$, $p = 0.008$, respectively).

Table 2.3. Total kinetic energy and ratio of forward, medial-lateral and upward COM kinetic energy at seat-off, swing-off and stance-off for three groups.

	YA	EA	EF	p-value	p-value (* ǂ #)
Total COM kinetic energy (J) Mean(SD)					
Seat-off	12.4ǂ (4.1)	10.1# (2.3)	6.7ǂ # (4.2)	< 0.001	0.176* < 0.001ǂ 0.039#
Swing-off	18.9*ǂ (11.5)	10.4* (6.6)	6.6 ǂ (5.9)	0.001	0.022* 0.001ǂ 0.448#
Stance-off	45.1*ǂ (13.8)	31.9 * (10.9)	20.6 ǂ (15.2)	< 0.001	0.027* < 0.001ǂ 0.066#
Ratio of Kinetic Energy (%) Mean(SD)					
Seat-off					
Forward	70.6*ǂ (13.6)	53.8* (13.7)	50.3ǂ (15.2)	0.001	0.007* 0.001ǂ 1.00#
Medial-lateral	0.8 (0.8)	0.5 (0.4)	0.8 (1.2)	0.614	N/A
Upward	28.5*ǂ (13.7)	45.7* (13.6)	48.9ǂ (15.0)	0.001	0.006* 0.001ǂ 1.00#
Swing-off					
Forward	61.5 (11.8)	62.1 (17.3)	57.6 (14.7)	0.660	N/A
Medial-lateral	3.8 (3.8)	2.7 (2.6)	12.0 (17.9)	0.044	1.00* 0.125ǂ 0.067#
Upward	34.6 (14.6)	35.1 (17.7)	30.4 (16.3)	0.685	N/A
Stance-off					
Forward	96.1 (2.1)	96.0 (2.3)	87.9 (17.6)	0.054	N/A
Medial-lateral	2.2 (2.1)	1.8 (2.1)	10.2 (17.6)	0.050	N/A
Upward	1.7 (1.1)	2.2 (1.1)	1.8 (1.1)	0.502	N/A

p_*: differences between YA and EA, *p*≠ differences between YA and EF, *p#* differences between EA and EF

Differences between STW and Gait

In YA, no significant differences were detected in the COM forward velocities between stance-off of STW and during gait (Table 2.1, p = 0.374). The step length during gait was significantly larger than that at stance-off (Table 2.1, p = 0.021). For EA and EF, the gait velocity was significantly faster than the COM forward velocity at stance-off (Table 2.1, p = 0.001, p = 0.002, respectively). The step length at stance-off did not differ from that during gait for EA or EF (Table 2.1, p = 0.372, p = 0.963, respectively).

The COM-Ankle angles at stance-off in YA were significantly smaller than the maximum posterior COM-Ankle angles during gait (Table 2.2, p = 0.004). The COM-Ankle angle at stance-off in EA and EF did not differ from the peak posterior COM-Ankle angle during gait (Table 2.2, p = 0.493, p = 0.056 respectively). Moreover, the peak posterior COM-Ankle angle during gait in EA and EF was correlated significantly with the COM-Ankle angle at stance-off (r=.626, p=0.012 and r=.713, p=0.003, respectively), while the peak posterior COM-Ankle angle during gait in YA was correlated significantly with the COM-ankle angle at seat-off (r = 0.729, p = 0.002).

Table 2.4. Turn types and number of turn steps for three groups.

	YA (n=13)	EA (n=15)	EF (n=15)
Turn types			
A: Forward-1	13 (100%)	6 (40%)	1 (6.7%)
B: Forward-2	0	0	3 (20%)
C: Backward-1	0	6 (40%)	2 (13.3%)
D: Backward-2	0	3 (20%)	8 (53.3%)
E: Weight Shift	0	0	1 (6.7%)
No. of turn steps			
3-4	9 (69%)	3 (20%)	0
5-6	3 (23%)	9 (60%)	7 (47%)
7-8	1 (8%)	3 (20%)	8 (53%)

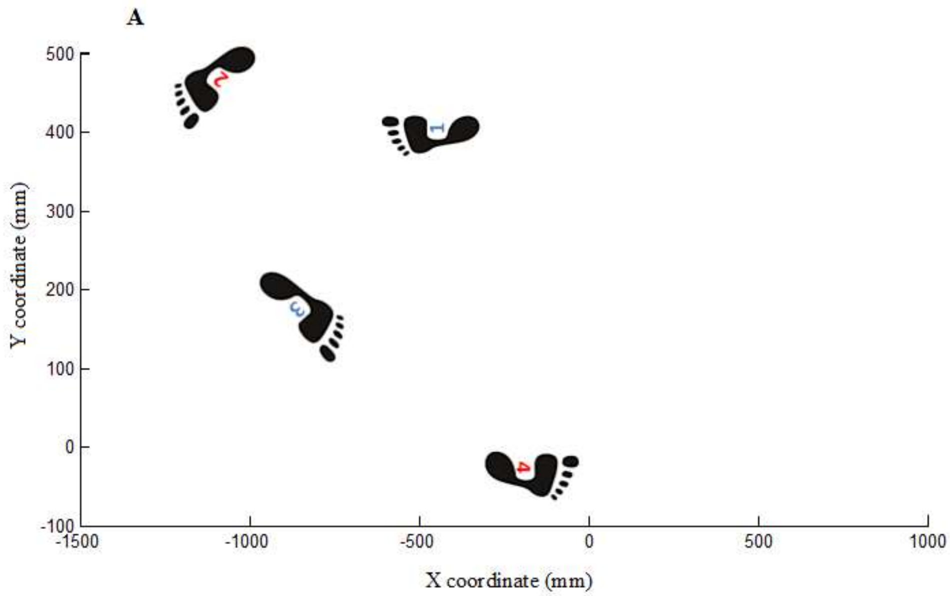

Figure 2.4. Turn pattern A (Forward-1): All steps are moving in a forward direction.

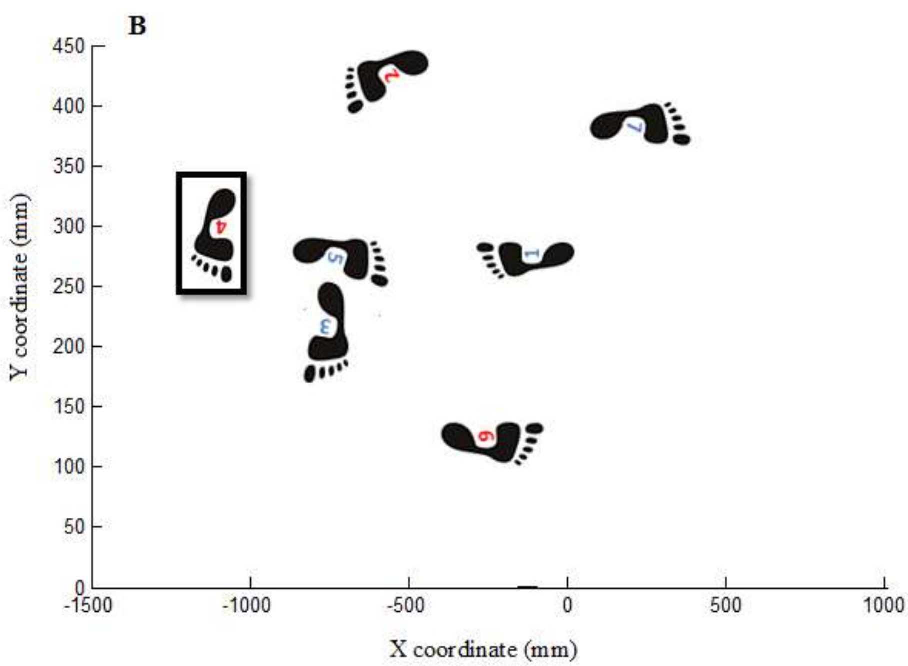

Figure 2.5. Turn pattern B (Forward-2): All steps are moving forward except the fourth step is not stepping forward but aligned approximately parallel to step 3.

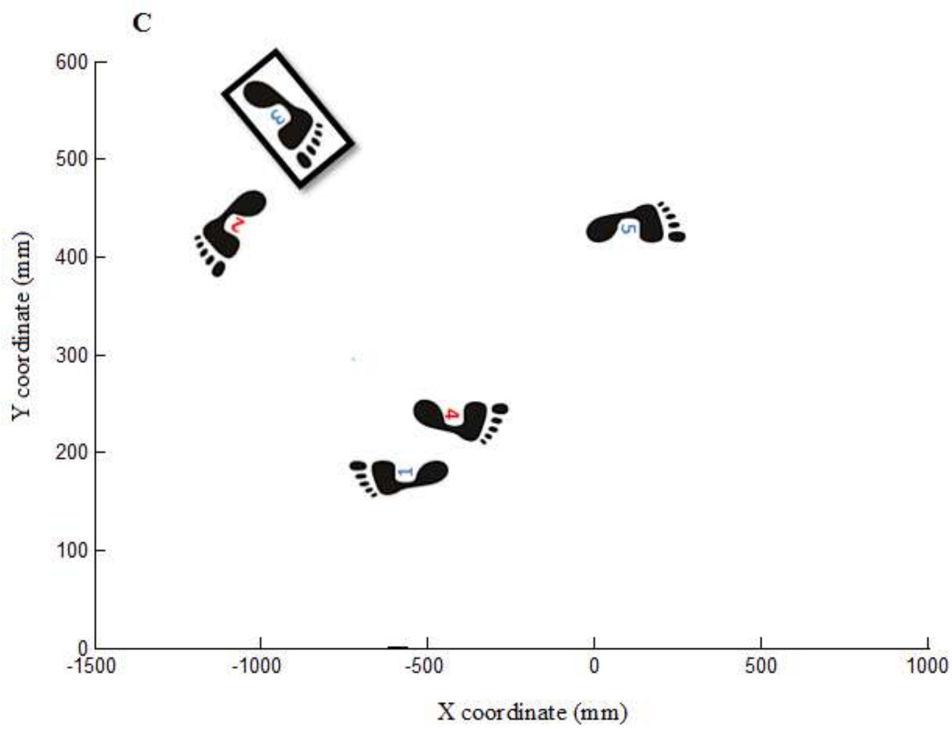

Figure 2.6. Turn pattern C **(**Backward-1): The third step is stepping backward and aligned approximately perpendicular to step 2.

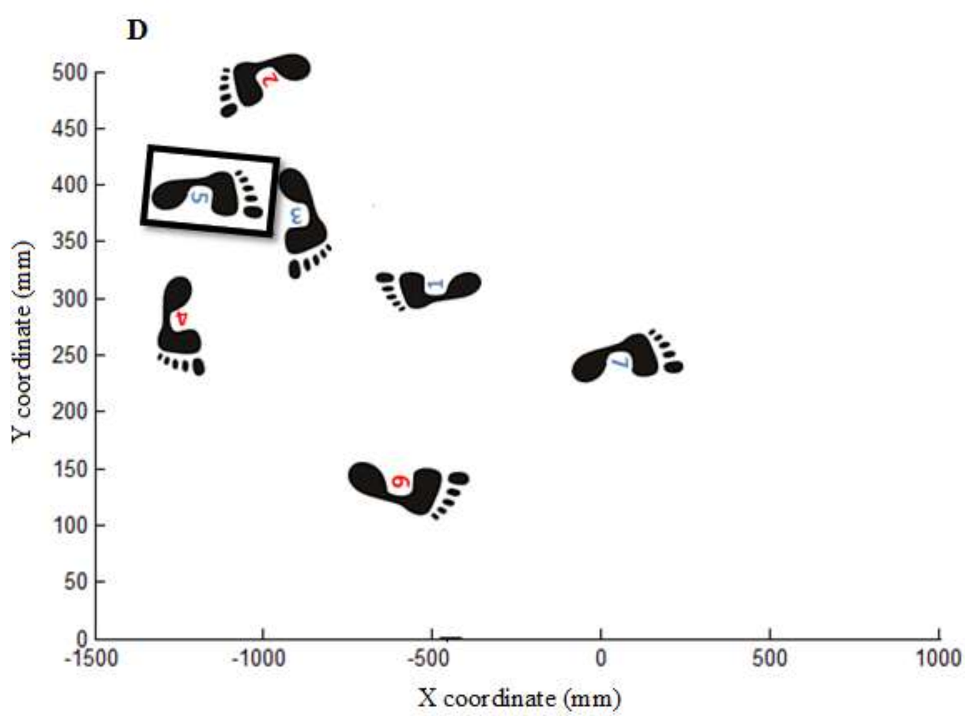

Figure 2.7. Turn pattern D **(**Backward-2): The fifth step is stepping backward and aligned approximately perpendicular to step 4. Step 3 is close to step 2 and aligned approximately perpendicular to step 2.

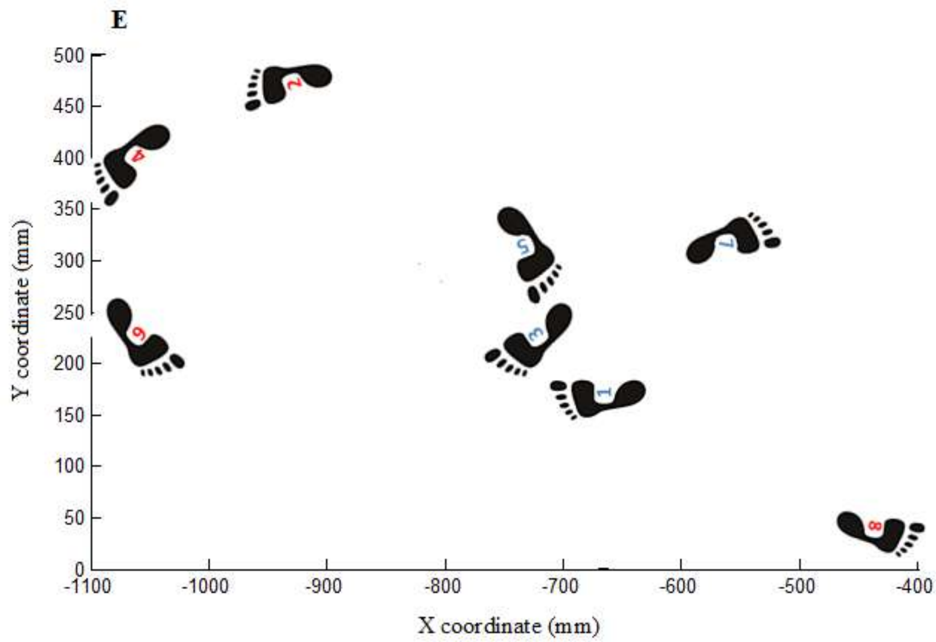

Figure 2.8. Turn pattern E (Weight shifting): wide step length, minimal forward step. Completing the turn with a series of weight shifting.

Differences between Gait and Turning

No significant differences were found between COM forward velocities at the start of the turn and during gait for all groups (Table 2.1, YA: $p = 0.138$, EA: $p = 0.151$, EF: $p = 0.637$). However, differences between COM forward velocities during gait and at the start of the turn ($Velocity_{gait} – Velocity_{start\ turn}$) were negatively correlated with the number of turn steps ($r = -0.582$, $p < 0.001$).

Discussion

This study examined COM motion control and its relationship to the BOS during the different phases and transitions of the TUG to demonstrate the effect of age and balance impairment. The fact that A-P COM-Ankle angle magnitudes were significantly greater in YA and EA compared to EF at seat-off during STW suggests that YA and EA are able to generate and more efficiently control a larger forward momentum during gait initiation. On the contrary, a small A-P COM-ankle angle at seat-off in EF suggests they place their COM within the BOS to achieve a stable posture before gait initiation. This decrease in COM-Ankle angle occurred in conjunction with reductions in both COM forward velocity and the initial step length in gait initiation. Taken together, these changes could reflect a strategy to prioritize balance maintenance over momentum generation during STW in EF.

At stance-off, YA already reached a similar COM forward velocity as during gait, but with a significantly shorter step length. In addition, the posterior COM-Ankle angle at stance-off was significantly smaller than the peak posterior angle during gait. These data suggested that YA did not wait to initiate gait at stance-off by taking a larger step. Instead, they carried over the COM momentum generated before swing-off to gait initiation. For EA and EF, the posterior COM-Ankle angles at stance-off were similar to those during gait. Moreover, the angles at stance-off and gait were found to be significantly correlated for EA and EF, while in YA, the angle at seat-off was significantly correlated with the angle during gait. These findings suggest that YA set the pace of gait from the beginning of the STW, while the older groups set their gait paces at stance-off.

EA completed the turning phase significantly slower than YA. However, no significant differences were found between these two groups for the TUG and STW durations. This suggests that healthy elderly adults encounter the most difficulty in the TUG with turning. Furthermore, significant differences in the ranges of A-P COM-Ankle angle were found between YA and EA after the turn. A smaller A-P angle suggests that

EA generated a smaller forward acceleration during the transition from turning to walking. Forty percent of EA demonstrated the turning patterns shown in Figure 2.6. Compared to the turning pattern shown in Figure 2.4, the pattern in Figure 2.6 includes a backward third step. This suggests that EA had a hesitancy to initiate walking immediately after the direction change.

Turning consists of decelerating the forward motion, rotating the body and stepping out toward the new direction [80]. Although EF demonstrated a smaller COM velocity at the beginning of the turn when compared with YA and EA, they still need to further decrease their COM velocity for direction change. Our findings suggested that individuals using more steps to complete a turn could demonstrate a failure to decelerate in the beginning of the turn, as EF took more steps to complete the turn than YA and EA. In addition, elderly fallers were found to have a smaller range of A-P COM-Ankle angles during turning than YA. It has been shown that the COM must be placed in a more posterior position in order to properly facilitate stepping toward to the new direction during turning [80]. A smaller A-P COM-Ankle angle suggests that EF were not able to place their COM at a more posterior position in relation to their ankle, and thus had trouble decelerating for the beginning of the turn [110]. Also, over half of EF demonstrated the turning pattern shown in Figure 2.7. Instead of placing the foot efficiently in the direction of forward progression, this pattern demonstrated a backward step (at Step five). The evidence indicates that elderly fallers did not have the ability to control their COM momentum during turning with fewer steps.

The length ratio between the actual COM trajectories and fitted parabolic curves was found to be the largest in the EF group. This ratio illustrated inefficient turning trajectories and excessive medial-lateral sway for EF. Since a large standard deviation was noticed, a further examination of data from each EF was performed. The two subjects demonstrating the largest length ratios (values) were identified as the ones with the most falls (# of falls). Thus, the feasibility of using this length ratio to identify individuals with a high risk of falling is worthy future investigation.

Several limitations need to be considered in this study. Seat-off was determined using peak vertical ground reaction force, which may not necessarily coincide with the seat-off timing determined by the pressure sensor on the bench. However, it has been reported that there was an excellent correlation between the two seat-off determination methods [84]. The COM-Ankle angle was used to represent the relationship between COM and BOS. Although COM-Ankle angle was able to demonstrate the COM control during single stance phase, future studies should investigate the control of COM with respect to BOS during double stance. Lastly, there were more women than men in the EF than in the EA group. Including gender matched control group would allow a better control for potential gender differences in balance control. A one-year prospective study found that a greater proportion of women (32.7%) than men (23.0%) experienced at least one fall and women had a higher risk of falling [111]. Our subject demographics represented this gender difference in the faller group.

In conclusion, our investigation of COM kinetic energy distribution and the relationship between COM and BOS revealed that both EA and EF (especially EF) prioritize balance maintenance over mobility during STW. More specifically, EA modified their STW strategy around the seat-off point so that they achieved a more upright position before walking. EF not only demonstrated altered COM control around the seat-off point, but also showed a limitation in COM control from swing-off to stance-off. The most challenging task during TUG for EA is turning, especially to initiate walking immediately after the direction change. Elderly fallers were found to require more steps to perform turning in order to control their COM momentum.

Bridge

Chapter II connected the clinical balance measure, the TUG, and the biomechanical balance parameter, COM-Ankle angle, by investigating the TUG using a motion analysis system. This chapter provided kinematics information of the TUG, especially the transition phases such as sit-to-walk (STW) and turning. It examined the differences in COM control during STW and turning between three subject groups: young adults, elderly adults, and elderly adults with fall histories. In Chapter III, we provide kinetics information of the STW phase of the TUG and examine the differences in joint kinetics and movement strategies among these three subject groups.

CHAPTER III

SAGITTAL PLANE MOVEMENT STRATEGY AND JOINT KINETICS DURING SIT-TO-WALK IN BALANCE IMPAIRED ELDERLY FALLERS

The study included in this chapter was developed by a number of individuals, including Dr. Li-Shan Chou and Dr. Chien-Chi Chang. Dr. Chou contributed substantially to this work by participating in the development of methodologies and providing critiques and editing advice. Dr. Chang was also helpful in providing critiques and editing advice. I was the primary contributor to the development of the protocol, data collection, data analysis and did all the writing.

Introduction

Rising from a chair is a common activity that requires both sufficient muscle strength and precise balance control [112, 113]. Difficulty in rising from a chair has been associated with increased fall risk in the elderly [3, 4]. This sit-to-stand (STS) motion typically precedes the initiation of gait, collectively the sit-to-walk (STW) motion, which is one of the common activities of daily living. During STW, it has been reported that elderly adults demonstrate a reduction in the horizontal center of mass (COM) momentum at seat-off in order to maintain a more stable upright posture before walking [86, 87]. Similar decreases in the COM forward velocity after seat-off were also observed in elderly fallers [88]. However, the underlying mechanism of such altered COM motion during STW has not been investigated. In order to develop effective screening and rehabilitation programs to prevent falling in the elderly, it is important to examine the association between COM movement strategies and joint kinetics during STW.

Balance control during locomotion could be examined with the COM motion and its interaction with the base of support (BOS) [74, 105]. Older adults with difficulties in performing STS were reported to more frequently utilize a strategy that shifts the center of pressure anterior to the ankle joint center to enhance postural stability [113, 114]. When compared to young adults, elderly adults generated a significantly larger ankle plantarflexor torque during STS in order to reduce the horizontal COM momentum to zero for achieving the final stable upright posture [112]. Greater challenges in balance control are expected when performing STW as opposed to STS in isolation [85]. STW motion requires the generation of a sufficient and controlled COM horizontal momentum for gait initiation and accommodation of a simultaneous narrowing base of support. The braking impulse was found to persist throughout the rising period during STS, while during STW it was rapidly switched to a propulsive impulse for gait initiation [85]. However, there is a lack of knowledge on associations among altered COM movement control, joint kinetics and ground reaction force (GRF) patterns to accommodate the increased biomechanical demands of STW in older adults.

Therefore, the purpose of this study was to examine differences in COM movement strategies and lower extremity joint kinetics during STW among three groups: healthy young adults, healthy elderly adults and elderly fallers with balance impairments. We hypothesized that, similar to STS, elder fallers would alter their COM movement strategy during STW, as compared to healthy young and older adults, to achieve a stable posture before gait initiation. This movement strategy could be demonstrated by a smaller

COM-Ankle inclination angle in the sagittal plane and would result in a delayed gait initiation. We also hypothesize that this different COM movement control in elderly fallers would be accompanied by different lower extremity joint moments and GRF patterns.

Methods

We recruited fifteen healthy young adults (YA) ranging from 18 to 35 years of age, fifteen healthy elderly adults (EA) and fifteen elderly fallers (EF) with balance impairment over the age of 70 years. Inclusion criteria for participants were 1) the ability to walk without the use of an assistive device; 2) no history of neurological or musculoskeletal deficits that might contribute to gait instability or falls, such as amputation, cerebral vascular accident, significant head trauma or Parkinson's disease; and 3) no uncorrectable visual impairment, vestibular dysfunction, or dementia. The EA were elderly individuals without fall histories and with a Fullerton Advanced Balance (FAB) scale score higher than 30 [68]. The FAB scale is a performance-based measure specifically designed for use with independently functioning elderly adults, with proven validity and reliability [67]. The EF in this study were defined as elderly individuals who had fallen twice or more in the year previous to the testing date. Furthermore, only falls that occurred during activities of daily living were included, so that falls caused by syncope or major intrinsic events such as stroke were excluded. In order to target fallers with balance impairments, a FAB scale score lower than 30 was required for EF. All participants agreed to the experimental procedure approved by the Institutional Review Board. A signed consent form was obtained from each subject before testing.

Participants performed the Timed Up and Go (TUG) test [61] while barefoot. They were asked to stand up from a bench, walk 3 meters, turn around, return to the bench and sit down. The following instructions were provided to all subjects: “Please complete the whole task at your comfortable speed, and we will time you.” The bench was set at each participant’s knee height. The bench for sitting was placed on one force plate in order to detect the seat-off point of the STW period. The other force plate was placed under the foot that provided the initial support to initiate walking (stance limb).

Twenty-nine markers were placed on selected bony landmarks of the subject [69]. Whole body motion data were captured with a 10-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA) with sample rate of 60 Hz and a low pass fourth-order Butterworth filter with 8 Hz cutoff frequency. Ground reaction forces were collected by two force plates (AMTI, Watertown, MA) and sampled at 960 Hz. Anthropometric reference data for both sexes were adapted from Dempster [109]. Whole body COM position was calculated as the weighted sum of a 13-segment model [69].

The overall time to complete the TUG was recorded. COM-Ankle angles were calculated as the inclination angles of the line formed by the COM and the lateral ankle (malleolus) marker in the sagittal plane for each frame during STW. Trunk angles in the sagittal plane were also calculated and were defined as the inclination angle of the line formed by the midpoint between two shoulder (acromion) markers and the midpoint between two pelvis (anterior superior iliac spine) markers. Flexion/extension moments of the hip, knee and ankle joint and sagittal plane ground reaction forces of the stance limb during STW were calculated using OrthoTrak™ software (Motion Analysis Corp., Santa Rosa, CA).

Data from the movement onset, seat-off, swing leg toe-off (swing-off), stance leg toe-off (stance-off) during STW were extracted for analysis. Movement onset was identified as the instant of initial change in COM forward position [106]. Seat-off was identified as the time when the magnitude of the vertical ground reaction force underneath the bench returned to baseline [84]. Swing-off and stance-off were identified as the times when the leading foot toe marker and trailing foot moved forward, respectively [106]. The duration of STW was defined as the period from movement onset to stance-off. Distances between the two lateral ankle markers in the frontal (step width) and sagittal plane (step length) were examined at stance-off (the first step during STW).

Group effects were assessed using multivariate analysis of variance (MANOVA) with an α level of 0.05. In order to control the effect of movement speed, STW duration was included as a covariate in the analysis for COM-Ankle angles, GRFs and hip and knee joint moments. STW duration was not included as a covariate for ankle joint moments due to its independence on speed [112]. Follow up analyses were performed using the Bonferroni adjustment. SPSS version 19.0 (IBM SPSS Inc., Chicago, IL) was used for all statistical analyses.

Results

All participants were able to successfully complete the TUG test. However, three elderly fallers needed to use their hands for support during STW. Also, due to multiple foot placements or technical issues, the force plate data from seven other elderly participants (5 EA/2 EF) and five young participants were not available. The data of the above fifteen participants were excluded from data analyses. Thus, results reported here are from 10 subjects in each group (Table 3.1). No significant differences were found among the investigated groups in weight and height (p = 0.393, p = 0.221 respectively). No significant age differences were found between the two elderly groups (p = 1.000). EA had a significantly higher FAB score than EF (p < 0.001).

Table 3.1. Subject characteristics [Mean (SD)].

Group	Age (yrs)	Sex (M/F)	Height (cm)	Weight (kg)	Falls	FAB
YA	24.5 (2.8)	5/5	168.7(6.2)	64.6(10.1)	0	N/A
EA	75.5 (3.0)	3/7	161.6(11.7)	67.8(16.6)	0	33.0(2.7)
EF	75.9 (4.1)	2/8	164.4(7.8)	74.7(21.2)	3.0(1.2)	25.1(3.3)

YA: Young adults, EA: Healthy elderly adults, EF Elderly fallers.
FAB: Fullerton Advanced Balance scale.

Significant group main effects were detected in the TUG duration, STW duration and step length at stance-off (Table 3.2). Post-hoc analysis showed that EF completed the entire TUG test with a significantly longer time than YA or EA. When compared to YA, STW duration was significantly longer for EF. At stance-off, YA took a significantly larger step than both EA and EF. Additionally, the EA had a larger step length at stance-off than EF. No significant differences were found among groups in the step width during STW. Sagittal plane trunk angles at seat-off did not differ significantly among groups (YA=36.5^{o} ± 7.1^{o}, EA=31.4^{o} ± 6.8^{o}, EF=32.9^{o} ± 9.3^{o}, p = 0.333).

Magnitudes of the anterior-posterior (A-P) COM-Ankle angle were significantly different between EF and YA or EA at seat-off (Table 3.3). Both YA and EA placed their COM at a more posterior position to the ankle position than EF. No significant differences were detected between YA and EA. At swing-off, EF demonstrated a significantly greater anterior COM-Ankle angle than EA (Table 3.3). No significant group differences were found in the A-P COM-Ankle angle at stance-off.

Table 3.2. Timed Up and Go test and sit-to-walk duration, step length and step width at stance-off for three groups [Mean (SD)].

	YA	EA	EF	p-value	p-value (* ǂ #)
TUG duration (s)	7.94 (0.95)	9.90 (1.53)	12.71 (3.84)	0.001	0.085* < 0.001ǂ 0.016#
STW duration (s)	1.33 (0.16)	1.48 (0.15)	1.90 (0.57)	0.003	0.343* 0.001ǂ 0.012#
Step length (m)	0.59 (0.06)	0.52 (0.08)	0.42 (0.06)	< 0.001	0.144* < 0.001ǂ < 0.001#
Step width (m)	0.21 (0.08)	0.21 (0.07)	0.23 (0.07)	0.714	N/A

*p**: differences between YA and EA, *p*≠ differences between YA and EF, *p*# differences between EA and EF.

Table 3.3. A-P COM ankle inclination angle at seat-off, swing-off, stance-off during STW for three groups (a negative value indicates that COM is located posterior to the ankle position).

A-P COM-Ankle inclination angle (degrees)	YA	EA	EF	p-value	p-value (* ǂ #)
Seat-off	-4.06 (2.31)	-3.03 (2.57)	0.51 (2.19)	< 0.001	0.667* < 0.001ǂ 0.003#
Swing-off	7.82 (1.49)	6.76 (2.64)	8.10 (2.61)	0.038	0.556* < 0.057ǂ 0.012#
Stance-off	-7.59 (2.46)	-6.82 (2.66)	-3.54 (2.86)	0.184	N/A

*p**: differences between YA and EA, *p*≠ differences between YA and EF, *p*# differences between EA and EF.

Patterns of the supporting limb GRFs in anterior-posterior and vertical directions were similar for all participants (Figure 3.1), demonstrating a braking impulse followed by a propulsive impulse at push-off (Figure 3.1-a). When compared to YA and EA, EF demonstrated a significantly greater braking and smaller propulsive impulses (Table 3.4, $p < .001$). The time elapsed between seat-off and onset of the propulsion impulse, as percentage of total STW duration, was the largest in EF (YA: 4.8 ± 5.3, EA: 9.0 ± 5.5, EF: 13.1 ± 6.5, $p = 0.013$). When compared to YA, EF demonstrated a larger peak breaking GRF (YA: -0.03 ± 0.02, EA: 0.04 ± 0.01, EF: 0.06 ± 0.02, $p = 0.001$). Additionally, when compared to YA, both EA and EF demonstrated a smaller peak propulsive GRF (YA: 0.30 ± 0.06, EA: 0.22 ± 0.06, EF: 0.15 ± 0.05, $p = 0.003$, $p < 0.001$, respectively) and a smaller peak vertical GRF (YA: 1.19 ± 0.09, EA: 1.04 ± 0.05, EF: 1.02 ± 0.08, $p < 0.001$) at push-off.

Flexion/extension moments of the hip, knee, and ankle of the stance limb were examined between seat-off and stance-off (Figure 3.2). YA demonstrated a trend of generating a larger hip extensor moment at seat-off than EA and EF (Table 3.4). The knee extensor moment did not differ among groups (Table 3.4). No significant group differences were found in hip and knee moment at swing-off. However, significant group differences in the ankle moment were found at seat-off and swing-off (Table 3.4). Both YA and EA showed a dorsiflexor moment while EF demonstrated a plantarflexor moment at both seat-off and swing-off.

Discussion

This study analyzed the STW phase of the clinical mobility assessment, TUG, to demonstrate changes in movement strategies and joint moments associated with aging and balance impairment. In general, when compared to healthy adults, elderly fallers exhibited significant changes in their movement control when performing STW. These

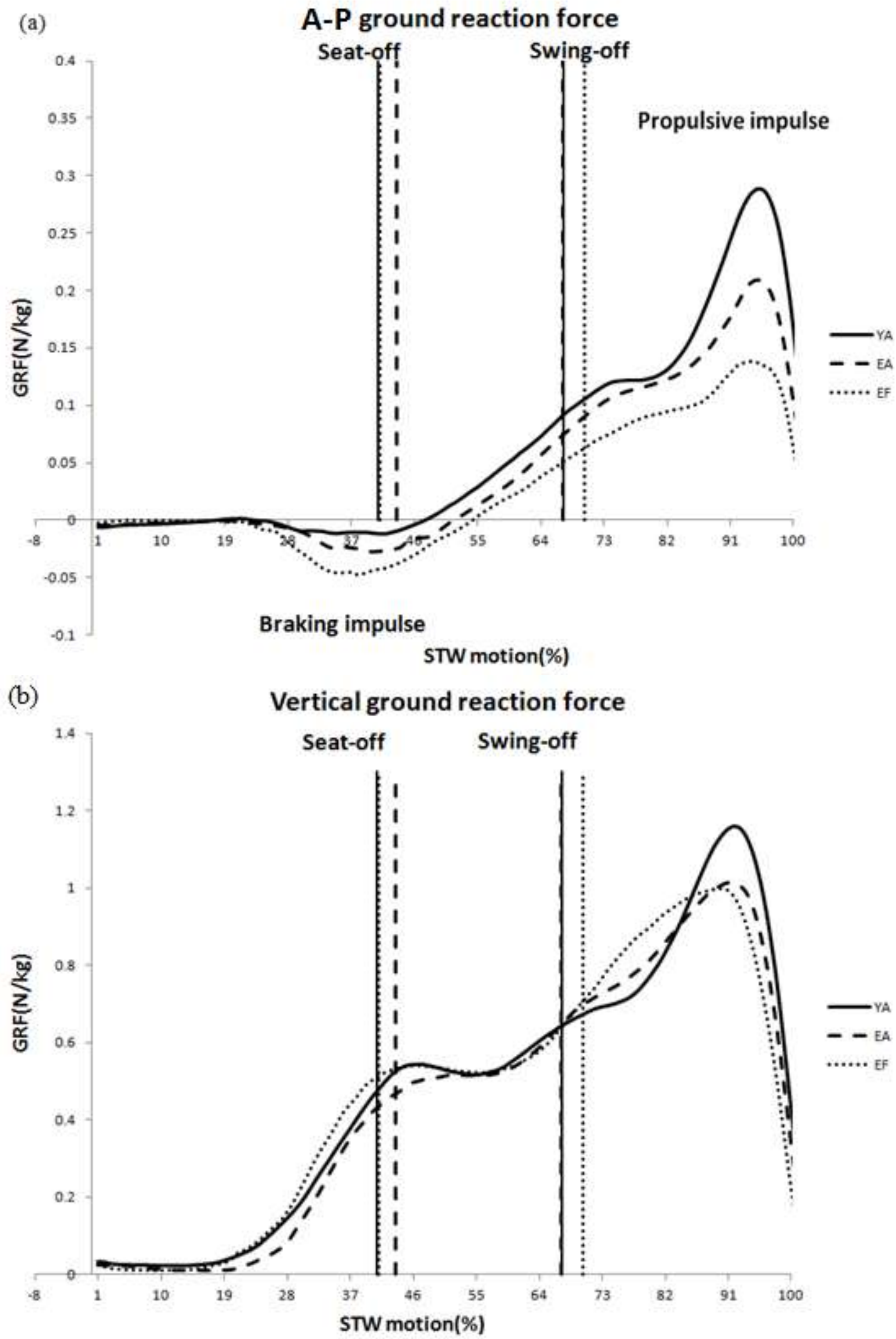

Figure 3.1. Average ground reaction forces during STW for each group (a) anterior-posterior (A-P) direction (b) vertical direction.

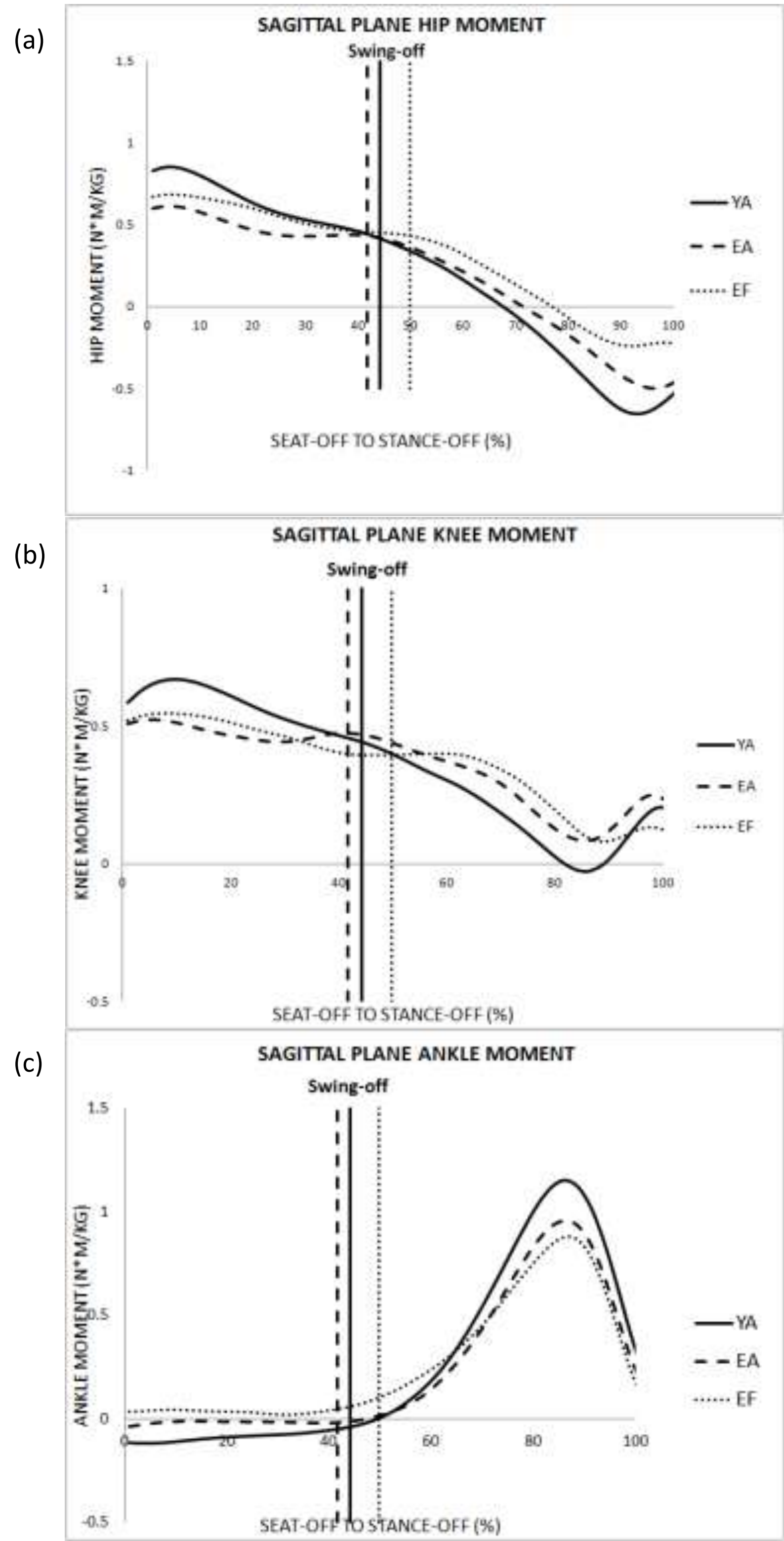

Figure 3.2. Average joint moment from seat-off to stance-off during STW for each group (a) Hip, (b) Knee, (c) Ankle.

Table 3.4. Braking, propulsion impulse and sagittal plane joint moment during STW for three groups.

	YA	EA	EF	p-value	p-value (* ǂ #)
Impulse (N/kg*s)					
Braking	-0.24 (0.25)	-0.43 (0.18)	-0.83 (0.41)	< 0.001	0.168* < 0.001ǂ 0.005#
Propulsion	6.64 (1.37)	5.04 (1.27)	3.48 (1.13)	< 0.001	0.009* < 0.001ǂ 0.010#
Joint Moments (N-m/kg)					
At seat-off					
Hip	0.83 (0.16)	0.66 (0.31)	0.67 (0.14)	0.266	N/A
Knee	0.58 (0.19)	0.56 (0.24)	0.52 (0.17)	0.759	N/A
Ankle	-0.11 (0.06)	-0.05 (0.10)	0.03 (0.14)	0.022	0.217* 0.006ǂ 0.103#
At swing-off					
Hip	0.39 (0.27)	0.48 (0.24)	0.45 (0.23)	0.370	N/A
Knee	0.53 (0.16)	0.54 (0.19)	0.42 (0.23)	0.834	N/A
Ankle	-0.08 (0.09)	-0.03 (0.08)	0.11 (0.25)	0.035	0.422* 0.012ǂ 0.071#

*p**: differences between YA and EA, *p*≠ differences between YA and EF, *p*# differences between EA and EF

changes include a smaller posterior COM-Ankle angle, a larger braking impulse and a prominent ankle plantarflexor moment at seat-off and a greater anterior COM-Ankle angle and plantarflexor moment at swing-off.

The fact that YA and EA demonstrated a greater posterior COM-Ankle angle as compared to EF at seat-off suggests that YA and EA were able to generate and more efficiently control a larger COM forward momentum for gait initiation. These results were similar to those reported for STS [112, 113]; the strategy demonstrated by the YA in the STS paradigm was documented as "momentum transfer" [115]. The momentum-transfer strategy utilizes the momentum generated prior to seat-off to assist in standing up;

the forward velocity of the COM at seat-off is used to facilitate a smooth transition into gait initiation. However, only individuals who are able to control a large COM forward momentum after seat-off are capable of performing a momentum-transfer strategy, or the body will fall forward and become imbalanced [115].

On the other hand, a smaller posterior COM-ankle angle at seat-off in EF suggested that they placed their COM above the BOS to achieve a stable posture prior to standing up. This is similar to the "zero momentum" strategy during STS [115], in which elderly adults were found to increase their truck flexion prior to seat-off and reposition the COM. However, such a strategy could impose a greater muscular demand on the lower extremities than a "momentum-transfer" strategy [115]. In the current study, we found that sagittal plane trunk angles at seat-off did not differ significantly among groups. This suggests that during STW, elderly fallers placed their ankle posteriorly to achieve a more stabilized posture at seat-off. This posterior placement of ankle position has been reported to reduce the COM forward displacement and momentum and make STS less difficult [116].

EF demonstrated a significantly greater braking impulse and smaller propulsive impulse as compared to YA or EA. This greater braking impulse is used to reduce the forward COM momentum, and results in a longer STW time. The braking impulse in STS has been reported to be greater than the braking impulse during the STW motion in YA [85]. Our data showed that the time elapsed between seat-off and onset of the propulsion impulse was the largest in EF.

At swing-off, elderly fallers demonstrated a greater anterior COM-ankle angle than EA. This increased forward inclination could imply EF's attempt to induce a forward momentum to initiate walking in order to compensate for the lack of momentum generation at seat-off. In addition, magnitudes of COM-ankle angle at swing-off were found to be similar to those at stance-off for EA and YA, which demonstrated their ability to reach the rhythm of walking at swing-off. On the contrary, EF had a smaller value of A-P COM-Ankle angle at stance-off than at swing-off. Taken together, these results suggest that EF prioritized balance maintenance over momentum generation at seat-off and delayed their gait initiation till swing-off.

Significant differences in the sagittal plane ankle moment were found between groups at seat-off and swing-off. Both YAs and EAs showed a dorsiflexor moment while EF demonstrated a plantarflexor moment at seat-off and swing-off. This could be related to the reduced ankle dorsiflexor strength that has been reported in elderly adults with a history of multiple falls [117]. In addition, due to a reduced forward COM momentum and smaller hip/knee extensor moments, EF had to increase the use of their ankle plantarflexor to push up to a standing position. The findings of reduced initial hip extensor moments and increased ankle plantarflexor moments among EF individuals resulted in diminishing the horizontal momentum during seat-off. This provides evidence that EF make changes in their movement strategy to ensure greater stability during STW.

Limited by the laboratory setting, only one force place was used to measure GRFs under the stance limb. Future studies would be benefited by the evaluation of both the loading and unloading limb during STW. The COM-Ankle angle was used to represent the relationship between COM and the BOS. Although COM-Ankle angle was able to demonstrate the COM control during the single stance phase of gait initiation, future studies should include the control of COM with respect to BOS during the double stance phase of gait initiation. Lastly, the study sample size was reduced since some participants made multiple foot placements on the force plate during gait initiation; therefore their data were not included in the data analysis. However, a post-hoc power analysis was performed using the ankle joint moment at seat-off. Our sample size of 30 subjects had 77% power to detect group differences at the 0.05 level with an effect size of 0.27.

In conclusion, the results obtained from the investigation of movement strategies and joint kinetics during STW showed that elderly adults, especially elderly fallers, place a greater priority on maintaining a stable posture than gait initiation during STW. Our results, taken together, demonstrated that EF adopted a movement strategy that includes a more stabilized posture at seat-off with a more posterior foot placement, a longer time duration, and a reduced initial step during gait initiation. In addition, EF increased the braking impulse to reduce their forward COM momentum. The delayed onset of propulsive impulse (gait initiation) ensured a stable upright position before gait initiation. When the hip extensor moments were reduced in EF, a greater ankle plantarflexor moment was initiated at seat-off to achieve a successful STW movement. While the increased ankle plantarflexor moment might negatively impact generation of COM momentum, it could be a movement strategy to improve stability during STW.

Bridge

Chapter III shows that elderly fallers modified their ankle joint moment to ensure a movement strategy that provided more stability for STW. The elderly fallers demonstrated a larger braking impulse and a delayed onset of propulsion impulse in the forward ground reaction profiles. All these adjustments could explain why elderly fallers demonstrated a longer STW phase in the TUG. Chapter IV investigates the relationship between longer TUG component' times (STW and turn durations) and two underlying physiological factors (muscle strength and balance control).

CHAPTER IV

IMPACTS OF MUSCLE STRENGTH AND BALANCE CONTROL ON SIT-TO-WALK AND TURNING DURATIONS IN THE TIMED UP AND GO TEST

The study included in this chapter was developed by a number of individuals, including Dr. Li-Shan Chou and Dr. Marjorie Woollacott. Dr. Chou contributed substantially to this work by participating in the development of methodologies and providing critiques and editing advice. Dr. Woollacott was also helpful in providing critiques and editing advice. I was the primary contributor to the development of the protocol, data collection, data analysis and did all the writing.

Introduction

The Timed Up and Go test (TUG) is a commonly used clinical assessment for screening mobility impairment and fall risk [59, 61, 66]. The TUG measures the amount of time required for a person to stand up from a chair, walk three meters, turn around, return to the chair and sit down. Overall, the TUG time increases with age and use of assistive devices [62]. The TUG has been able to identify individuals who have cognitive impairment [63] and a higher risk of falling [64]. The TUG time was reported to moderately correlate with gait speed, the Berg Balance Scale, and the Barthel Index [65]. Findings from these studies provide important information on how overall TUG time could correlate with some age-related functional declines; however, they do not reveal how these declines affect each of the tasks included in the TUG, especially sit-to-walk (STW) and turning.

Many factors have been shown to influence the overall TUG time in elderly adults [62-65]; however, their specific effects on each of the individual tasks, such as sit-to-stand or turning are not clear. Some factors are not modifiable (such as age), while others can be improved with training programs (e.g. strength and balance). Exercises comprising balance and strength trainings have been reported to be able to efficiently reduce fall incidents [23]. Muscle weakness is highly associated with the sit-to-stand component of the TUG [118]. Impaired balance control could be more likely to affect TUG time than gait time alone [119]. A better understanding of the association between declines in muscle strength or balance control ability and changes in performance during sit-to-walk (STW) and turning of the TUG in the elderly could enhance our ability to detect fall risk and develop preventive interventions.

Although clinical balance tests can indicate balance abilities [56, 65, 68], they could be unable to reveal underlying mechanisms causing imbalance during dynamic tasks. The measurement of the instantaneous position of the whole body center of mass (COM) with respect to the center of pressure (COP) can reflect postural alignment and quantify balance control during gait. Elderly adults with balance impairment demonstrated a greater medial and smaller anterior COM-COP inclination angle than healthy elderly adults during walking [74, 105]. When the COP is not available, the COM-Ankle inclination angle during the single stance phase provides an alternative assessment [75]. Examining the COM-Ankle inclination angles during the performance of the TUG test could allow us to better quantify how balance control is maintained or perturbed in the types of functional tasks included in the TUG and their transitions.

The purpose of this study was to examine the association of muscle strength and balance control ability with increased sit-to-walk or turning durations during the TUG test in the elderly. To enhance our understanding of possible mechanisms underlying a longer sit-to-walk (STW) or turn duration, COM-Ankle inclination angles derived from each of these tasks in the TUG were selected to indicate balance control ability. Strength measurements of hip abductors, knee extensors, and ankle plantar-flexors were included in this study due to their important roles in weight support. We hypothesized that the muscle strength would be more associated with STW time as compared to turning time, and the balance control would be better correlated with turning time as compared to STW time.

Methods

To achieve power of .80 and a large effect size ($f^2 = .35$), a sample size of 46 is required to detect a significant linear multiple regression model with six predictors (four confounding variables and two predictors). We recruited sixty elderly adults over the age of 70 years from the community. Inclusion criteria for elderly participants were individuals who (1) could walk without the use of an assistive device; (2) had no history of neurological or musculoskeletal deficits that might contribute to gait instability or falls such as amputation, cerebral vascular accident, significant head trauma, or Parkinson's disease; and (3) had no uncorrectable visual impairment, vestibular dysfunction, or dementia. All participants agreed to the experimental procedure approved by the Institutional Review Board and signed consent forms prior to their participation in the study.

Factors that could affect the TUG performance time were recorded and treated as potential confounding variables in the analysis, which include age, cognitive status, activity of daily living function (ADL) and history of falls. Berg balance scores (BBS), Fullerton Advanced Balance (FAB) scales, and whole body motion data while performing the TUG test were collected.

Participants performed the TUG while barefoot. They were asked to stand up from a bench, walk three meters, turn around, return to the bench and sit down. The following instruction was provided to all of the subjects: "Please complete the whole task at your comfortable speed, and we will time you." Participants placed both feet on a force plate as they stood up from sitting on a bench. The height of the bench was adjusted to each participant's knee height.

Twenty-nine markers were placed on selected bony landmarks of the subject [69]. Whole body motion data during the TUG were captured with a 10-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA) with sample rate of 60 Hz and a low pass fourth-order Butterworth filter with an 8 Hz cutoff frequency. Two force plates (AMTI, Watertown, MA) were used to collect ground reaction forces with a sampling rate of 960 Hz. Anthropometric reference data for both sexes were adapted from Dempster [109]. Whole body COM position was calculated as the weighted sum of a 13-segment model [69].

Peak joint moments during the isometric maximal voluntary contraction (MVC) test for bilateral hip abductors, knee extensors, and ankle plantar flexors were determined using the Biodex dynamometer (Biodex Medical Systems, Inc., NY). These peak joint moments were used to indicate the strength for each muscle group. The MVC strength of the hip abductors was measured in a standing position with neutral hip position. The testing leg was strapped to the force measuring device with the point of application at the lateral distal end of the femur. Furthermore, in order to maintain one leg standing position during the testing, subjects were allowed to hold the test administrator's shoulders. Subjects then were instructed to abduct the hip against the application pad without rotating the lower extremity or moving the trunk in any plane. The knee extensor isometric strength was measured in a seated position at 60 degrees of knee flexion with respect to the longitudinal axis of the thigh segment. The force measuring device was placed at four fingers width above the bilateral malleolus. Isometric strength of ankle plantar flexor was measured in the seated position at 20 degree of knee flexion and neutral ankle position.

While testing each joint, the dynamometer was aligned so that the axis of the lever arm coincided with the joint axis of rotation. Each subject was allowed two sub-maximal practice trials for each joint function. Subjects were instructed to push as hard as they could for a period of 5 seconds. A resting period of one minute or longer as necessary was given between repetitions to minimize the joint fatigue. Data from three trials were collected for each joint function. For each joint, the maximum value from three trials was selected for further analysis. The data from both sides for each joint were then averaged and normalized to body weight (kg).

The overall time to complete the TUG was recorded. COM-Ankle angles were calculated as the inclination angles of the line formed by the COM and lateral ankle (malleolus) markers in the sagittal and frontal planes for each frame during the TUG. Duration for STW was determined from the movement onset to stance leg toe-off (stance-off). Movement onset was identified as the instant when COM forward position changed [106]. Seat-off was identified as the time when the vertical ground reaction force peaked [84]. Swing leg toe-off (swing-off) and stance-off were identified as the times when the leading foot toe marker and trailing foot moved forward, respectively [106]. The sagittal plane COM-Ankle angles from seat-off, swing-off, and stance-off, and the frontal plane COM-Ankle angles from swing-off and stance-off during STW were extracted for analysis. The COM range of motion in the frontal plane during STW was also examined.

Turning duration was calculated from start-of-turn to end-of-turn. The start-of-turn was defined as the toe-off immediately prior to the instant when the transverse plane pelvic rotation exceeds the range of rotation detected during walking. The end-of-turn was defined as the sagittal plane COM position after the turn reaches the same position at the start-of-turn. The supporting limb that made the first direction change is defined as the "turning limb," and this stance period is termed "pivoting." Ranges of the COM-Ankle angles during pivoting and stances periods of the step prior to and after pivoting were extracted for analysis.

Correlations between the time duration for each TUG component (STW and turning) and the following variables were examined: muscle strength (hip abductors, knee extensors and ankle dorsiflexors), clinical balance measures (BBS and FAB), and laboratory balance measures derived from TUG testing (frontal plane COM motion during STW and COM-Ankle angles at different events during STW and Turning). Statistical analyses were performed in the following steps. First, the Pearson correlation was used to examine the correlation between each TUG component time and each individual variable (univariate approach). The factors that were significantly correlated with the TUG component time were included in the multiple regression analysis (multivariate approach). Second, due to the existence of a moderate to high inter-correlation ($r > 0.5$) between several variables in each category (muscle strength or balance control), principle component regression, which combines linear regression with principal component analysis (PCA) [120], was conducted. PCA generated a principle component for each category and ensured that multicollinearity did not compromise the reliability of the regression. Third, we constructed a series of multiple regression models to examine the combined contribution of all factors to predict each TUG component time after controlling for potential confounding factors (age, cognitive impairment, ADL, and past fall history). The potential confounding factors for TUG performance times were entered into Model 1. In Model 2, variables used in Model 1 plus the muscle strength variables were entered. Finally, in Model 3, variables of Model 2 plus the balance control variables were entered. The significant level for all statistical tests was set at $\alpha = 0.05$. SPSS version 19.0 (IBM SPSS Inc., Chicago, IL) was used for all statistical analyses.

Results

Data from three subjects were excluded because their TUG durations exceeded three times the standard deviation of the group mean. The average age of the 57 participants included in the analysis was 77.8 years (Table 4.1). Approximately one-third of the participants (28.1%) had a history of two or more falls. The mean overall TUG time, FAB score and BBS score were 11.9 ± 3.0 seconds, 27.6 ± 7.6 and 52.8 ± 5.4, respectively.

Table 4.1. Subject demographics and characteristics (N= 57).

	Mean	Std. Deviation
Age (year)	77.8	5.8
Height (cm)	163.4	9.3
Weight (kg)	71.8	18.2
Past fall history ($\geq$ 2 falls) %	28.1	N/A
Activities of daily living (Max score = 6)	5.83	0.5
SLUMS Examination (Max score = 30)	26.8	3.1
Fullerton advanced balance scale (Max score = 40)	28.1	6.9
Berg balance scale (Max score = 56)	53.3	3.9
Activities-specific Balance Confidence scale (Max score = 100)	85.8	15.7
Hip abductor strength (N-m/kg)	0.55	0.16
Knee extensor strength (N-m/kg)	1.30	0.43
Ankle plantar-flexor strength (N-m/kg)	1.07	0.39
Average of Hip abductor and Knee extensor strengths (N-m/kg)	0.92	0.27
Timed up and go test duration (s)	11.85	3.02
Sit-to-walk duration (s)	1.73	0.43
Turn duration (s)	2.77	0.97
	Value	
% Female subjects	66.7	N/A
Use of assistive device (%)	7.0	N/A

Both STW and turning durations were significantly correlated with FAB and BBS (Table 4.2). STW duration was significantly correlated with hip abductor strength, knee extensor strength, sagittal and frontal plane COM-Ankle angles at swing-off, sagittal plane COM-Ankle angle at stance-off, and frontal plane COM range of motion. Turning duration was significantly correlated with hip abductor strength and the sagittal plane COM-Ankle angles before and after pivoting (Table 4.2).

Significant correlations were found among all strength measurements (hip/knee: $r = 0.559$, hip/ankle: $r = 0.643$, knee/ankle: $r = 0.521$, $p < 0.001$) and between two clinical

Table 4.2. Pearson correlation coefficients between STW and turning durations and muscle strength, clinical and laboratory balance measures.

a. Predictors	STW duration Pearson coefficients	p-value
Strength		
HIP	-0.369*	0.002
KNEE	-0.248*	0.031
ANKLE	-0.205	0.063
Balance		
FAB	-0.480*	< 0.001
BBS	-0.473*	< 0.001
Sagittal plane COM-Ankle angle seat-off	-0.088	0.258
Sagittal plane COM-Ankle angle swing-off	-0.325*	0.007
Sagittal plane COM-Ankle angle stance-off	0.460*	0.000
Frontal plane COM-Ankle angle swing-off	0.247*	0.032
Frontal plane COM-Ankle angle stance-off	0.100	0.230
Frontal plane COM range of motion	0.407*	0.001

b. Predictors	Turning duration Pearson coefficients	p-value
Strength		
HIP	-0.248*	0.032
KNEE	-0.166	0.108
ANKLE	-0.023	0.433
Balance		
FAB	-0.463*	< 0.001
BBS	-0.499*	< 0.001
Sagittal plane COM-Ankle angle before pivoting	-0.325*	0.007
Sagittal plane COM-Ankle angle during pivoting	-0.198	0.070
Sagittal plane COM-Ankle angle after pivoting	-0.319*	0.008
Frontal plane COM-Ankle angle before pivoting	-0.089	0.257
Frontal plane COM-Ankle angle during pivoting	-0.022	0.437
Frontal plane COM-Ankle angle after pivoting	-0.066	0.313

* Items that were selected to enter multiple regression analysis, $p < 0.05$

balance measurements (FAB/BBS: $r = 0.807$, $p < 0.001$). Several biomechanical balance parameters during STW were significantly correlated with clinical balance measures (e.g., sagittal plane COM-Ankle angle at stance-off with FAB, $r = 0.608$ and with BBS, r

= 0.392). Many biomechanical balance parameters during turning were also significantly correlated with clinical balance measurements (e.g., sagittal plane COM-Ankle angle before pivoting with FAB, $r = 0.439$ and with BBS $r = 0.336$)

One principle component, named Balance_STW, was generated from PCA for balance variables predicting STW duration (FAB, BBS, sagittal and frontal plane COM-Ankle angles at swing-off, sagittal plane COM-Ankle angle at stance-off, and frontal plane COM range of motion). The other principle component, named Balance_Turn, was generated for balance variables predicting turn duration (FAB, BBS and sagittal plane COM-Ankle angle before and after pivoting). Moderate correlations were found between Balance_STW and STW duration ($r = -0.529$, $p < 0.001$) as well as Balance_Turn and turning duration ($r = -0.437$, $p = 0.001$). Hip abductor strength was selected for representing strength category to predict turning duration due to its significant correlation. Both hip abductor and knee extensor strengths were significantly correlated with STW duration; therefore, the average value of hip abductor and knee extensor strength was used to represent strength in the prediction of STW duration (Table 4.1).

In multiple regression analysis, the R square values revealed that 3.5 % of the variance in STW duration and 13.7% of the variance in turning duration were explained by the confounding variables in Model 1 (Table 4.3). When strength variables were included (Model 2), the explained variance was increased by 8.9% for STW duration and 1.8% for turning duration. Significant F change was detected between Models 1 and 2 for STW duration prediction (Table 4.3, $p = 0.030$). When balance variables were added to the model (Model 3), the explained variance was significantly increased by 18.7% for STW duration and 7.4% for turning duration ($p = 0.001$ and 0.038, respectively, for significant F changes). Model 3 explained a statistically significant portion of variance in STW and turning durations (p = 0.005, p = 0.049, respectively). After controlling for demographics, cognition, fall history, and ADL function, balance control was significantly related to STW and turning durations whereas muscle strength was not (Table 4.4).

Table 4.3. Variance predicted by strength and balance variables: R Square and R Square change from model 1 to model 3.

Model		R Square	R Square Change	F Change	Sig. F Change
STW time					
	1	0.035	0.035	0.457	0.767
	2	0.124	0.089	4.965	0.030*
	3	0.311	0.187	13.055	0.001*
Turning time					
	1	0.132	0.132	1.898	0.125
	2	0.150	0.018	1.030	0.315
	3	0.224	0.074	4.575	0.038*

1. Predictors: (constant), confounding variables
2. Predictors: (constant), confounding variables, strength
3. Predictors: (constant), confounding variables, strength, balance

*p < 0.05

Table 4.4. Results of model 3 multivariate logistic regression assessing the relation between TUG components' times (STW and Turn duration) and two factors (strength and balance).

Predictors	Beta Coefficient		p-value	
	STW	Turning	STW	Turning
STRENGTH	-0.166	-0.059	0.211	0.671
BALANCE	-0.494	-0.326	0.001	0.038

Discussion

This study examined the association between two age-related functional declines (muscle strength and balance control) and the amount of time taken to perform STW or the turning component of the TUG using both univariate and multivariate regression analyses. Using the univariate analysis, STW time demonstrated a moderate correlation with hip abductor strength. This was expected given that the hip abductor is a key muscle in maintaining single limb support, which is an important component during STW transition.

Both clinical balance measures, BBS and FAB, moderately correlated with STW and turn times. However, our findings indicated that biomechanical balance parameters derived directly from the TUG test correlated better with FAB than BBS. This indicates that FAB could better capture balance deficits associated with tasks included in the TUG than BBS, especially when applied to highly functional elderly adults.

Several biomechanical balance parameters also significantly correlated with STW or turning durations. Shorter STW durations were associated with larger anterior COM-Ankle angles at swing-off. At swing-off, a single leg support phase, an individual needs to manage the transition from a larger base of support (BOS) of seating to a smaller BOS (single leg standing). Participants with larger COM-Ankle angles indicate that they have the ability to better maintain or sustain a greater anterior inclination between the COM and BOS. Shorter STW durations were also associated with larger posterior COM-Ankle angles at stance-off. A smaller posterior COM-Ankle angle indicated that a smaller anterior COM acceleration at stance-off was generated [110]. This implies participants with poor balance control chose a strategy that avoided large forward momentum at stance-off, thus, culminating in a longer STW duration. In addition, frontal plane COM excursion demonstrated a positive correlation with STW duration. Decreased medial-lateral stability during STW has been reported in elderly individuals with a fear of falling [89], and elderly adults with balance impairment were reported to walk with a larger COM sway angle [74]. Our finding also suggests that the ability to efficiently control COM in the frontal plane during this forward movement transition was challenged for people who had a longer STW time.

Shorter turning durations were associated with a larger range of sagittal plane COM-Ankle angles during turning (steps before and after “pivoting”). Similar to STW, a larger range of A-P angles during turning indicates one’s ability to control a larger COM momentum within a step [110]. To begin the turn, forward COM momentum must be reduced [80]. A smaller range of A-P angles before turning suggests an inability to decelerate the forward motion efficiently within a step, or walking with a slower velocity. A smaller range of A-P angles after the turn could suggest an individual’s inability to immediately generate a forward acceleration and restore the waking speed.

The multivariate analyses also partially supported the second part of the hypothesis that strength has a higher association with STW than turning duration. Although strength was not significantly associated with STW and turning durations after controlling for the confounders, a significant F change was detected when strength was added to the regression model for STW duration. This indicated that significantly greater variance in STW duration was explained after muscle strength was included in the model. This phenomenon was not observed in turning duration.

Our findings did not support our hypothesis that balance control is more strongly associated with turning than STW duration. Instead, balance control was found to be significantly associated with both STW and turning durations even after controlling for muscle strength and confounders. Specifically, a person with better balance control performs STW and turning faster. Muscle strength of the lower extremities is essential for balance maintenance and mobility. Reduced muscular strength can significantly impact an older adult's functional ability [29, 32]. Thus, in our regression model, balance was entered after controlling muscle strength. Our results suggest that a smooth STW motion not only requires muscle strength but also relies heavily on one's balance control ability. We argue that it is important for clinicians to further investigate the balance ability of people who have longer STW and turn durations in the TUG.

The average TUG time was below 12 seconds in our study. This suggests that our participants are relatively active individuals [66]. Including more frail elderly participants could potentially expand the data range and yield different results. Also, we were only able to investigate a few of the many factors that have been shown to affect performance of individual tasks included in the TUG. A significant but small R square value indicated that a moderate portion of variance was not explained by the variables examined in this study. Many other factors could influence TUG performance, e.g. health status, use of assistive devices, balance perception, and cognitive impairment [62, 63, 122].

In conclusion, investigating correlations between biomechanical balance parameters and individual task performance in the TUG provided valuable information that aids in explaining possible mechanisms behind a longer STW or turning performance. Our findings suggest that balance control is an important factor that contributes to longer STW and turning durations on the TUG. A person who requires a longer time to complete STW not only tries to avoid a large momentum generation in the forward direction but also has difficulties in controlling frontal plane COM motion. A longer turning duration is associated mainly with forward COM momentum control during the step before and after making the direction change. Therefore, the feasibility of timing each component of the TUG in a clinical setting is worthy of further investigation.

Bridge

Chapter IV examined correlations between underlying impairments (muscle strength and balance control ability) and both TUG component times (STW and turn durations) to demonstrate what factors cause elderly fallers to have longer STW and turn durations. Additionally, correlations found between biomechanical balance parameters and TUG components' times provided valuable information that aids in explaining possible mechanisms behind a longer STW and turn duration. In Chapter V, we investigate the feasibility of using these biomechanical balance parameters to predict prospective falls.

CHAPTER V

BIOMECHANICAL BALANCE PARAMETERS IN FRONTAL PLANE PREDICT PROSPECTIVE FALLS IN ELDERLY ADULTS

The study included in this chapter was developed by number of individuals, including Dr. Li-Shan Chou and Dr. Roland Good. Dr. Chou contributed substantially to this work by participating in the development of methodologies and providing critiques and editing advice. Dr. Good was helpful in providing statistical analysis advice. I was the primary contributor to the development of the protocol, data collection, data analysis and did all the writing.

Introduction

One of the major problems associated with aging is the increased risk of falling. Approximately a third of elderly adults fall annually [1, 2]. Elderly adults with balance and gait impairments are three times more likely to fall than those without such impairments [123]. Exercises with balance and strength training were reported to be effective in reducing fall incidents [23]. The ability of a balance assessment outcome to predict the prospective risk of falling in elderly adults is critical to a timely prescription of preventive interventions.

Several clinical balance assessments, such as the Timed Up and Go test (TUG), Berg Balance Scale (BBS), and Fullerton Advanced Balance scale (FAB), are currently used to evaluate balance control ability and fall risks. Several studies have suggested that the TUG is sensitive to identify elderly adults at risk for falls [59, 66]. Low BBS scores have been shown to predict falls [57, 58]. However, BBS has also found to be less predictive of falls in elderly adults who are active and live independently [57, 124]. To address this limitation, the FAB scale was recently developed; it is a performance-based measure that is specifically designed for use with independently functioning elderly adults. The FAB has moderate sensitivity (74.6%) and lower specificity (52.6%) in predicting faller status [68]. Measuring balance perception is another way to assess balance ability. Research has found that simply asking about elderly adults' own perception of balance could predict risk of fracture from falls [125]. The Activity-Specific Balance Confidence (ABC) scale is a questionnaire designed to measure balance perception in the elderly. ABC has been shown to be able to distinguish between elders at various levels of functional mobility as well as fall risks [126, 127].

Many studies have examined the use of laboratory measurements in predicting fall risks. Hillard et al. (2008) found that frontal plane balance recovery performance and lateral balance stability are significant predictors of prospective falls in community-living elderly adults [98]. Maki et al. (1994) reported that lateral spontaneous-sway amplitude under blindfolded conditions was the single best predictor of future falling risk [99]. Hausdorff et al. (2001) demonstrated that stride time variability during gait was able to predict falls in a one-year prospective study [1]. Studies also examined the ability of both laboratory measures and clinical tests to predict elderly fallers and non-fallers. While some clinical balance tests (BBS and Tinetti balance scale) were not able to predict fallers, the laboratory parameters correlated with future falls [100].

Most falls occur during various forms of locomotion [6]. However, few studies have investigated the ability of laboratory parameters, which are derived from balance control during dynamic activities to predict the occurrence of future falls [1, 98]. Most studies examined balance control during static postural tasks [99, 100, 103] or used clinical balance scales to predict falls [57-59]. Examining the placement of the center of mass (COM) in relation to the center of pressure (COP) or ankle joint of the supporting limb at every instant during walking could quantify gait balance control and detect elderly individuals with balance disorders in a retrospective study design [74, 105]. However, the ability of using gait balance measures to predict the risk of prospective falls is still to be explored.

Therefore, the purpose of this study was to assess the feasibility of using biomechanical measures of gait imbalance (COM-Ankle angles in frontal and sagittal planes) to prospectively predict a fall (or falls) in community-dwelling elderly adults. Several clinical balance tests were also included in the analysis and their abilities to predict falls served as a reference for the biomechanical balance parameters. We hypothesized that the biomechanical parameters would be able to predict future falls in community-dwelling elderly adults. Secondly, we expect that combining the clinical balance and biomechanical balance measures would demonstrate a better prediction of prospective fall incidents than clinical balance measures alone.

Methods

Power analysis was performed for logistic regression analysis using preliminary results of frontal plane COM-ankle inclination angle range during STW from 10 elderly adults (6 non-fallers, 4 fallers). A total of 42 subjects were determined with an odds ratio of 4.856, and $\alpha = 0.05$. Considering an attrition rate of 10 percent over a one-year period, a minimum of 50 subjects were required for this study. Therefore, we recruited sixty elderly adults over the age of 70 years from the community. Inclusion criteria for individuals included: (1) ability to walk without the use of an assistive device; (2) had no history of severe neurological or musculoskeletal deficits that might contribute to gait instability or falls such as amputation, cerebral vascular accident, significant head trauma, or Parkinson's disease; and (3) had no uncorrectable visual impairment, vestibular dysfunction, or dementia. All participants agreed to the experimental procedure approved by the Institutional Review Board and signed consent forms prior to study participation.

This was a one-year longitudinal study. Within this study period, monthly phone check-ups were conducted to document any fall incidents. Elderly adults were then separated into two subgroups based on their fall incidences at the end of the study period. Elderly adults who had reported two or more falls were classified as fallers; the others were classified as non-fallers.

Possible risk factors for falling were recorded and treated as potential confounding variables in the analysis, including age, impaired vision, impaired hearing, cognitive impairment, use of medication, depression, strength, and past fall history. Clinical tests (ABC Scale, TUG, BBS, FAB) and a comprehensive laboratory motion analysis on the TUG test were performed.

Participants performed the TUG while barefoot [61]. They were asked to stand up from the bench, walk three meters, turn around, return to the bench and sit down. The following instruction was provided to all of the subjects: "Please complete the whole task at your comfortable speed, and we will time you." Participants placed both feet on a force plate as they stood up from seating on a bench. Height of the bench was adjusted to each participant's knee height.

Twenty-nine markers were placed on selected bony landmarks of the subject [69]. Whole body motion data were captured with a 10-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA) with a sample rate of 60 Hz and a low pass fourth-order Butterworth filter with an 8 Hz cutoff frequency. Two force plates collected ground reaction forces (AMTI, Watertown, MA) with a sampling rate of 960 Hz. Anthropometric reference data for both sexes were adapted from Dempster [109]. Whole body COM position was calculated as the weighted sum of a 13-segment model [69]. The 13 segments are: head and neck, trunk, pelvis, and right and left segment of upper arms, forearms, thighs, shanks, and feet.

Peak joint moments during the isometric maximal voluntary contraction (MVC) test for bilateral hip abductors, knee extensors, and ankle plantar flexors were determined using the Biodex dynamometer (Biodex Medical Systems, Inc., NY). These peak joint moments were used to indicate the strength for each muscle group. The MVC strength of the hip abductor was measured in a standing position with the hip at neutral position. The testing leg was strapped to the force measuring device with the point of application at the lateral distal end of the femur. Furthermore, in order to maintain the one leg standing position, the subject was allowed to hold the test administrator's shoulders. Subjects then were instructed to abduct the hip against the application pad without rotating the lower extremity or moving the trunk in any plane. The knee extensor isometric strength was measured in a seated position at 60 degrees of knee flexion with respect to the longitudinal axis of the thigh segment. The force measuring device was placed at four fingers width above the bilateral malleolus. Isometric strength of ankle plantar flexor was measured in seated position at 20 degree of knee flexion and neutral ankle position.

While testing each joint, the dynamometer was aligned so that the axis of the lever arm coincided with the joint axis of rotation. Each subject was allowed two sub-maximal practice trials for each joint function. Subjects were instructed to push as hard as they could for a period of 5 seconds. A resting period of one minute or longer as necessary was given between repetitions to avoid fatigue effects on the joints. Data from three trials were collected for each joint function. For each joint, the maximum value from three trials was selected for further analysis. The data from both sides for each joint were then averaged and normalized by body weight (kg).

COM-Ankle angles were calculated as the inclination angles of the line formed by the COM and lateral ankle (malleolus) markers in the sagittal and frontal planes for each frame during TUG. Duration for STW was determined from the movement onset to stance leg toe-off (stance-off). Movement onset was identified as the instant when COM forward position changed [106]. Seat-off was identified as the time when the vertical ground reaction force peaked [84]. Swing leg toe-off (swing-off) and stance-off were identified as the times when the leading foot toe marker and trailing foot moved forward, respectively [106]. The supporting limb that made the first direction change is defined as the "turning limb," and this stance period is termed "pivoting."

We used logistic regression to test the ability of balance tests to predict prospective falls. Fall status was included as the dependent variable after dichotomization (0= non-fallers, 1 = fallers: ≥ 2 falls in one-year follow-up). A fall was defined based on the description from The International Classification of Diseases as “an unexpected event where a person falls to the ground from an upper level or the same level [5].” To select biomechanical balance predictors that could be used to identify fallers, univariate logistic regression analyses were first conducted with fall status as the dependent variable and each of the biomechanical balance predictors as the independent variable. Predictors showing a $p<0.20$ for the Wald-test were included in the multivariate logistic regression analysis [18]. The multivariable logistic regression analysis was then performed in three stages: In stage 1, the potential confounding factors for fall status were entered into model 1. In stage 2, we conducted five model 2 logistic regressions to assess the ability of each clinical balance test and all biomechanical balance parameters to predict falls due to high inter-correlations among all clinical balance tests. For model 2A, model 1 variables plus ABC were entered. For model 2B, model 1 variables plus BBS were entered. For model 2C, model 1 variables plus FAB were entered. For model 2D, model 1 variables plus TUG were entered. For model 2E, model 1 variables plus biomechanical balance predictors were entered. In stage 3, four models were constructed to examine the combined effort of each clinical balance test and all biomechanical balance parameters to predict falls. Model 3A includes model 1 variables plus ABC and all biomechanical balance predictors. Model 3B includes model 1 variables plus BBS and all biomechanical balance predictors. Model 3C includes model 1 variables plus FAB and all biomechanical balance predictors. Model 3D includes model 1 variables plus TUG and all biomechanical balance predictors. A significant level of α equal to or less than .05 was used for all tests. SPSS 19.0 (SPSS Inc., Chicago, IL) was used for all statistical analyses.

<u>Results</u>

Characteristics of all study participants are shown in Table 5.1. Two subjects were excluded due to loss of contact or death. Two other subjects were excluded because they refused to complete the ABC test. In addition, the biomechanical data of one subject was excluded due to poor data quality. Thus, data from these five subjects were not entered in to the regression models. The average age of the 55 participants was 77.5 years. There were more female (67.3%) than male participants in the study. Approximately one-third of the participants had a past fall history with two or more falls. Among all participants, 21.8 % fell two or more times in the one-year study period. The mean TUG time, FAB score and BBS score were 9.5 ± 3.6 seconds, 28.4 ± 7.0 and 53.3 ± 4.6 respectively.

The biomechanical balance predictors that were more highly associated with fall status ($p<0.20$) were entered into model 2E and model 3. These variables were sagittal plane COM-Ankle angle at seat-off, frontal plane COM-Ankle angle during pivoting, and frontal plane COM range of motion during STW (Table 5.2). Vision was not included in the analysis due to the large number of missing values (> 10%) [18]. Results from regression model 1 are presented in Table 5.3. This model was not able to predict fall status with an R square = .253 ($p = 0.197$). No significant predictors were found in model 1 variables.

In models 2A, 2B, 2C and 2D, each individual clinical balance test (ABC, BBS, FAB and TUG, respectively) was added to all confounders in model 1. Only model 2A was significantly related to fall status after controlling for the confounding variables (Table 5.4). When ABC was added to the confounders of model 1, the explained variance

Table 5.1. Subject demographics (N= 55).

	Mean	Std. Deviation
Age (year)	77.5	5.4
Height (cm)	163.4	9.4
Weight (kg)	72.0	18.4
BMI	26.8	6.2
Vision (N=49)	0.65	0.17
Number of medications	4.1	3.3
General depression scale	1.7	2.3
SLUMS Examination (Max score = 30)	26.5	3.8
Timed up and go test (s)	9.5	3.6
Fullerton advanced balance scale (Max score = 40)	28.4	7.0
Berg balance scale (Max score = 56)	53.3	4.6
Activities-specific Balance Confidence scale (Max score = 100)	85.5	16.4
Number of falls	1.2	1.8
	Value	
Female/Male (%)	67.3	N/A
Past fall history (≥ 2 falls) %	30.9	N/A
Impaired hearing (%)	20.0	N/A
Fallers (≥ 2 falls) %	21.8	N/A

was increased by 14.1% (from 25.3 % to 39.4 %). Participants with higher ABC scores are less likely to become fallers. For models 2A to 2D, the sensitivity ranged from 50.0% to 58.3% and specificity ranged from 81.4% to 88.4% with a probability of fall cutoff equal to 0.3. When the biomechanical balance predictors were added to model 1 (model 2E) the explained variance was increased by 24.9% (from 25.3% to 50.2%), and the sensitivity and specificity were 66.7% and 88.4%, respectively (cutoff = 0.3). The frontal plane COM-Ankle angle during pivoting while turning and COM range of motion during STW was significantly associated with fall status after controlling for confounding variables (Table 5.4). Participants with a larger frontal plane COM-Ankle angle during pivoting were less likely to become fallers.

Table 5.2. Results of univariate logistic regression assessing the ability of potential biomechanical balance parameters to predict falls.

Predictors: COM-Ankle angles	Beta Coefficient	p-value	R square	Odds ratio
Sagittal plane				
Seat-off	0.204	0.081	0.088	1.227
Swing-off	0.073	0.610	0.007	1.076
Stance-off	0.063	0.533	0.010	1.065
Before pivoting	-0.057	0.272	0.037	0.945
Pivoting	-0.144	0.394	0.021	0.866
After pivoting	-0.063	0.259	0.038	0.939
Frontal plane				
Swing-off	0.103	0.435	0.016	1.109
Stance-off	0.112	0.391	0.020	1.119
Before pivoting	0.120	0.216	0.040	1.127
Pivoting	-0.226	0.063	0.103	0.798
After pivoting	-0.053	0.564	0.009	0.949
COM range during STW	0.231	0.086	0.079	1.260

Table 5.3. Results of model 1 multivariate logistic regression assessing the ability of potential confounding factors to predict falls (R Square = 0.253).

Predictors	Beta Coefficient	p-value	Odds ratio
Age	-0.081	0.313	0.922
General depression scale	-0.086	0.645	0.918
Past fall history	1.272	0.111	3.567
Number of Medications	0.155	0.182	1.167
SLUMS exam	0-.128	0.276	0.880
Impaired hearing	0.992	0.334	2.696
Strength	-0.367	0.820	0.693

Table 5.4. Results of model 2 multivariate logistic regression assessing the ability of each clinical balance predictor and all biomechanical balance parameters to predict falls.

Predictors	R Square	Beta	Odds ratio	p-value
Model 2A	0.394			0.038
ABC		-0.075	0.928	0.029
Model 2B	0.350			0.077
BBS		-0.282	0.754	0.085
Model 2C	0.268			0.230
FAB		-0.064	0.938	0.421
Model 2D	0.253			0.275
TUG		-0.007	0.993	0.963
Model 2E	0.502			0.017
Sagittal plane COM-Ankle angle (seat-off)		-0.211	0.809	0.269
Frontal plane COM range (STW)		-0.561	0.571	0.026
Frontal plane COM angle (pivoting)		0.650	1.915	0.028

Results of the combined effort of each clinical balance test and all biomechanical balance predictors to predict falls are presented in Table 5.5 (Model 3A, 3B, 3C and 3D). All models 3 were significantly related to fall status with 50.2% to 57.7% of the variance in fall status explained. The sensitivity ranged from 66.7% to 75.0% and the specificity ranged from 88.4% to 93.0% (cutoff = 0.3). In model 3A, only the frontal plane COM-Ankle angle during pivoting was significantly associated with fall status after controlling for confounding variables. In models 3B, 3C and 3D, both the frontal plane COM-Ankle angle during pivoting and the frontal plane COM range of motion during STW were significantly correlated with fall status after controlling for confounding variables.

Table 5.5. Results of model 3 multivariate logistic regression assessing the ability of clinical and biomechanical balance parameters to predict falls.

Predictors	R Square	Beta	Odds ratio	p-value
Model 3A	0.562			0.009
ABC		-0.063	0.939	0.107
Sagittal plane COM-Ankle angle (seat-off)		-0.204	0.815	0.317
Frontal plane COM range (STW)		-0.506	0.603	0.058
Frontal plane COM angle (pivoting)		0.650	1.916	0.047
Model 3B	0.577			0.007
BBS		-0.358	0.699	0.110
Sagittal plane COM-Ankle angle (seat-off)		-0.192	0.825	0.361
Frontal plane COM range (STW)		-0.567	0.567	0.043
Frontal plane COM angle (pivoting)		0.602	1.826	0.026
Model 3C	0.502			0.026
FAB		-0.006	0.994	0.946
Sagittal plane COM-Ankle angle (seat-off)		-0.211	0.810	0.270
Frontal plane COM range (STW)		-0.557	0.573	0.030
Frontal plane COM angle (pivoting)		0.644	1.904	0.035
Model 3D	0.509			0.024
TUG		-0.098	0.907	0.545
Sagittal plane COM-Ankle angle (seat-off)		-0.220	0.802	0.253
Frontal plane COM range (STW)		-0.577	0.561	0.022
Frontal plane COM angle (pivoting)		0.698	2.010	0.028

Discussion

After controlling for demographics, cognition, depression, strength, and past fall history, we found that biomechanical balance parameters measured during TUG were associated with future fall status in our sample of elderly adults. Furthermore, when adding biomechanical parameters to each clinical balance parameter, significantly higher variance in fall status was explained. Specifically, elderly adults who have a greater frontal plane COM range of motion during STW and a smaller frontal plane angle during turning were more likely to become fallers.

Lateral stability under static condition and frontal plane balance recovery performance were reported to be important predictors of prospective falls in community-living elderly adults [98, 99]. Our investigations of balance control during dynamic tasks (STW and turning) aligned with these earlier findings. Additionally, previous retrospective studies have shown that people with balance impairment and past fall histories showed larger frontal plane COM sway when walking [71, 74]. In this study, we found that a larger frontal plane COM range of motion during STW could be a strong fall predictor. In Chapter IV, we also found that frontal plane COM range of motion during STW had a positive correlation with STW duration. In addition, decreased medial-lateral stability during STW has been reported in elderly individuals with a fear of falling [89]. These findings, taken together, suggest that inability to efficiently control COM in the frontal plane during this forward movement transition resulted in a longer STW duration as well as indicated a higher risk for future falls.

The other dynamic biomechanical balance parameter that can significantly predict future fall status is the range of the frontal plane COM-Ankle angle during pivoting of the turn. This parameter measures the separation between COM and base of support (ankle position) in the frontal plane in the single stance phase of the leg making direction change and accounts for each individual’s body height. Our data suggested that individuals having a larger range of COM-Ankle angles are less likely to fall. These current results aligned with the findings reported in Chapter II, a retrospective study design, that the elderly fallers demonstrated a smaller frontal plane COM-Ankle angle during pivoting. One may ask why elderly adults with a higher risk of falling exhibited a larger frontal plane angle during walking and STW but a smaller angle during turning. The ultimate goal of walking or STW is to move forward, during which frontal plane movement is controlled and minimized. Thus, an inability to efficiently confine the frontal plane COM motion within the line of progression could induce gait imbalance. Unlike walking or STW, the goal of turning is to direct the body onto a new direction, which requires an efficient movement generation in the frontal plane. Individuals with a smaller frontal plane COM-Ankle angle at turning seem to avoid separating the COM away from the base of support, which suggests that they prioritize balance maintenance over direction change.

Past studies have investigated extensively the ability to predict falls with the clinical balance measures included in this study, and our results were in good agreement with previous findings. However, the purpose of this study was not to re-examine their predictive abilities; instead we used them as references to evaluate how well the biomechanical balance parameters can serve as fall predictors. In our study, we entered each clinical balance parameter individually into different regression models to avoid mulitcollinearity due to their significant inter-correlations [126].

Among all clinical balance tests included in this study, only ABC demonstrated a significant association with fall status when entered individually into the model. This could be due to several fundamental differences between our and past studies. First, most of previous studies were retrospective investigations while our study involved a prospective design. Second, before adding each clinical balance test into the regression model, we controlled for various confounding variables for falling. Third, our study participants were mainly recruited from the community while some past studies recruited their participants from nursing homes [56, 126].

The strong association between ABC score and future fall status is somewhat unexpected. Various performance-based balance tests and self-report questionnaires are often used in physical therapy practices to gain insights into a patient's balance ability. Performance-based tests and self-reports provide different but complementary information about one's balance ability. Self-reports of balance and health perception are seldom used in physical therapy clinics [128]. Our results indicated that balance perception may provide valuable information about an individual's fall risks beyond what can be explained by demographics, cognition, depression, strength, and past fall history. Also, administrating the questionnaire requires minimal effort from clinicians and takes little time to complete. Therefore, we suggest that obtaining balance perception should be considered routine in fall risk assessments of the elderly.

A prior power analysis was performed to calculate the required sample size; we have confidence that our sample size supports our findings. However, our participants were volunteers from the community, so the implication of our findings to individuals living in a skilled nursing facility would be speculative. In addition to the confounders included in this study, there are many other factors that could be included in fall risk predictions (such as somatosensory function, social status, health status). Lastly, vision was examined but was not included in data analysis due to many missing values. Including one or more confounders could affect the study results. However, our goal was to test the feasibility of assessing prospective fall risk using biomechanical parameters but not to identify all the risk factors for fall.

Despite these limitations, our results indicated that dynamic biomechanical balance parameters could provide valuable information about a participant's future fall risks beyond what can be explained by demographics, cognition, depression, strength, and past fall history. Among all biomechanical parameters investigated, frontal plane balance control parameters appear to be the most significant predictors for future falls. Our findings also provided valuable information for specifying target areas (frontal plane COM controls) for designing clinical interventions to prevent falls.

CHAPTER VI

DISCUSSION AND CONCLUSION

Main Findings

In the first study, we examined the TUG to determine changes in COM motion control related to age and balance impairment. Our findings suggest that elderly adults, especially fallers, have reduced balance control ability in making transitions between tasks during the TUG test. Moreover, both EA and EF (especially EF) prioritized balance maintenance over momentum generation.

During STW transition, when compared to YA, EA showed slight changes in sagittal plane COM motion, but EF demonstrated significant changes. These changes included a smaller forward COM velocity, a smaller A-P COM-Ankle angle and a larger upward kinetic energy ratio at seat-off during STW. Furthermore, the medial-lateral COM motion in EF was perturbed due to their significant reduction in the forward COM velocity. Our study found that the AP COM-Ankle angle is sensitive in detecting changes between EA and EF during STW.

During turning transitions, smaller A-P COM-Ankle angles, taking more steps during the turn and being unable to control large momentum during turning, demonstrated reduced balance control ability in EA and EF. The most challenging part of turning for EA is the transition back to walking after making the turn. Additionally, when compared to EA, EF needed more steps to control their COM momentum and complete the turn.

In the second study, the results obtained from the investigation of movement strategies and joint kinetics during STW showed that EF placed a greater priority on maintaining a stable posture than gait initiation during STW. EF adjusted their foot (ankle) positions such that their COM moved slightly anterior to the ankle. Additionally, EF increased the braking impulse to reduce their forward COM momentum. The delayed onset of propulsive impulse (gait initiation) ensured that a stable upright position is maintained before gait initiation. When the hip and knee extensor moments were reduced in EF, a greater ankle plantar-flexor moment was initiated at seat-off to achieve STW. The increased ankle plantar-flexor moment negatively impacted the horizontal momentum during STW. We speculate that this is a strategy to improve stability.

The third study examined the association between two underlying factors (muscle strength and balance control) and time durations for STW and turning during TUG test. We found that lower extremity muscle strength and several balance parameters moderately correlated with STW and turn durations. More specifically, STW duration demonstrated a moderate correlation with hip abductor strength and a weak correlation with knee extensor strength. Furthermore, investigating correlations between biomechanical balance parameters and STW duration time provided insights to explain possible mechanisms behind a longer STW or turn duration. A person who takes a longer time to complete STW not only tries to avoid generating a large momentum in forward direction, but also has trouble controlling frontal plane COM motion. Longer turn duration is associated mainly with forward COM momentum control during the step before and after making the direction change. Our multiple regression analyses suggested that balance control is an important factor that contributes to longer STW and turn duration of the TUG. Further, lower extremity muscle strength at hip and knee joints has a significantly higher association with STW than turn duration.

In the final study, we found that biomechanical balance parameters measured during the TUG were associated with future fall status in our sample of elderly adults. Moreover, when adding biomechanical parameters to each clinical balance parameter, significantly more variance in fall status could be explained. Specifically, elderly adults who have a larger frontal plane COM range of motion during STW and a smaller frontal plane angle during turning were more likely to become fallers. Our findings indicated that biomechanical balance parameters may provide valuable information about a participant's fall risks beyond what could be explained by demographics, cognition, depression, strength, and past fall history. Among all biomechanical parameters investigated, frontal plane balance control parameters appear to be the most significant predictors for future falls.

Limitations of the Study

Several limitations should be noted form the current study. First, although a prior power analysis was performed to calculate the required sample size, we were not able to obtain a large enough sample size for the second study because not all subjects made clean contact with the force plate during gait initiation. A larger number of subjects would have provided greater power in statistical analysis for the second study. Moreover, our participants were volunteers from the community. The average TUG time for our study participants was below twelve seconds. This suggested that our participants were mostly active populations [66]. Thus, the implication of our findings to individuals living in a skilled nursing facility or older adults with a lower functional ability would be speculative. Future study should consider including a wider spectrum of individuals, which will allow investigators to have more confidence in generalizing the study results to all elderly adults.

Second, the COM-Ankle angle was used to represent the relationship between COM and BOS. Although COM-Ankle angle was able to demonstrate the COM control during the single stance phase, future studies should include investigating the control of COM with respect to BOS during the double stance phase. The COM estimation in our study was based on previously published anthropometric data. These data have been commonly accepted due to a lack of anthropometric information for specific age and body composition. Additionally, past studies investigating errors associated with COM estimation found that changing only the mass distribution proportions has minor effects on the whole body COM parameters during walking.

Third, seat-off was determined using peak vertical ground reaction force, which may not necessarily coincide with the seat-off timing determined by the pressure sensor on the bench. However, it has been reported that there was an excellent correlation between the two seat-off determination methods [84]. Due to the laboratory setting, only one force place was used to detect forces under the supporting limb. Future studies would be benefited by the evaluation of both the loading and unloading limbs during STW.

Finally, we were only able to investigate a few of the many factors that have been shown to affect TUG performance and fall risks. Although our goal was to test the feasibility of assessing fall risk using biomechanical parameters but not to explain as much of the variance as possible in fall risk, we understand that there are many other factors that could influence TUG performance (such as health status, use of assistive devices, balance perception, cognitive impairment) and fall risk predictions (such as somatosensory function, social status, health status). Additionally, impaired vision was measured but was not included in data analysis due to too many missing values. Including one or more additional confounders might affect the study results.

Future Research

This study connects a clinical balance test, the TUG, and the biomechanical balance parameter, COM-Ankle angle, by performing a whole body motion analysis during the TUG. The duration of TUG components was precisely measured with the motion analysis system in our study. In order to apply our results to a clinical setting, the feasibility of timing each of the TUG components using a multi-memory stopwatch in clinical settings needs to be assessed and validated with our results. This could aid clinicians in developing more specific prevention and training strategies.

The second study investigating joint kinetic during STW suggests that weakness in hip muscles and ankle dorsiflexor muscles could possible affect STW performance in EF. However, the joint moments are the net result of muscular, ligamentous forces and moments, and joint reaction force. It cannot determine the functional challenges imposed on a specific muscle. Therefore, a quantitative parameter that is able to evaluate the functional challenge imposed on a specific muscle is needed to better understand the mechanism of increasing the risk of falling in the elderly associated with muscle weakness. Such a quantitative measure could also provide information to the development of strength training intervention programs. We suggest that future study should investigate muscle activation using electromyography during the transitional phases to examine the reasons behind such functional declines in EF.

The third study's results suggest that hip abductor and knee extensor strength are associated higher with STW than turning duration. The effectiveness of strengthening intervention programs to improve STW performance needs further study. Our final study included a one-year prospective study design to investigate the ability of several biomechanical balance parameters to predict fall incidences. Our results revealed valuable information to identify target areas (frontal plane COM controls during STW and turning) for possible clinical interventions to prevent falls. Therefore, we suggest that future studies explore the intervention programs that target frontal plane COM control during these task transitions and examine the effectiveness of such intervention programs. Although we have identified possible biomechanical factors (frontal plane COM-Ankle angles during STW and turning) in fall prediction, we did not retest the subjects to document the age-related changes in these predictors. Therefore, the ability to quantify age-related changes in these specific parameters needs to be investigated further. Finally, research that follows older adults for longer than one year needs to be conducted.

APPENDIX A

ACTIVITIES SPECIFIC BALANCE CONFIDENCE SCALE

For each of the following activities, please indicate your level of self-confidence by choosing a corresponding number from the following rating scale:

0% 10 20 30 40 50 60 70 80 90 100%

no confidence   completely confident

"How confident are you that you will not lose your balance or become unsteady when you…

1. …walk around the house? ____%
2. …walk up or down stairs? ____%
3. …bend over and pick up a slipper from the front of a closet floor ____%
4. …reach for a small can off a shelf at eye level? ____%
5. …stand on your tiptoes and reach for something above your head? ____%
6. …stand on a chair and reach for something? ____%
7. …sweep the floor? ____%
8. …walk outside the house to a car parked in the driveway? ____%
9. …get into or out of a car? ____%
10. …walk across a parking lot to the mall? ____%
11. …walk up or down a ramp? ____%
12. …walk in a crowded mall where people rapidly walk past you? ____%
13. …are bumped into by people as you walk through the mall?____%
14. … step onto or off an escalator while you are holding onto a railing? ____%
15. … step onto or off an escalator while holding onto parcels such that you cannot hold onto the railing? ____%
16. …walk outside on icy sidewalks? ____%

APPENDIX B

BERG BALANCE SCALE

SITTING TO STANDING
INSTRUCTIONS: Please stand up. Try not to use your hand for support.
(   ) 4 able to stand without using hands and stabilize independently
(   ) 3 able to stand independently using hands
(   ) 2 able to stand using hands after several tries
(   ) 1 needs minimal aid to stand or stabilize
(   ) 0 needs moderate or maximal assist to stand

STANDING UNSUPPORTED
INSTRUCTIONS: Please stand for two minutes without holding on.
(   ) 4 able to stand safely for 2 minutes
(   ) 3 able to stand 2 minutes with supervision
(   ) 2 able to stand 30 seconds unsupported
(   ) 1 needs several tries to stand 30 seconds unsupported
(   ) 0 unable to stand 30 seconds unsupported

If a subject is able to stand 2 minutes unsupported, score full points for sitting unsupported. Proceed to item #4.

SITTING WITH BACK UNSUPPORTED BUT FEET SUPPORTED ON FLOOR OR ON A STOOL
INSTRUCTIONS: Please sit with arms folded for 2 minutes.
(   ) 4 able to sit safely and securely for 2 minutes
(   ) 3 able to sit 2 minutes under supervision
(   ) 2 able to able to sit 30 seconds
(   ) 1 able to sit 10 seconds
(   ) 0 unable to sit without support 10 seconds

STANDING TO SITTING
INSTRUCTIONS: Please sit down.
(   ) 4 sits safely with minimal use of hands
(   ) 3 controls descent by using hands
(   ) 2 uses back of legs against chair to control descent
(   ) 1 sits independently but has uncontrolled descent
(   ) 0 needs assist to sit

TRANSFERS

INSTRUCTIONS: Arrange chair(s) for pivot transfer. Ask subject to transfer one way toward a seat with armrests and one way toward a seat without armrests. You may use two chairs (one with and one without armrests) or a bed and a chair.

( ) 4 able to transfer safely with minor use of hands
( ) 3 able to transfer safely definite need of hands
( ) 2 able to transfer with verbal cuing and/or supervision
( ) 1 needs one person to assist
( ) 0 needs two people to assist or supervise to be safe

STANDING UNSUPPORTED WITH EYES CLOSED

INSTRUCTIONS: Please close your eyes and stand still for 10 seconds.

( ) 4 able to stand 10 seconds safely
( ) 3 able to stand 10 seconds with supervision
( ) 2 able to stand 3 seconds
( ) 1 unable to keep eyes closed 3 seconds but stays safely
( ) 0 needs help to keep from falling

STANDING UNSUPPORTED WITH FEET TOGETHER

INSTRUCTIONS: Place your feet together and stand without holding on.

( ) 4 able to place feet together independently and stand 1 minute safely
( ) 3 able to place feet together independently and stand 1 minute with supervision
( ) 2 able to place feet together independently but unable to hold for 30 seconds
( ) 1 needs help to attain position but able to stand 15 seconds feet together
( ) 0 needs help to attain position and unable to hold for 15 seconds

REACHING FORWARD WITH OUTSTRETCHED ARM WHILE STANDING

INSTRUCTIONS: Lift arm to 90 degrees. Stretch out your fingers and reach forward as far as you can. (Examiner places a ruler at the end of fingertips when arm is at 90 degrees. Fingers should not touch the ruler while reaching forward. The recorded measure is the distance forward that the fingers reach while the subject is in the most forward lean position. When possible, ask subject to use both arms when reaching to avoid rotation of the trunk.)

( ) 4 can reach forward confidently 25 cm (10 inches)
( ) 3 can reach forward 12 cm (5 inches)
( ) 2 can reach forward 5 cm (2 inches)
( ) 1 reaches forward but needs supervision
( ) 0 loses balance while trying/requires external support

PICK UP OBJECT FROM THE FLOOR FROM A STANDING POSITION
INSTRUCTIONS: Pick up the shoe/slipper, which is place in front of your feet.
( ) 4 able to pick up slipper safely and easily
( ) 3 able to pick up slipper but needs supervision
( ) 2 unable to pick up but reaches 2-5 cm(1-2 inches) from slipper and keeps balance independently
( ) 1 unable to pick up and needs supervision while trying
( ) 0 unable to try/needs assist to keep from losing balance or falling

TURNING TO LOOK BEHIND OVER LEFT AND RIGHT SHOULDERS WHILE STANDING
INSTRUCTIONS: Turn to look directly behind you over toward the left shoulder. Repeat to the right. Examiner may pick an object to look at directly behind the subject to encourage a better twist turn.
( ) 4 looks behind from both sides and weight shifts well
( ) 3 looks behind one side only other side shows less weight shift
( ) 2 turns sideways only but maintains balance
( ) 1 needs supervision when turning
( ) 0 needs assist to keep from losing balance or falling

TURN 360 DEGREES
INSTRUCTIONS: Turn completely around in a full circle. Pause. Then turn a full circle in the other direction.
( ) 4 able to turn 360 degrees safely in 4 seconds or less
( ) 3 able to turn 360 degrees safely one side only 4 seconds or less
( ) 2 able to turn 360 degrees safely but slowly
( ) 1 needs close supervision or verbal cuing
( ) 0 needs assistance while turning

PLACE ALTERNATE FOOT ON STEP OR STOOL WHILE STANDING UNSUPPORTED
INSTRUCTIONS: Place each foot alternately on the step/stool. Continue until each foot has touch the step/stool four times.
( ) 4 able to stand independently and safely and complete 8 steps in 20 seconds
( ) 3 able to stand independently and complete 8 steps in > 20 seconds
( ) 2 able to complete 4 steps without aid with supervision
( ) 1 able to complete > 2 steps needs minimal assist
( ) 0 needs assistance to keep from falling/unable to try

STANDING UNSUPPORTED ONE FOOT IN FRONT

INSTRUCTIONS: (DEMONSTRATE TO SUBJECT) Place one foot directly in front of the other. If you feel that you cannot place your foot directly in front, try to step far enough ahead that the heel of your forward foot is ahead of the toes of the other foot. (To score 3 points, the length of the step should exceed the length of the other foot and the width of the stance should approximate the subject's normal stride width.)

( ) 4 able to place foot tandem independently and hold 30 seconds
( ) 3 able to place foot ahead independently and hold 30 seconds
( ) 2 able to take small step independently and hold 30 seconds
( ) 1 needs help to step but can hold 15 seconds
( ) 0 loses balance while stepping or standing

STANDING ON ONE LEG

INSTRUCTIONS: Stand on one leg as long as you can without holding on.

( ) 4 able to lift leg independently and hold > 10 seconds
( ) 3 able to lift leg independently and hold 5-10 seconds
( ) 2 able to lift leg independently and hold ≥ 3 seconds
( ) 1 tries to lift leg unable to hold 3 seconds but remains standing independently.
( ) 0 unable to try of needs assist to prevent fall

( ) TOTAL SCORE (Maximum = 56)

APPENDIX C

FULLERTON ADVANCED BALANCE SCALE

Name: Date of Test: ____________

1. Stand with feet together and eyes closed
( ) 0 Unable to obtain the correct standing position independently
( ) 1 Able to obtain the correct standing position independently but unable to maintain the position or keep the eyes closed for more than 10 seconds
( ) 2 Able to maintain the correct standing position with eyes closed for more than 10 seconds but less than 30 seconds
( ) 3 Able to maintain the correct standing position with eyes closed for 30 seconds but requires close supervision
( ) 4 Able to maintain the correct standing position safely with eyes closed for 30 seconds

2. Reach forward with outstretched arm to retrieve an object (pencil) held at shoulder height
( ) 0 Unable to reach the pencil without taking more than two steps
( ) 1 Able to reach the pencil but needs to take two steps
( ) 2 Able to reach the pencil but needs to take one step
( ) 3 Can reach the pencil without moving the feet but requires supervision
( ) 4 Can reach the pencil safely and independently without moving the feet

3. Turn 360 degrees in right and left directions
( ) 0 Needs manual assistance while turning
( ) 1 Needs close supervision or verbal cueing while turning
( ) 2 Able to turn 360 degrees but takes more than four steps in both directions
( ) 3 Able to turn 360 degrees but unable to complete in four steps or fewer in one direction
( ) 4 Able to turn 360 degrees safely taking four steps or fewer in both directions

4. Step up onto and over a 6-inch (15 cm) bench
( ) 0 Unable to step up onto the bench without loss of balance or manual assistance
( ) 1 Able to step up onto the bench with leading leg but trailing leg contacts the bench or swings around the bench during the swing-through phase in both directions
( ) 2 Able to step up onto the bench with leading leg, but trailing leg contacts the bench or swings around the bench during the swing-through phase in one direction
( ) 3 Able to correctly complete the step up and over in both directions but requires close supervision in one or both directions
( ) 4 Able to correctly complete the step up and over in both directions safely and independently

5. Tandem walk
( ) 0 Unable to complete 10 steps independently
( ) 1 Able to complete the 10 steps with more than five interruptions
( ) 2 Able to complete the 10 steps with three to five interruptions
( ) 3 Able to complete the 10 steps with one to two interruptions
( ) 4 Able to complete the 10 steps independently and with no interruptions

6. Stand on one leg
( ) 0 Unable to try or needs assistance to prevent falling
( ) 1 Able to lift leg independently but unable to maintain position for more than 5 seconds
( ) 2 Able to lift leg independently and maintain position for more than 5 but less than or equal to 12 seconds
( ) 3 Able to lift leg independently and maintain position for more than 12 but less than 20 seconds
( ) 4 Able to lift leg independently and maintain position for the full 20 seconds

7. Stand on foam with eyes closed
( ) 0 Unable to step onto foam or maintain standing position independently with eyes open
( ) 1 Able to step onto foam independently and maintain standing position but unable or unwilling to close eyes
( ) 2 Able to step onto foam independently and maintain standing position with eyes closed for 10 seconds or less
( ) 3 Able to step onto foam independently and maintain standing position with eyes closed for more than 10 seconds but less than 20 seconds
( ) 4 Able to step onto foam independently and maintain standing position with eyes closed for 20 seconds

Do not perform test item 8 if score is 2 or lower on test item 4. Also do not introduce test item 8 if test item 4 was not performed safely and/or it is contraindicated to perform this test-item (review test administration instructions for contraindications). Give test item 8 a score of 0 and proceed to test item 9.

8. Two-footed jump
( ) 0 Unable to attempt or attempts to initiate jump but one or both feet do not leave the floor
( ) 1 Able to initiate jump with both feet but one foot either leaves the floor or lands before the other
( ) 2 Able to perform jump with both feet but unable to jump farther than the length of feet
( ) 3 Able to perform jump with both feet and achieve a distance greater than the length of feet
( ) 4 Able to perform jump with both feet and achieve a distance greater than twice the length of feet Revised Aug 2011

9. Walk with head turns
( ) 0 Unable to walk 10 steps independently while maintaining 30o head turns at an established pace
( ) 1 Able to walk 10 steps independently but unable to complete required number of 30o head turns at an established pace
( ) 2 Able to walk 10 steps but veers from a straight line while performing 30o head turns at an established pace
( ) 3 Able to walk 10 steps in a straight line while performing 30o head turns at an established pace but head turns less than 30o in one or both directions
( ) 4 Able to walk 10 steps in a straight line while performing required number of 30o head turns at established pace

10. Demonstrate reactive postural control
( ) 0 Unable to maintain upright balance; makes no observable attempt to step; requires manual assistance to restore balance
( ) 1 Unable to maintain upright balance; takes two or more steps and requires manual assistance to restore balance
( ) 2 Unable to maintain upright balance; takes more than two steps but is able to restore balance independently
( ) 3 Unable to maintain upright balance; takes two steps but is able to restore balance independently
( ) 4 Unable to maintain upright balance but able to restore balance independently with only one step

TOTAL POINTS SCORED: _______________
40 POINTS POSSIBLE MAXIMUM SCORE

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