

DYNAMIC LIMITS OF BALANCE CONTROL DURING DAILY FUNCTIONAL  
ACTIVITIES ASSOCIATED WITH FALLING

by

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## DISSERTATION ABSTRACT

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Doctor of Philosophy

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Title: Dynamic Limits of Balance Control during Daily Functional Activities Associated with Falling

Falls are one of the most serious problems among the elderly, resulting in fatal physical injuries. Early identification of people at a high risk of falling is needed to facilitate rehabilitation to reduce future fall risk. The overall goal of this dissertation was to develop biomechanical models that identify dynamic limits of balance control in daily functional activities associated with falling, including sit-to-stand (STS) movement, standing (stance perturbation), and walking.

Poor performance of STS movement has been identified as one of the risk factors of falls among elderly individuals. We proposed a novel method to identify dynamic limits of balance control during STS movement using whole body center of mass (COM) acceleration and assessed its feasibility to differentiate individuals with difficulty in STS movement from healthy individuals. The results demonstrated that our model with COM acceleration could better differentiate individuals with difficulty in STS movement from healthy individuals than the traditional model with COM velocity.

Poor postural control ability is also a risk factor of falls. Postural recovery responses to backward support surface translations during quiet standing were examined for healthy young and elderly adults. The results demonstrated that functional base of

support (FBOS) and ankle dorsiflexor strength could be sensitive measures to detect elderly individuals with declined balance control. Our biomechanical model, which determines a set of balance stability boundaries, showed a better predictive capability than the statistical model for identifying unstable balance recovery trials, while the statistical model better predicted stable recovery trials.

Lastly, walking requires a fine momentum control where COM acceleration could play an important role. Differences in control of dynamic stability during walking were examined with our proposed boundaries of dynamic stability. Elderly fallers adapted a more conservative gait strategy than healthy individuals, demonstrating significantly slower forward COM velocity and acceleration with their COM significantly closer to the base of support at toe-off, which could be indicative of a poor momentum control ability.

Overall, this study demonstrated that COM acceleration would provide further information on momentum control, which could better reveal underlying mechanisms causing imbalance and provide an insightful evaluation of balance dysfunction.

This dissertation includes unpublished co-authored material.

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CHAPTER I  
GENERAL INTRODUCTION

Functional Daily Activities Associated with Falling

Falls are one of the most serious problems among the elderly due to age-related declines in balance control, resulting in fatal physical injuries [1-3]. The prevalence rate of fallers is at 33% after 65 years and reaches 50% after 80-85 years [4, 5]. Early identification of people at a high risk of falling is needed to facilitate the provision of rehabilitation treatment to reduce future fall risk [6].

Sit-to-stand (STS) movement is one of the most commonly performed functional activities in daily life [7, 8]. STS movement can be more biomechanically demanding than other activities, which requires more leg strength and greater joint ranges of motion than walking or stair climbing [9, 10] and is often associated with falling [11-15]. Poor performance of STS movement has been identified as one of the risk factors that may be predictive of the likelihood of falls among elderly individuals [2, 16, 17]. Difficulty in performing STS movement is commonly reported in older adults [18-21], and increased odds of falls have been found for elderly individuals having difficulty in STS movement [2, 22]. Multiple fallers perform significantly worse than non-multiple fallers in a STS task [23]. A better understanding of dynamic balance control of individuals with difficulty in STS movement would be helpful in developing effective methods to identify elderly individuals at a risk of falling and to understand the mechanisms underlying the increased incidence of falls in the elderly.



Poor postural control is also one of the risk factors in the increased incidence of falls in the elderly, as indicated by a significant increase in body sway even in healthy older adults when compared with young adults, with the greatest amount of sway in the elderly with a history of recent falls [24]. Research studies have also shown that functional stability boundaries, defined as the excursion of center of pressure (COP) trajectories during maximum sway efforts, decrease with age [25, 26]. During quiet standing, the COP is regulated by the neural system to keep the whole body center of mass (COM) within a desired position to achieve stable upright standing. This suggests that the elderly have a reduced allowable area to effectively displace the COP to control the COM movement. In addition to studies measuring spontaneous sway of the body during quiet standing, responses to unexpected perturbations during standing, such as support surface translations, have been commonly used to examine dynamic postural control ability [27-31]. It has been reported that elderly individuals take a compensatory step more often than young adults with less severe balance perturbations [32]. These findings suggest that older adults are at higher risk of instability that would necessitate taking a step or risking a fall; however, causes of the increased rate of stepping in the elderly still remain undetermined. Examining postural recovery responses to perturbations would help us to assess dynamic balance control ability of the individual, which would be useful for early identification of individuals at a higher risk of falling.

Finally, most falls in the elderly resulting in fatal physical injuries occur while walking [33]. Age-related changes in the neural and musculoskeletal systems would contribute to such declined gait performance in the elderly. Kinematic analysis revealed that elderly showed a slower walking speed, shorter stride length, smaller vertical

movement of the head, larger lateral movement of the head, wider stride width, greater toeing out, longer stance phase, shorter swing phase, smaller hip, knee and ankle flexion. Kinetic analysis showed that plantar flexors of elderly adults generated less power at push-off, which could explain the shorter step length, and increased stance phase. This reduced push-off could be an adaptive change used to ensure a safer gait [34]. Those age-related changes in gait function could result in changes in COM motion. The COM motion has been shown to differentiate elderly people who are at a higher risk of falling from healthy individuals. Elderly patients with balance impairment demonstrated greater and faster medio-lateral motion of the COM during walking, which was more pronounced when crossing an obstacle [35-37]. Those results suggest that examining COM motion during walking would allow us to identify individuals with balance impairment, which would facilitate designing risk assessments for fall prevention.

Overall, these daily functional activities associated with falling, including STS movement, standing/stance perturbation, and walking, involve the displacement of the COM. Balance maintenance during these movements requires a proper dynamic control of the COM in relation to the base of support (BOS) confined by supporting feet. An inability to properly control the COM due to age-related declines in the neural and musculoskeletal systems contributes to an increased likelihood for imbalance or falls in the elderly.

#### Traditional and Novel Approaches to Assess Dynamic Balance Control

Motion of the whole body COM is passively regulated within a desired region to achieve postural stability by the neural and musculoskeletal systems through an active

control of the COP displacement. Biomechanical interactions between the COM motion and COP position provide information about the ability to maintain balance, and thus are used to assess dynamic balance control [38, 39].

COM motion and its relative position to the COP have been used to examine balance control during gait [38, 39]. Linear measures of COM motion in the frontal plane during obstacle crossing [35-37] and narrow base walking task [40] have been used to examine age-related declines in frontal plane stability. Elderly with imbalance showed greater and faster lateral COM motion when crossing over obstacles, although they did not have differences in temporal-distance gait parameters. Lee and Chou (2006) further proposed COM-COP inclination angles, which account for the instantaneous COM height as well as horizontal COM-COP separation distance to exclude inter subject variability in their height since magnitudes of those linear measures of the COM motion and the COM-COP separation distance may be affected by a subject's posture. They demonstrated that the COM-COP inclination angle was a sensitive measure of balance control during gait with a smaller angle indicating better balance performance [41, 42].

In addition, since the BOS is defined as the possible range of the COP movement, the occurrence of imbalance has been traditionally regarded as a consequence of the COM movement beyond the boundaries of the BOS [32]. The relationship between COM and BOS, as limits of COP control, has been investigated to assess dynamic stability during the movement. Steps could occur in anticipation of an impending collision or fall, or in reaction to imposed horizontal movement of the COM with respect to the BOS [30]. A protective step would be engaged when the COM travels beyond the boundaries of the BOS. However, this COM displacement-dependent view presents mostly a static model

of standing balance, and it does not guarantee that stability during a movement will be maintained [30].

In dynamic conditions involving large displacement of the COM, the horizontal velocity of the COM also needs to be considered when describing the feasible movements for the control of balance. Even with a location inside the BOS, balance might not be maintained if the COM has a sufficiently large horizontal velocity [27, 29, 30, 43]. Conversely, it is possible for the COM to be located outside the BOS without initiating a fall if sufficient horizontal COM velocity directed toward the BOS exists. Thus, besides the horizontal location of the COM with respect to the BOS, the magnitude and the direction of its corresponding velocity may also provide critical information pertaining to one's ability to control balance [43].

Pai and Patton (1997) determined a set of feasible COM velocity-position combinations that guarantee stability in dynamic conditions based on a single-link-plus-foot inverted pendulum model, which has a mass point at the COM and has a center of rotation at the ankle joint. They estimated the limiting COM velocities in a given COM position that would permit successful termination of the movement within the BOS in the sagittal plane. A backward balance loss would occur if an initial forward momentum is insufficient to carry the COM within the BOS, while an excessive forward momentum would result in a forward loss of balance [43-45]. Yang et al. extended this concept to the frontal plane motion and described medio-lateral momentum during gait [45]. They also predicted the threshold of the COM velocity relative to the BOS required to prevent backward balance loss during single stance recovery from a slip during gait [46]. Hof and colleagues (2005) also suggested the importance of COM velocity to assess dynamic

stability and derived the extrapolated center of mass (XcoM) to quantify stability using a simple inverted pendulum model, where the condition of dynamic stability was that the XcoM should be confined within the BOS [47, 48]. Time-to-Contact, which uses the instantaneous kinematics, i.e., position and velocity of the COM, to predict a future time at which the COM will contact the BOS boundary, has also been reported to characterize dynamic stability [49]. Those studies suggest that dynamic stability is achieved as a function of both the COM position and velocity.

However, COM velocity, which reflects momentum, only describes an instantaneous state of motion, not providing how the momentum is regulated by skeletal muscles. Even though the resultant velocities are similar, different accelerations could be possible, depending on how the joint torque is generated. It is commonly reported that muscle strength declines with aging [18, 50-53], and elderly adults have difficulties in generating muscle torques at a higher rate [10, 54, 55]. Considering that acceleration results from joint torques produced by skeletal muscles, COM acceleration would directly reflect their momentum control ability, and elderly adults seem to have less ability to regulate momentum by acceleration due to age-related decline in their muscle function, which would result in imbalance during the movement. Examining COM acceleration would provide further insights into balance control during the movement, which would help us to better differentiate individuals with a higher risk of falling from healthy individuals.

### Purpose of the Study and Specific Aims

While COM velocity describes an instantaneous state of motion, in terms of momentum, COM acceleration could provide further insights into how the momentum is regulated by skeletal muscles. An examination of COM acceleration, in addition to its velocity, could help us understand how balance is controlled during the movement, which could better reveal underlying mechanisms causing imbalance and provide an insightful evaluation of balance dysfunction.

The overall goal of this dissertation study was to develop biomechanical models focusing on COM acceleration that identify dynamic limits of balance control in daily functional activities associated with falling, including STS movement, standing/stance perturbation, and walking.

As one of the most performed functional activities, sit-to-stand (STS) movement from a seated position is often associated with falling [11-15]. Poor performance of STS movement has been identified as one of the risk factors that may be predictive of the likelihood of falls among elderly individuals [2, 16, 17]. Identifying individuals with difficulty in STS would allow us to detect individuals who have a higher risk of falling.

Poor postural control ability is also one of the risk factors in the increased incidences of falls in the elderly [24]. Responses to unexpected perturbations during standing, such as support surface translations, have been commonly used to examine dynamic postural control ability [27-31]. Examining postural recovery responses to perturbations would help us to assess balance control ability of the individual, which would be useful for early identification of individuals at a higher risk of falling.

Most falls resulting in fatal physical injuries occur while walking [33]. Walking requires fine momentum control. Considering that muscle strength typically decreases with age, the elderly seem to have less ability to regulate momentum by acceleration, which would result in imbalance during the movement. Investigating how the momentum is controlled during walking would allow us to identify potential fallers.

Accordingly, three studies along with the following specific aims were proposed in this dissertation study:

*Study 1. Identifying Individuals with Difficulty in Sit-to-Stand Movement Using Regions of Stability*

COM control during STS movement for healthy young and elderly adults, and elderly individuals with difficulty in STS were compared using regions of stability determined based on the COM position at seat-off and its instantaneous velocity (ROS<sub>v</sub>) or its peak acceleration (ROS<sub>a</sub>). Three specific aims were proposed in this study:

1. Establish a region of stability using COM acceleration, which identifies dynamic limits of balance control during STS movement.
2. Assess its feasibility to differentiate individuals with difficulty in STS movement from healthy individuals.
3. Test the hypothesis that age- and strategy-related differences in COM control could be more clearly distinguished with its acceleration as compared to velocity, and thereby the ROS<sub>a</sub> would better differentiate individuals with difficulty in STS movement from healthy individuals.

*Study 2. Identifying Biomechanical Factors Affecting Postural Recovery Responses and Acceleration Threshold Prediction during a Backward Stance Perturbation*

Postural recovery responses during backward support surface translations for healthy young and elderly adults were examined. The relationships between ankle muscle strength and functional BOS (FBOS), the effective limits of the BOS, as well as perturbation acceleration threshold to maintain balance were investigated. A biomechanical model that predicts perturbation acceleration threshold to maintain balance was developed. Four specific aims were proposed in this study:

1. Examine the association of ankle muscle strength, FBOS, or perturbation acceleration with the ability to recover balance.
2. Test the hypothesis that FBOS would be a sensitive measure to predict threshold perturbation acceleration for taking a step or heel-rise, and ankle plantarflexor strength would be a significant predictor of FBOS as well as threshold perturbation acceleration.
3. Establish a biomechanical model that predicts threshold perturbation acceleration to maintain upright standing.
4. Compare the predictive capability of the biomechanical model with that of a statistical model.

*Study 3. Assessment of Dynamic Momentum Control during Walking Using Regions of Stability*

Differences in control of dynamic stability during walking for healthy young and elderly adults, and elderly fallers were compared using regions of stability determined



based on the COM position at toe-off and its instantaneous velocity (ROS<sub>v</sub>) or its peak acceleration (ROS<sub>a</sub>). Three specific aims were proposed in this study:

1. Establish a region of stability using COM acceleration, which identifies dynamic limits of balance control during walking.
2. Examine age- and strategy-related differences in dynamic momentum control in the antero-posterior and medio-lateral directions during walking among healthy young and elderly adults, and elderly fallers.
3. Test the hypothesis that elderly fallers would demonstrate a more conservative gait strategy in terms of their momentum control.

#### Flow of the Dissertation

This dissertation includes previously unpublished co-authored material. Chapter II through V correspond to Study 1 to 3 stated in the previous section.

Chapter II addresses the specific aims in Study 1. Differences in COM control during STS movement among healthy young and elderly adults, and elderly individuals with difficulty in STS were investigated. Feasibility of regions of stability to differentiate individuals with difficulty in STS movement from healthy individuals was also assessed. This chapter includes co-authored material with Dr. Li-Shan Chou.

Chapter III addresses the specific aims 1&2 in Study 2. Postural recovery responses during backward support surface translations for healthy young and elderly adults were examined. The relationship between ankle muscle strength and functional BOS (FBOS), the effective limits of the BOS, as well as perturbation acceleration

threshold to maintain balance was investigated. This chapter is intended to be published with co-authors, Dr. Wei-Li Hsu, Dr. Marjorie Woollacott, and Dr. Li-Shan Chou.

Chapter IV addresses the specific aims 3&4 in Study 2. A biomechanical model which predicts perturbation acceleration threshold to maintain balance was developed and its predictive capability was compared with that of a statistical model. This chapter is intended to be published with co-authors, Dr. Marjorie Woollacott, and Dr. Li-Shan Chou.

Chapter V addresses the specific aims in Study 3. Differences in control of dynamic stability during walking for healthy young and elderly adults, and elderly fallers were compared using regions of stability. This chapter is intended to be published with a co-author, Dr. Li-Shan Chou.

Finally, a concluding summary is provided in Chapter VI, with conclusions drawn from the major findings of each study. Limitations of the studies are discussed and suggestions are made for future research. Appendices are provided after the Bibliography, showing detailed derivations of the equations used in each study.

CHAPTER II  
REGION OF STABILITY DERIVED BY CENTER OF MASS ACCELERATION  
BETTER DIFFERENTIATES INDIVIDUALS WITH DIFFICULTY IN SIT-TO-  
STAND MOVEMENT

The study described in this chapter was developed with Dr. Li-Shan Chou. Dr. Li-Shan Chou contributed substantially to the work by providing critiques about data analysis and development of methodologies. I was the primary contributor to the data collections, data analysis, model development, implementation of the procedure, and did all the writing.

Introduction

Early identification of people at a high risk of falling is needed to facilitate the provision of rehabilitation treatment to reduce future fall risk [6]. Falls most frequently occur during daily functional activities involving the displacement of the whole body center of mass (COM), such as walking, sit-to-stand (STS), and stair ascent or descent [2]. Standing up from a seated position could be biomechanically more demanding than walking or stair climbing, as it requires greater muscular efforts and greater joint ranges of motion [9, 10]. Difficulty in performing STS movement is commonly observed and associated with falls in older adults [18-21], being identified as one of the risk factors that may be predictive of the likelihood of falls [2, 16, 17]. Increased odds of falls have been found for elderly individuals having difficulty in STS movement [2, 22], and multiple fallers performed significantly worse than non-multiple fallers in performance of the STS

task [23]. Therefore, understanding biomechanical differences in balance control of older adults with difficulty in STS would enhance our identification of elderly fallers.

Different movement strategies are used to achieve a STS task, depending on the balance control ability of the individual. These strategies can be classified by the horizontal COM momentum and its location in relation to the base of support (BOS). At the time of transfer from the chair (seat-off), the body is often in a statically unstable position, with the COM located posterior to the heel and outside of the BOS [18, 56]. However, an adequately generated COM horizontal momentum allows a successful completion of the STS movement [16]. This strategy (momentum-transfer) requires a precise control of momentum generation to confine the COM motion within the BOS and avoid imbalance [11]. Another strategy (zero-momentum/stabilization) is moving the COM into the BOS prior to seat-off, such as bending the trunk forward. This strategy allows a direct transition from a three-point BOS to the two-point feet support position, and imposes minimal perturbation to balance control [11]. Because the body remains inherently stable at seat-off with the COM brought over the area of the BOS [57], individuals who have balance problems might prefer this strategy. Thus, horizontal COM position and its velocity observed during the STS task could directly reflect their balance control ability.

An inverted pendulum model (IPM) in the sagittal plane has been used to investigate the COM control during STS movement [43, 58, 59], as well as to assess an individual's ability to recover balance from an induced slip during STS movement [60, 61]. A feasible stability region (FSR) was derived to show all feasible combinations of horizontal COM velocities and positions that allow a successful movement termination

[43, 62]. Similar to the FSR, Mourey and colleagues [63] defined the dynamic equilibrium area (DEA) with the inverted pendulum model and compared the COM control between young and elderly during STS. Examination of COM velocity and position in relation to this region would allow us to distinguish the balance control strategy selected by the individual to perform the STS task. However, it is still not clear how the COM momentum is controlled during STS, and whether elderly adults with difficulty in STS movement would differ from healthy adults in such control.

It is possible that the same COM velocities could be achieved with different accelerations, and inability to properly generate COM acceleration would lead to a poor momentum control and result in imbalance. Even though the resultant velocities are similar, different accelerations could be possible, depending on how the joint torque is generated. It is commonly reported that muscle strength declines with aging [18, 50-53], and elderly adults have difficulties in generating muscle torques at a higher rate [10, 54, 55]. Considering body movements are modulated by acceleration resulting from muscle forces, elderly adults would demonstrate a greater difficulty in movement control. Examining COM acceleration, in addition to its velocity, could provide further insights into how balance is controlled. Understanding how the COM motion is regulated could better reveal underlying mechanisms causing imbalance and enhance the development of interventions for fall prevention.

The objectives of this study were, therefore, (1) to establish a region of stability using COM acceleration, which identifies dynamic limits of balance control during STS movement, and (2) to assess its feasibility to differentiate individuals with difficulty in STS movement from healthy individuals. It was hypothesized that age- and strategy-

related differences in COM control could be more clearly distinguished with its acceleration as compared to velocity, and thereby the ROSa would better differentiate individuals with difficulty in STS movement from healthy individuals.

## Methods

### *Regions of Stability*

The region of stability was derived in two ways: one used COM velocity (ROSV), and the other used COM acceleration (ROSa). A single-link-plus-foot IPM in the sagittal plane was used (Fig.2.1). Detailed derivation is presented in Appendix A. Boundaries of the ROSv were defined using the following equation:

$$-\tilde{X}_s \leq \tilde{\dot{X}}_s \leq 1 - \tilde{X}_s \quad (2-1)$$

where  $\tilde{X}_s$  and  $\tilde{\dot{X}}_s$  are normalized COM position and velocity at seat-off (the time when the body is lifted off the seat of a chair), defined as  $\tilde{X}_s = (X_s - X_h) / L_f$ ,  $\tilde{\dot{X}}_s = \dot{X}_s / (L_f \omega_0)$  ( $\omega_0 = \sqrt{g/l}$ ,  $L_f = X_t - X_h$ : foot length,  $X_h$  and  $X_t$ : the heel and toe positions,  $l$ : pendulum length). For the COM position at seat-off, this relationship provides the allowable range of COM velocities that permits a successful STS movement termination.

To derive boundaries of the ROSa, the shape of COM acceleration function was modeled as a triangle-shape positive acceleration followed by a triangle-shape peak negative acceleration (Fig.2.1), which was observed from our experimental data (Fig.2.2). Furthermore, the seat-off was assumed to take place at two different instants: (a) immediately prior to the generation of negative acceleration (Fig.2.1a), and (b) when the negative acceleration reaches its peak (Fig.2.1b). Boundaries of the ROSa with the seat-off instant (a) [ROSa(a)] could be defined as:

$$-\frac{X_h - X_s}{X_s - X_i} \tilde{X}_s \leq \tilde{X}_p \leq \frac{X_t - X_s}{X_s - X_i} (1 - \tilde{X}_s) \quad (2-2)$$

where  $\tilde{X}_p$  is the normalized peak COM acceleration, defined as  $\tilde{X}_p = A\ddot{X}_p / L_f$  ( $A = l/g$ ), and  $X_i$  is the initial COM position.

Similarly, boundaries of the ROSa with the seat-off instant (b) [ROSa(b)] were defined as:

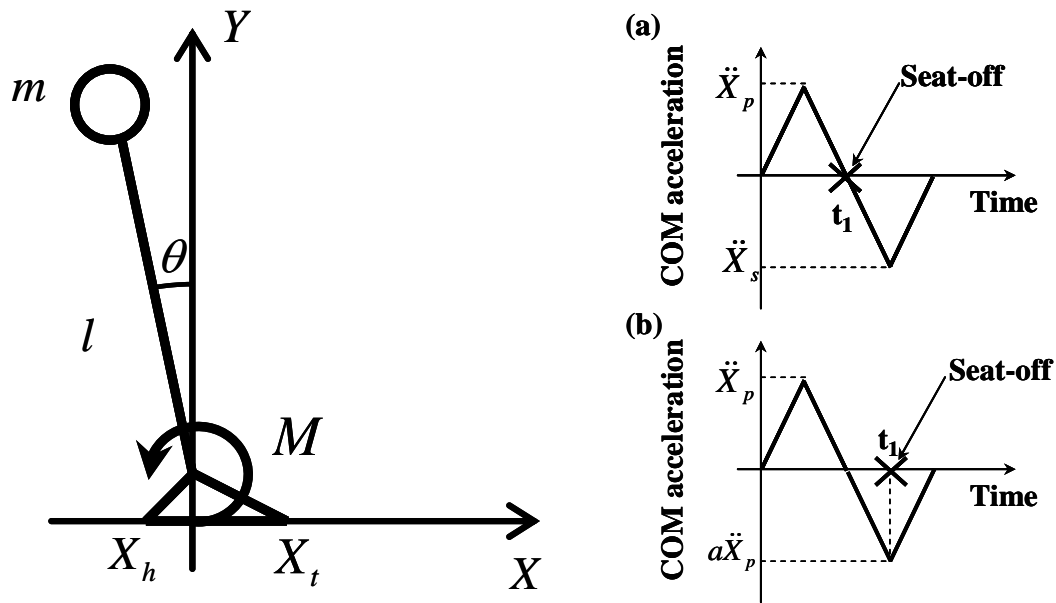
$$-G \frac{X_h - X_s}{X_s - X_i} \tilde{X}_s \leq \tilde{X}_p \leq G \frac{X_t - X_s}{X_s - X_i} (1 - \tilde{X}_s) \quad (2-3)$$

where  $G = \{6 + a(6 - a^2)\} / (2 - a^2)^2$ ,  $a$ : the ratio of the peak negative acceleration to the positive acceleration prior to seat-off, which was determined from each experimental trial (overall average:  $a=0.70$ ). The ROSa provides us the peak COM acceleration being generated prior to seat-off that allows a successful STS movement termination with a given COM position at seat-off.

In addition, the magnitude of the stability was defined as the shortest distance from the experimental data to the forward boundary of the ROSv and ROSa (Stability margin) [45].

### *Experimental Protocol*

Ten healthy young adults [Young: 6 men/4 women; mean age 22.2 years (SD 1.9), mean height 170.9 cm (SD 9.5), mean mass 67.9 kg (SD 11.3)], 10 healthy elderly adults [Elderly: 5 men/5 women; mean age 70.6 years (SD 3.4), mean height 172.8 cm (SD 8.4), mean mass 74.6 kg (SD 14.7)], and 10 elderly adults having difficulty in STS [Elderly (DIFF): 5 men/5 women; mean age 73.6 years (SD 5.1), mean height 167.8 cm (SD 8.1), mean mass 88.6 kg (SD 21.2)] participated in this study. Difficulty in STS were



**Figure 2.1.** A single-link-plus-foot inverted pendulum model in the sagittal plane and two horizontal COM acceleration scenarios included in the analysis.  $X$  indicates the COM position in the antero-posterior direction.  $X_h$  and  $X_t$  indicate the heel and toe positions.  $m$ ,  $l$  and  $M$  are whole body mass, pendulum length (distance from the ankle to the COM), and ankle joint moment. The average value of pendulum lengths between seat-off and standing was used as the pendulum length,  $l$ . The COM acceleration was modeled as a triangle-shape positive acceleration followed by a triangle-shape peak negative acceleration with the seat-off taking place at two different instants: (a) immediately prior to the generation of negative acceleration, and (b) when the negative acceleration reaches its peak ( $a\ddot{X}_p$ ).  $a$ : the ratio of the peak negative acceleration to the positive acceleration prior to seat-off, which was determined for each experimental trial (overall average:  $a = 0.70$ ).

defined as requiring a longer time to perform 5 repetitions of STS task than the following durations for each of the age groups: 11.4 sec (60-69 years), 12.6 sec (70-79 years), and 14.8 sec (80-89 years) [64]. All participants did not have a history or clinical evidence of neurological, musculoskeletal or other medical conditions. The experimental protocol was approved by the Institutional Review Board. Written and verbal instructions of testing procedures were provided, and written consent was obtained from each subject prior to testing.



Subjects were instructed to stand up from a chair adjusted to their knee height (floor to the head of the fibula) at a self-selected speed while barefoot. They were instructed to sit on the chair with trunk positioned vertically with self-selected foot positions. Several practice trials were performed to find their comfortable starting position. The starting position and arm movement were not constrained to allow a natural STS movement, as constrained protocols could limit practical relevance and often preclude the use of different strategies [11]. Data from 6 trials were collected for each subject. It has been reported that elderly adults with functional limitations, such as muscle weakness, would prefer the strategy with exaggerated trunk flexion prior to seat-off [18, 20, 65-68]. Therefore, after completing the STS in a self-selected manner [Young (Norm)], young subjects were asked to perform another block of six STS trials with their trunk purposely bent prior to seat-off, which would require them to use a different strategy [Young (Bend)]. Therefore, there were 1 group with 2 conditions [Young (Norm) and Young (Bend)] and 2 groups [Elderly and Elderly (DIFF)].

Whole body motion was captured with an eight-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA). A total of 29 reflective markers were placed on the subject [37]. Three-dimensional marker trajectories were collected at 60Hz and smoothed using a fourth-order Butterworth filter with a cut-off frequency of 8Hz. Ground reaction forces (GRFs) were collected by two force plates located under each foot (Advanced Mechanical Technology Inc., MA) at 960 Hz. The force data were time-synchronized to the video sampling. The whole-body COM was calculated as the weighted sum of 13 body segments, including head and neck, trunk, pelvis, 2 upper arms, 2 forearms with hands, 2 thighs, 2 shanks, and 2 feet [37]. Anthropometric reference data

were adopted from the initial work of Dempster [34]. Seat-off was determined as the instant when the vertical GRF reached its peak [69, 70].

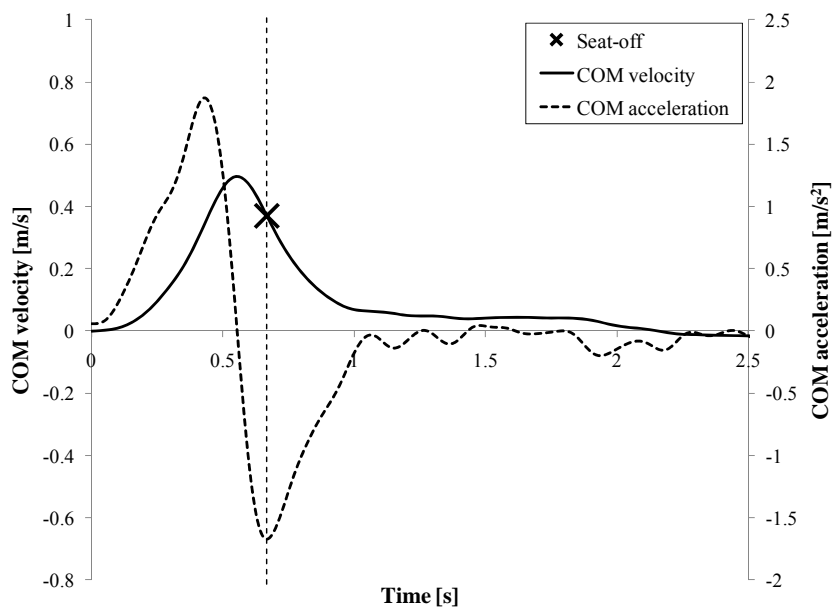
The COM position was referenced to the average heel position and normalized to the foot length (average of both feet). Normalized horizontal COM position and velocity at seat-off, and normalized peak horizontal COM acceleration prior to seat-off were identified to construct the ROSv and ROSa. Stability margins for the ROSv and ROSa were calculated as the shortest distance from the experimental data to the forward boundary of the ROSv and ROSa, respectively. Maximum trunk flexion angle was also calculated as the angle formed between the trunk segment and vertical line at its maximum during STS task.

A paired t-test was performed to detect differences in COM motion between Young (Norm) and Young (Bend) conditions. An independent t-test was used for the other comparisons. Statistical analyses were performed using SPSS (Chicago, IL). Significance level was adjusted by Bonferroni correction to account for multiple comparisons ( $\alpha=.008$ ).

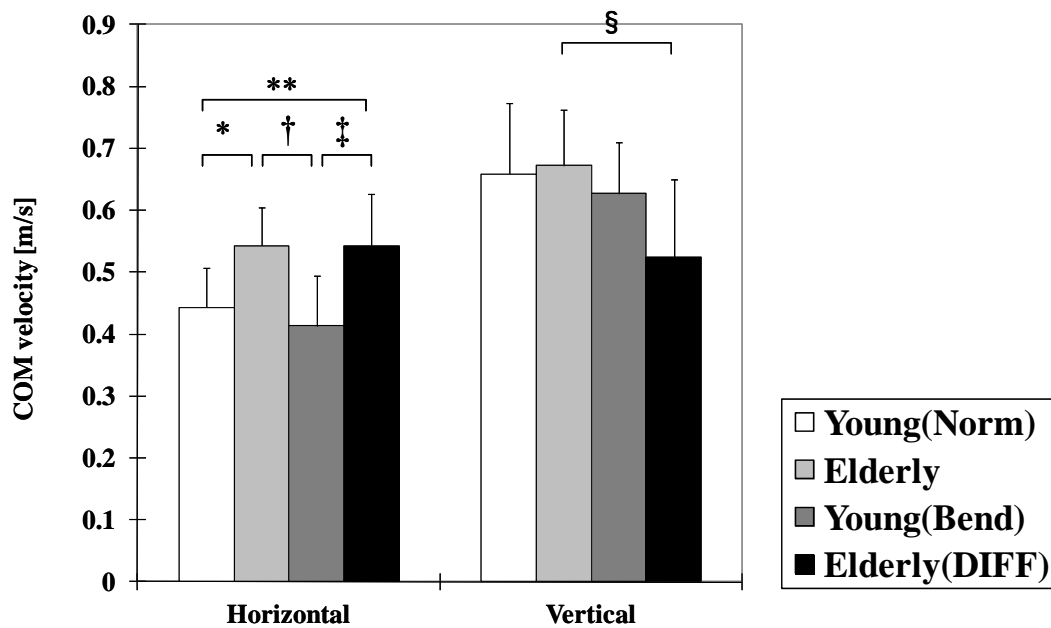
## Results

Young (Bend) group showed significantly larger maximum trunk flexion angle than the other groups ( $p<.001$ ; Young (Bend):  $69.0^\circ$  (SD 9.9); Young (Norm):  $42.5^\circ$  (SD 8.0); Elderly:  $41.3^\circ$  (SD 6.3); Elderly (DIFF):  $50.3^\circ$  (SD 7.9)). Elderly (DIFF) group showed a larger trunk flexion angle than Young (Norm) or Elderly groups but it was not statistically significant ( $p\leq.043$ ).

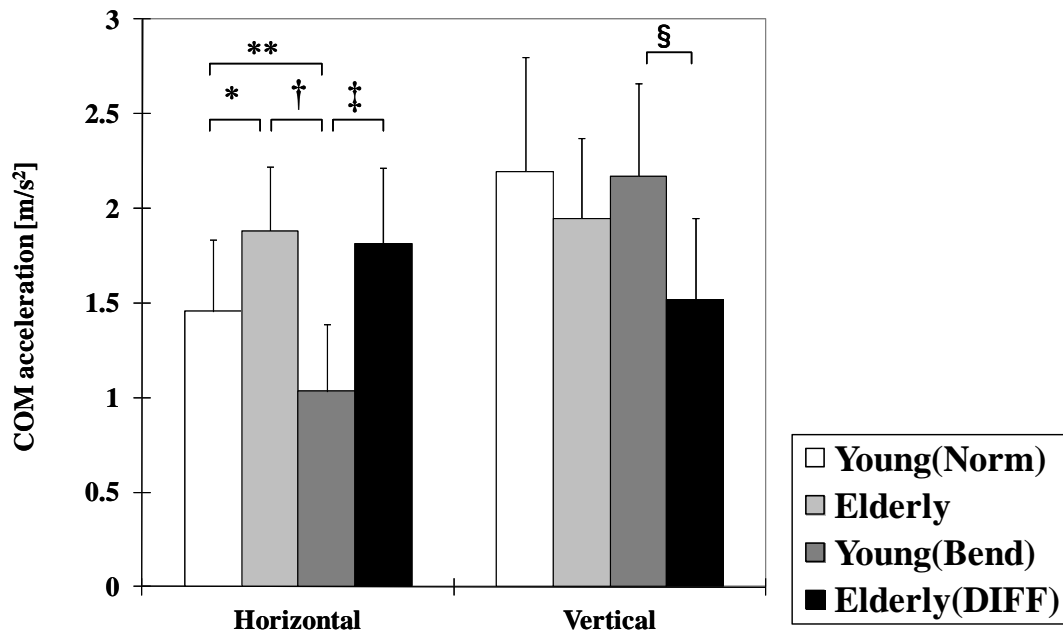
For most trials, seat-off was initiated right after COM velocity reached its peak, and COM velocity decreased with a backward COM acceleration after seat-off (Fig.2.2). Elderly and Elderly (DIFF) groups demonstrated a significantly greater peak horizontal COM velocity than the two Young conditions ( $p \leq .006$ ; Fig.2.3). No significant differences were detected either between the two Young conditions or between Elderly and Elderly (DIFF) groups. Elderly group also demonstrated a significantly larger peak horizontal COM acceleration than did the Young for the two conditions ( $p \leq .008$ ; Fig.2.4). Young (Bend) condition demonstrated significantly smaller peak horizontal COM acceleration than Young (Norm) condition although no significant difference was detected in the peak horizontal COM velocity between them ( $p < .001$ ; Fig.2.4). Elderly (DIFF) showed a significantly smaller vertical COM velocity than Elderly group ( $p = .003$ ; Fig.2.3), while their vertical COM acceleration was significantly smaller than Young (Bend) group ( $p = .003$ ; Fig.2.4).



**Figure 2.2.** Representative time-history plot of horizontal COM velocity and acceleration for Young (Norm) group. A similar trend was seen in the other groups.

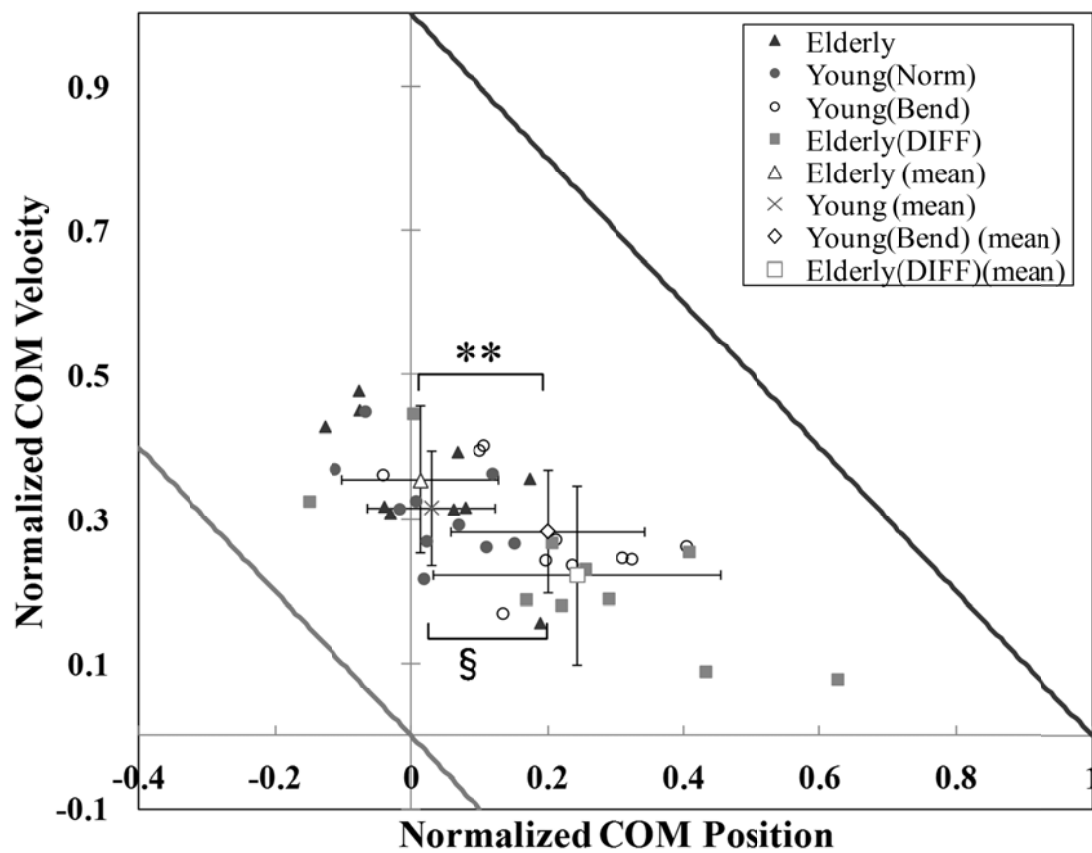


**Figure 2.3.** Peak COM velocity in horizontal and vertical directions for Young (Norm), Elderly, Young (Bend), and Elderly (DIFF) groups. Values are mean  $\pm$  SD. (\* $p$ =.002, \*\* $p$ =.006, † $p$ =.001, ‡ $p$ =.002, § $p$ =.003.)

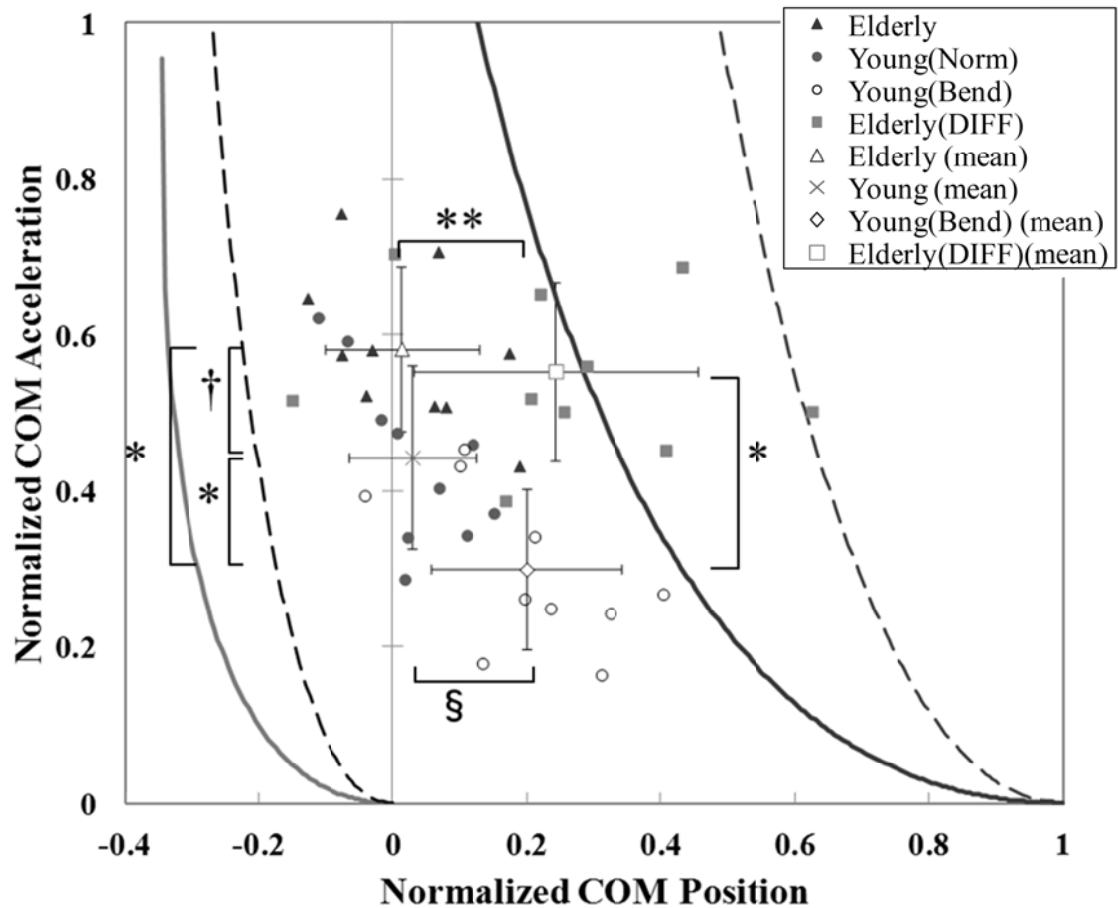


**Figure 2.4.** Peak COM acceleration in horizontal and vertical directions for Young (Norm), Elderly, Young (Bend), and Elderly (DIFF) groups. Values are mean  $\pm$  SD. (\* $p$ =.008, \*\* $p$ <.001, † $p$ <.001, ‡ $p$ <.001, § $p$ =.003.)

Data from all trials of each group were located within the boundaries of the ROSv (Fig.2.5). While data for Young (Norm), Young (Bend), and Elderly groups were all located inside the ROSa(a), 15 out of 62 trials from Elderly (DIFF) group were located outside the forward boundary of the ROSa(a), but within that of the ROSa(b) (Fig.2.6). 8 of them were located outside the forward boundary of the ROSa(b).



**Figure 2.5.** Normalized COM velocity at seat-off with respect to normalized COM position at seat-off for Young (Norm) (●), Elderly (▲), Young (Bend) (○), and Elderly (DIFF) (■) groups. Two solid lines indicate the boundaries of the ROSv derived by equation (1). Each plot indicates the mean data for each subject. Mean value with standard deviation (error bars) for each group is also indicated. (\*\* $p=0.003$ , §  $p<0.001$ .)



**Figure 2.6.** Normalized peak COM acceleration prior to seat-off with respect to normalized COM position at seat-off for Young (Norm) (●), Elderly (▲), Young (Bend) (○), and Elderly (DIFF) (■) groups. Each plot indicates the mean data for each subject. Mean value with standard deviation (error bars) for each group is also indicated. Solid and dashed lines represent boundaries of the ROSa (a) and ROS(b) derived using different COM acceleration profiles with the equation (2) and (3), respectively. Since the boundaries are dependent on the initial COM position varying between subjects, those boundaries were averaged for all trials. (\* $p < .001$ , \*\* $p = .003$ , † $p = .007$ , §  $p < .001$ .)

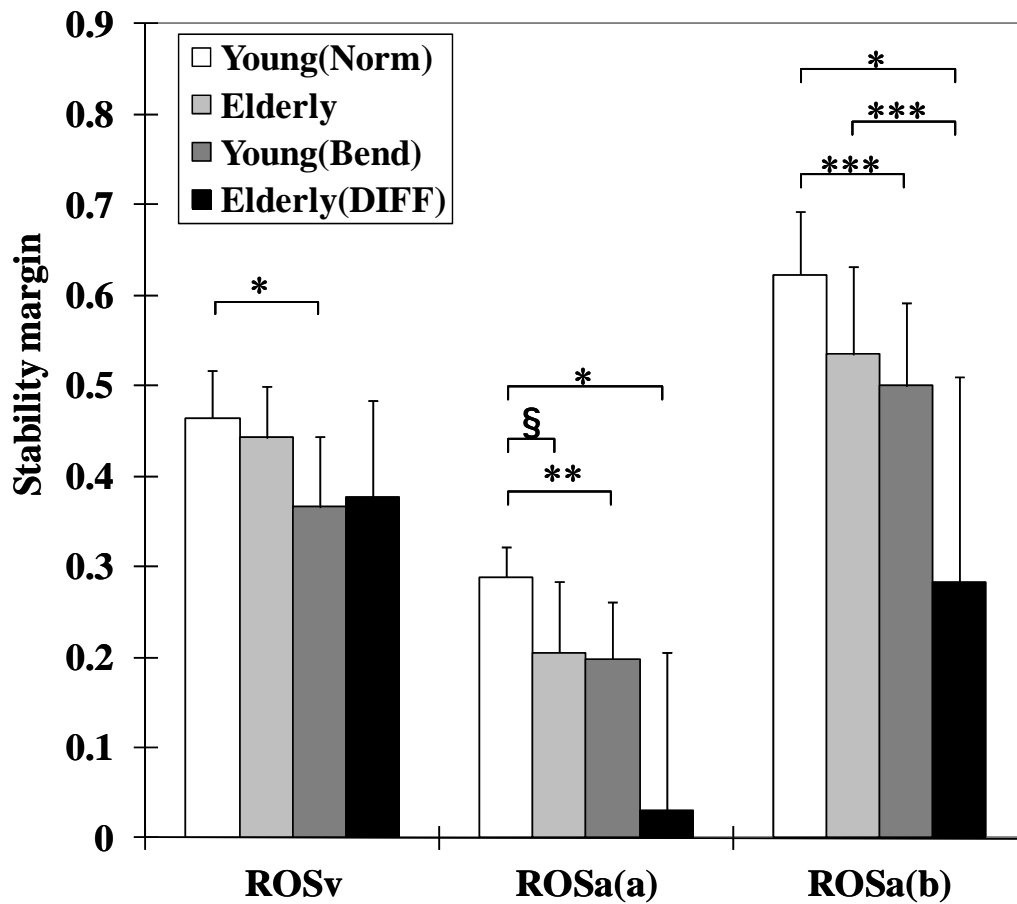
For COM positions, 29 out of 73 trials from Young (Norm), 36 out of 75 trials from Elderly, 7 out of 65 trials from Young (Bend), and 8 from 62 trials from Elderly (DIFF) were located posterior to the heel (0 on the X-axis), outside the BOS.

Mean normalized COM position at seat-off for Young (Bend) group (0.20 (SD 0.13)) was found to be significantly anterior to those for Young (Norm) and Elderly

groups ( $p \leq 0.003$ ; Elderly: 0.02 (SD 0.11); Young (Norm): 0.03 (SD 0.08)) (Fig.2.5 and 2.6). The COM position for Elderly (DIFF) group was also more anteriorly located than those two groups, but it was not significantly different from them ( $p \leq 0.10$ ; 0.25 (SD 0.22)).

No significant differences were detected in normalized COM velocity at seat-off (Young (Norm): 0.31 (SD 0.07); Elderly: 0.35 (SD 0.09); Young (Bend): 0.28 (SD 0.08); Elderly (DIFF): 0.23 (SD 0.11)) (Fig.2.5). Mean normalized peak COM acceleration significantly differed among all subject groups, except between Elderly and Elderly (DIFF) and between Young (Norm) and Elderly (DIFF) groups ( $p \leq 0.007$ ; Young (Norm): 0.44 (SD 0.11); Elderly: 0.58 (SD 0.10); Young (Bend): 0.30 (SD 0.10); Elderly (DIFF): 0.55 (SD 0.10)) (Fig.2.6). Elderly group demonstrated significantly larger normalized peak COM acceleration than the Young group for the two conditions ( $p \leq 0.007$ ). Elderly (DIFF) and Young (Norm) group showed a significantly larger normalized peak COM acceleration than that for Young (Bend) group ( $p < 0.001$ ).

Young (Bend) condition showed significantly smaller stability margin than Young (Norm) condition in the ROSv ( $p \leq 0.001$ ), while no significant differences were detected in the other comparisons (Fig.2.7). For the stability margins based on the ROSa(a) or ROSa(b), we further detected significant group differences between Young (Norm) and Elderly or Elderly (DIFF) groups ( $p \leq 0.007$ ) in the ROSa(a), and between Elderly (DIFF) and Young (Norm) or Elderly groups ( $p \leq 0.005$ ). Elderly (DIFF) group showed the smallest stability margins, followed by Young (Bend), Elderly, and Young (Norm) groups. Stability margins for Elderly (DIFF) group were significantly smaller than those for Young (Norm) and Elderly groups when the ROSa (b) was used ( $p \leq 0.005$ ).



**Figure 2.7.** Stability margins based on the ROSv and ROSa for Young (Norm), Elderly, Young (Bend), and Elderly (DIFF) groups. Error bars indicate standard deviation.

(\* $p < .001$ , \*\* $p = .001$ , \*\*\* $p = .005$ , §  $p < .007$ .)

### Discussion

The objectives of this study were to establish a region of stability using COM acceleration and to assess its feasibility to differentiate individuals with difficulty in STS movement from healthy individuals. Patterns of the boundaries for the ROSa were similar to those of the ROSv, where the COM velocity and acceleration for a successful movement termination are greater when the COM at seat-off is located posterior to the BOS. The ROSa showed quadratic boundaries, unlike the linear boundaries of the ROSv,



since the ROSa boundaries were dependent on the initial COM position as well as that at seat-off.

We hypothesized that age- and strategy-related differences in COM control could be more clearly distinguished with its acceleration as compared to velocity, and thereby the ROSa would better differentiate individuals with difficulty in STS movement from healthy individuals. Although no significant group differences were detected in normalized horizontal COM velocity at seat-off, peak normalized horizontal COM acceleration differed significantly among all subject groups, except between Elderly and Elderly (DIFF) and between Young (Norm) and Elderly (DIFF) groups. The four subject groups were more clearly distinguished with the use of COM acceleration. Elderly (DIFF) group showed a significantly smaller stability margin than the healthy young and elderly groups when the ROSa(b) was used, even though there were no detectable differences in the ROSv. These results suggest that COM acceleration could be a more sensitive measure than COM velocity to distinguish individuals with different balance control abilities, and that the ROSa could better differentiate individuals with difficulty in STS movement from healthy individuals than the ROSv.

Young (Bend) and Elderly (DIFF) groups placed their COM significantly anterior to the other two groups with a similar COM velocity at seat-off. Both groups placed their COM over the BOS before seat-off (in more than 87% of the trials) with larger trunk flexion, which has been reported as a stabilization strategy used by older adults with functional limitations [10, 57, 71]. In contrast, for Young (Norm) and Elderly groups, the COM was located posterior to the heel, outside the BOS, for more than 40% of the trials, where the body is in a statically unstable position. However, their relatively larger

horizontal COM velocity (i.e. momentum) at seat-off provided a dynamic stability, as confirmed by the ROS<sub>v</sub>, where the data points outside the BOS are still within the ROS<sub>v</sub>. They appear to use a momentum transfer strategy, which has been recognized as the ideal and efficient strategy used by nondisabled persons of all ages [10, 43, 57, 68, 72, 73]. The ROS<sub>v</sub> provides insights about strategies used during STS movement, where healthy young and elderly groups, and Young (Bend) and Elderly (DIFF) groups appeared to use momentum transfer and stabilization strategies, respectively.

Although a similar strategy was observed between the two groups, their momentum control was different. Despite demonstrating a similar COM velocity at seat-off, they exhibited distinctly smaller and larger accelerations prior to seat-off. Both Elderly and Elderly (DIFF) groups generated significantly larger acceleration prior to seat-off, compared to the two Young conditions. They also showed significantly larger peak horizontal COM velocity than the Young for the two conditions prior to seat-off ( $p \leq .006$ ; Elderly: 0.54 m/s (SD 0.06); Elderly (DIFF): 0.54 m/s (SD 0.08); Young (Norm): 0.44 m/s (SD 0.06); Young (Bend): 0.41 m/s (SD 0.07)), in consistent with previous reports [59]. Elderly subjects developed a larger forward momentum with a larger acceleration prior to seat-off, and such momentum was used to bring the upper body into the BOS. In fact, using the forward momentum to move the upper body prior to seat-off could reduce lower extremity forces required to elevate the body [57]. This could be a compensation for age-related muscle weakness. Results from our muscle strength assessment (maximum isometric voluntary contraction) revealed significantly weaker knee extensor strength in both Elderly and Elderly (DIFF) groups as compared to Young group ( $p \leq .016$ ; Significance level was set at  $\alpha = .017$  for these three comparisons, adjusted

by Bonferroni correction: Elderly: 1.13 Nm/kg (SD 0.32); Elderly (DIFF): 1.01 Nm/kg (SD 0.35); Young: 1.49 Nm/kg (SD 0.29)). For Elderly (DIFF) group, they further reduced the generated momentum prior to seat-off, as seen in smaller COM velocity at seat-off compared to Elderly group.

Subjects having difficulty in STS task have been reported to place more importance on achieving postural stability [55]. It seems that Elderly (DIFF) group used a forward momentum to confine their COM over the BOS, and then initiated seat-off with a reduced velocity, which decreases the destabilizing effect of the forward momentum when the COM is closer to the forward boundary of the BOS. This decrease in COM velocity at seat-off was later confirmed with time-history plots of COM velocity and acceleration (Fig.2.5a, 2.5b). Both Young (Bend) and Elderly (DIFF) groups showed similar strategy, where they placed their COM significantly anterior to the other groups with a similar COM velocity at seat-off. However, Elderly (DIFF) group generated significantly larger acceleration prior to seat-off. Although they used similar strategies, in terms of COM position and momentum observed at seat-off, how momentum was controlled could still be different between healthy individuals and individuals with difficulty in STS.

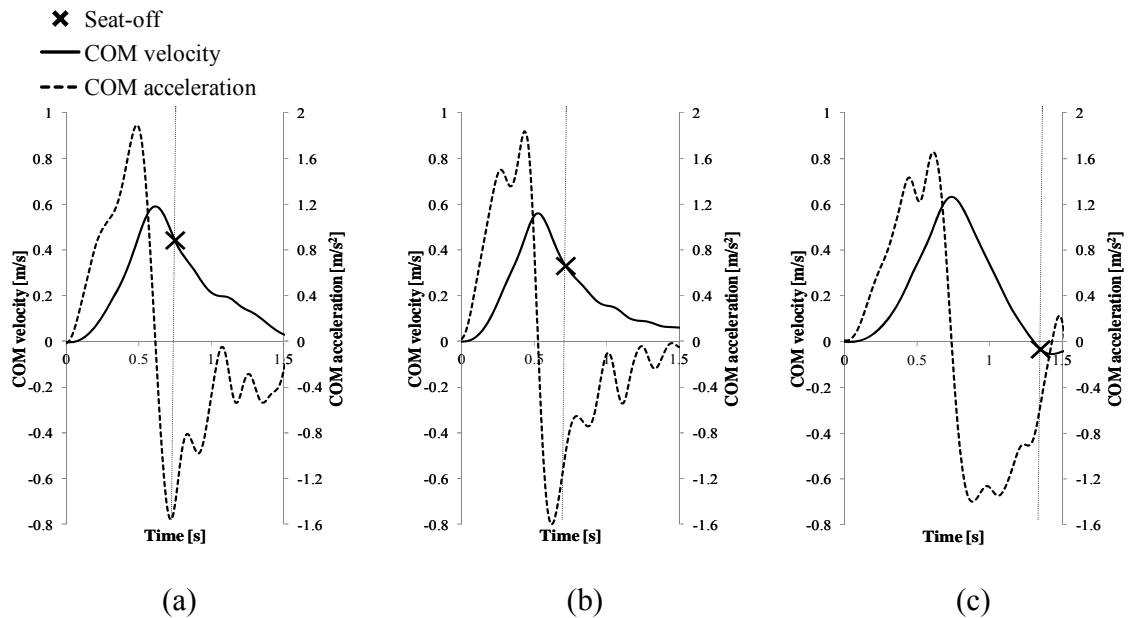
The ROSa could be sensitive to detect this difference. When the ROSa was used, Elderly (DIFF) group showed significantly smaller stability margins than the healthy young and/or elderly groups, while only Young (Bend) condition showed significantly smaller stability margin than Young (Norm) condition based on the ROSv. The ROSa could detect individuals with difficulty in STS even though there were no detectable differences in the ROSv. The ROSa could be useful to further classify strategies used by

individuals in terms of momentum control, which could further differentiate individuals with different balance control abilities, even though they fall into the same category in the ROSv.

Fifteen out of 62 trials from Elderly (DIFF) group were located outside the forward boundary of the ROSa(a), but within that of the ROSa(b). With the COM acceleration model (a) (Fig.2.1a), we assumed that the COM only undergoes forward acceleration prior to seat-off. Such an assumption would represent an extreme scenario, not allowing an individual to reduce the forward momentum prior to seat-off, and result in a conservative estimation for the forward boundary of ROSa. This explains why these data were located outside the ROSa(a) even with successful STS performance. The acceleration model (b) included the backward acceleration portion, which reduces the momentum prior to seat-off. It was confirmed that this profile would better represent the empirically measured COM acceleration profile (Fig.2.2 and 2.8a). Thus, it is reasonable to have those data points included within the forward boundary of the ROSa(b).

Eight data points from 2 subjects were still located outside the forward boundary of the ROSa(b). Those individuals placed their COM even more anteriorly (mean 0.53; group mean 0.25) with much smaller COM velocity (mean 0.08; group mean 0.23) at seat-off, but with a similar peak COM acceleration (mean 0.59; group mean 0.55). They appeared to perform STS with a similar COM acceleration like other subjects in Elderly (DIFF) group, which resulted in a similar velocity profile. They might have difficulty in controlling a large COM momentum at seat-off and had to further reduce the COM velocity prior to seat-off, resulting in a delay in seat-off timing. Such a delay, therefore, allowed the COM to reach a more anterior position, which placed their data points

outside the ROSa(b). This delay was confirmed with time-history plots of COM velocity and acceleration (Fig.2.8c).

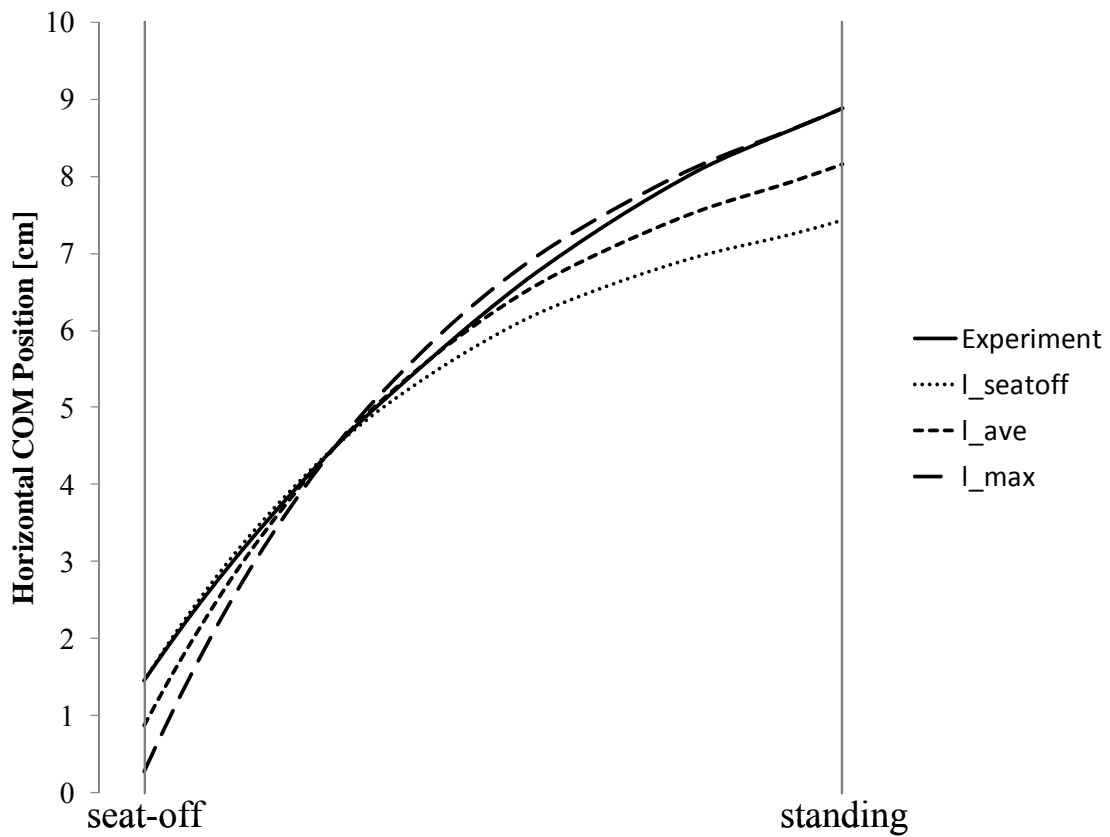


**Figure 2.8.** Representative time-history plots of horizontal COM velocity and acceleration for (a) healthy elderly adults (Elderly), (b) elderly adults with difficulty in STS (Elderly (DIFF)), and (c) elderly individuals with difficulty in STS whose data points were outside the ROSa(b).

The ROS<sub>v</sub> is determined based on the COM velocity and position at seat-off, while the ROS<sub>a</sub> is based on the peak COM acceleration prior to seat-off and COM position at seat-off, which would determine the possible amount of momentum generated prior to seat-off. While the ROS<sub>v</sub> provides information about COM motion state at seat-off, the ROS<sub>a</sub> provides insights into how that state is attained prior to seat-off, which could differentiate individuals with functional limitations from healthy individuals. The ROS<sub>a</sub> might be useful to further classify strategies used by individuals and provide an insightful evaluation of balance dysfunction. In supporting previous reports that difficulty in performing STS is associated with falls in older adults [11-15, 18-21], 7 out of 10

elderly subjects with difficulty in STS in this study had two or more falls in the year either prior to or after the testing, based on follow-up check-ups. Identifying older adults with difficulty in STS using regions of stability could enhance our identification of elderly fallers. Future studies should include elderly fallers to assess the capability of the model to differentiate them from healthy individuals.

The use of a constant length IPM to describe the multi-segmental STS movement could be a limitation in the study. We examined differences in COM position between the experimental data and that estimated from the IPM (Fig.2.9). Using the average pendulum length between seat-off and standing, the maximum differences were found to be 1.3 cm (SD 0.4) and 14.9 cm (SD 2.9) in the horizontal and vertical directions, respectively. This could indicate that the multi-segmental motion mainly affects the vertical COM movement during STS. We also used different pendulum lengths (length at seat-off or standing) in our model and found that it did not change our findings on the ROS<sub>v</sub> and ROS<sub>a</sub>. It has also been shown that the empirically measured COM positions and velocities during STS were adequately illustrated by the FSR [43], which is similar to our ROS<sub>v</sub>, and the model predictions from the FSR were consistent with experimental observations for STS followed by a volitional fall and balance recovery from induced slips during STS [43, 45, 60, 61, 74]. Therefore, the constant length IPM seems to adequately describe the horizontal COM movement during STS, although models taking into account changes in pendulum length, such as telescopic inverted pendulum [58, 59], would be more applicable.



**Figure 2.9.** Representative horizontal COM position from the experiment (solid line) and the estimated horizontal COM position based on 3 different fixed pendulum lengths ( $l_{\text{seat-off}}$ : the pendulum length at seat-off,  $l_{\text{ave}}$ : the average value in pendulum length between those at seat-off and standing,  $l_{\text{max}}$ : the pendulum length at standing (its maximum value), between seat-off and standing (when the pendulum reached its maximum length).

In conclusion, this study established regions of stability during STS based on the COM position at seat-off and its instantaneous velocity (ROSv) or its peak acceleration (ROSa), and assessed feasibility of these regions to differentiate individuals with difficulty in STS movement from healthy individuals. While the ROSv provides information about the COM motion state at seat-off, the ROSa provides insights into how that state is attained prior to seat-off, which could differentiate individuals with functional limitations from healthy individuals. Elderly adults with difficulty in STS

showed a significantly smaller stability margin than the other healthy groups when the ROSa was used, even though there were no detectable differences in the ROSv. The ROSa could provide a quantitative basis for assessing dynamic stability during STS movement, which allows us to differentiate individuals with difficulty in STS movement, who are most likely at a risk for imbalance or falls.

### Bridge

This chapter proposed a novel method to identify dynamic limits of balance control during STS movement using COM acceleration, and assessed its feasibility to differentiate individuals with difficulty in STS movement from healthy individuals. The results demonstrated that examining COM acceleration could provide further insights into how balance is controlled during STS, and that our model with COM acceleration could better differentiate individuals with difficulty in STS movement from healthy individuals than the traditional model with COM velocity. Identifying older adults with difficulty in STS using the region of stability we proposed could enhance our identification of elderly fallers.

In addition to difficulty in STS, poor postural control ability is a risk factor in the increased incidences of falls in the elderly. The following two chapters investigated postural recovery responses to backward support surface translations during quiet standing for healthy young and elderly adults. To identify biomechanical factors affecting postural recovery responses, the relationship between ankle muscle strength and functional BOS (FBOS), the effective limits of the BOS, as well as perturbation acceleration threshold to maintain balance was first investigated in Chapter III. Based on



the findings in Chapter III, a biomechanical model that predicts perturbation acceleration threshold to maintain balance was developed in Chapter IV.

## CHAPTER III

### ANKLE DORSIFLEXOR STRENGTH RELATES TO THE ABILITY TO RESTORE BALANCE DURING A STANCE PERTURBATION IN THE ELDERLY

The study described in this chapter was developed by a number of individuals, including Dr. Wei-Li Hsu, Dr. Marjorie Woollacott, and Dr. Li-Shan Chou. Dr. Wei-Li Hsu, Dr. Marjorie Woollacott, and Dr. Li-Shan Chou contributed substantially to the work by providing critiques about data analysis and development of methodologies. I was the primary contributor to the data analysis and interpretation, and did all the writing.

#### Introduction

Falls are one of the most serious problems among the elderly, resulting in fatal physical injuries [1-3]. Poor postural control is one of the contributing factors to increased incidence of falls in the elderly [24]. Responses to unexpected perturbations during standing, such as support surface translations, have been commonly used to examine dynamic postural control ability [27-31]. Elderly adults are reported to take compensatory steps more often than young adults when a less severe balance disturbance is applied. Occurrence of a stepping response has been traditionally regarded as a consequence of the whole body center of mass (COM) moving beyond the limits of the base of support (BOS) [32]. However, even with a location inside the BOS, a standing posture might not be maintained if the COM traverses with a sufficiently large horizontal velocity [27, 29, 30, 43]. Therefore, step initiation could depend on the COM position in relation to the BOS and its instantaneous velocity, i.e., momentum.

One possible explanation for the increased incidence of stepping in elderly adults would be an inability to generate the necessary muscle torques to control the horizontal COM momentum [32]. However, Hall and colleagues (1999) showed that neither the magnitude nor rate of ankle muscle torque production during the initial balance response differed between young and healthy elderly adults, suggesting that initial ankle muscle torque production may not be a limiting factor in the increased incidence of stepping in the elderly [32, 75].

Although initial muscle torque responses for momentum control may not be different between healthy young and elderly adults, differences could be found in the effective limits of the BOS, which provides stability margins for the COM movement. During stance perturbations, actively controlling the center of pressure (COP) position is required to keep the COM within a desired region, as the distance between the COP and COM correlates with the horizontal COM acceleration [76], which regulates momentum. The BOS could serve as the limits for COM control since it provides a possible range for COP movement. It has been shown that the functional base of support (FBOS), defined as the effective limits of COP movement, decreases with aging [26]. A decrease in FBOS indicates a constriction of the limits of stability for COM control, which would reduce an individual's ability to restore balance during perturbed stance.

Studies of quiet or perturbed standing have identified the dominance of the ankle muscles in controlling balance in the antero-posterior direction [77]. It has been suggested that reduced ankle muscle strength plays an important role in the loss of balance in elderly adults [75, 78]. Enhancement of ankle muscle strength was reported to lead to improvements in balance recovery during standing perturbations in the elderly

[79]. Weakness in ankle muscles would contribute to a decreased FBOS and explain the increased incidence of stepping in the elderly when stance is perturbed.

The objective of this study was, therefore, to examine the association of ankle muscle strength, FBOS, or perturbation acceleration with the ability to recover balance. We hypothesized that FBOS would be a sensitive measure to predict threshold perturbation acceleration for taking a step or heel-rise, and ankle plantarflexor strength would be a significant predictor of FBOS as well as threshold perturbation acceleration during a backward support surface translation, which induces forward body sway.

### Methods

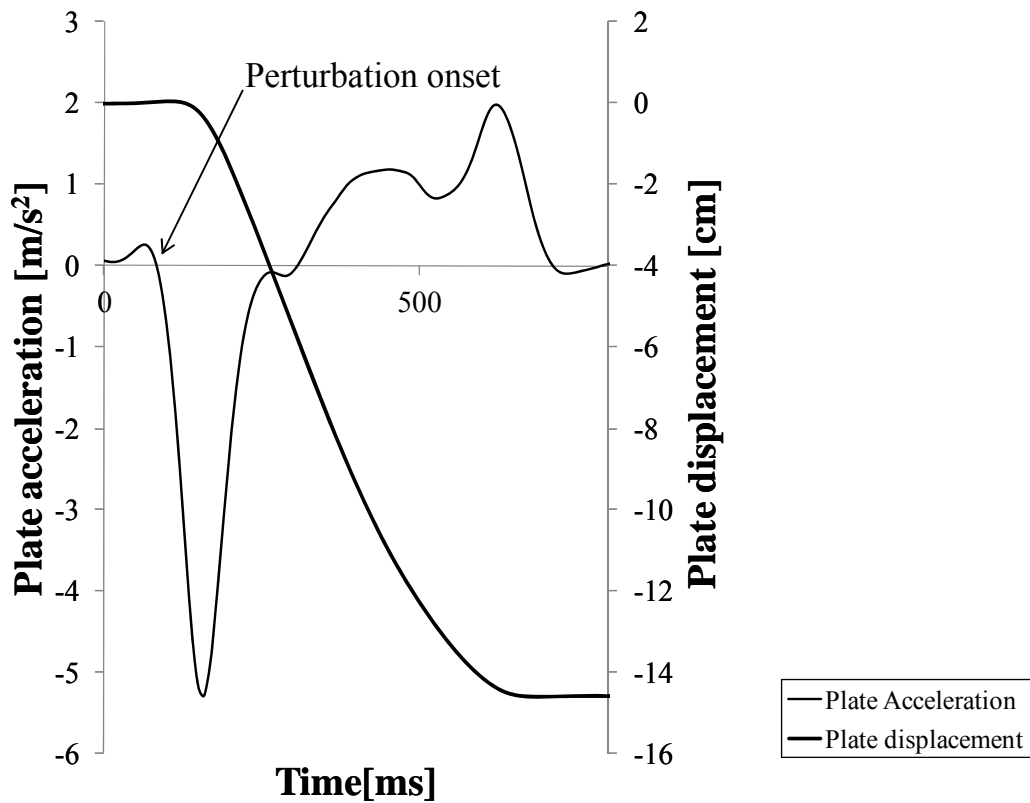
Standing posture of 16 healthy young adults [8 women; mean age  $25.4 \pm 4.3$  years, mean height  $171.3 \pm 8.7$  cm, mean mass  $67.7 \pm 12.5$  kg] and 16 healthy elderly adults [9 women; mean age  $74.9 \pm 6.2$  years, mean height  $163.1 \pm 9.7$  cm, mean mass  $67.5 \pm 10.4$  kg] was perturbed with a backward support surface translation. Subjects stood with one foot on each of two electronically synchronized force platforms (Institute of Neuroscience, University of Oregon) with their arms folded on their chest [80]. They wore a lightweight harness attached to an overhead trolley to ensure safety and an assistant remained by the side of the subject to prevent a fall. They were asked to try not to bend their trunk and to maintain their balance without moving their feet or taking a step. Six perturbations (nonlinear ramp-to-parabola acceleration waveforms) included one for each of the velocities 15, 30, 40, 50, 60, and 70 cm/s and the platform moved backward for 15 cm at all velocities. Perturbations were presented in a pseudo-randomized manner whereby less severe disturbances occurred within earlier trials [32]. This testing order served as a

precautionary measure to ensure that subjects would not be exposed to the more severe disturbances early in the test session. The experimental protocol was approved by the Institutional Review Board. No participants had a history or clinical evidence of neurological, musculoskeletal or other medical conditions.

Motion data were captured with an eight-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA) at 120Hz. A total of 29 reflective markers were placed on each subject's bony landmarks [37]. Additionally, 4 markers were placed on the corners of the moving platform to obtain platform movement. Three-dimensional marker trajectory data were then low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 8Hz.

Perturbation acceleration was used to describe the magnitude of the balance disturbance since previous studies have shown that platform acceleration provides the initial destabilizing input in the postural response [80-85]. Perturbation acceleration for each speed condition was calculated from the 2<sup>nd</sup> derivative of plate marker displacement. The onset of platform motion was defined as the first zero crossing in a backward acceleration phase of the platform acceleration (Fig.3.1) [80]. Ground reaction forces were collected at 960 Hz.

Responses to the platform perturbation were classified as in-place, step (STEP), and/or heel-rise (HR). An in-place response occurred when the feet remained in contact with the supporting surface throughout the trial. A STEP response occurred if a step was observed. A HR response occurred if the heel was raised more than 1% of body height. A trial was considered as both a HR and STEP trial when a HR response was observed prior to a STEP response.



**Figure 3.1.** Representative plate displacement and acceleration profiles for the 40cm/s condition. The onset of platform motion was defined as the first zero crossing in the backward acceleration phase of the platform acceleration.

Electromyographic (EMG) data were collected to identify onset of muscle activities. Bipolar surface electrodes (Motion Lab Systems Inc., Baton Rouge, LA) were placed on the skin over the right lateral gastrocnemius (GA) and sampled at 960 Hz. EMG data were rectified, but not filtered for analysis to determine muscle onset latencies. Criteria for determining muscle onset latencies were that EMG activity was greater than the baseline mean plus three standard deviations, and that half of the EMG sample points in the burst remained above this level for at least 40 ms [75]. Baseline data for each trial were collected 2 to 3 seconds prior to onset of plate movement.

In addition, COP positions in the antero-posterior (AP) direction during sustained maximal forward and backward leaning were measured to determine forward FBOS (FFBOS), backward FBOS (BFBOS), and total FBOS (TFBOS) [26]. FFBOS and BFBOS were calculated from the ankle joint with the forward direction as positive, and TFBOS was the difference between FFBOS and BFBOS, the total area between forward limit (FFBOS) and backward limit (BFBOS). Ankle plantar- and dorsi-flexor strengths (PF and DF, respectively) of the dominant leg were measured during isometric maximum voluntary contraction in a seated position at a neutral ankle position with a BIODEX dynamometer (Biodex Medical Systems, NY).

The combined sagittal plane joint moment from both ankles during each perturbation trial was calculated based on a foot segment model using collected ground reaction forces and kinematic data [75]. Anthropometric data were estimated using the initial work of Dempster [34]. Peak ankle joint moment was defined as the maximum ankle torque following onset of plate movement. If a step occurred during the trial, the maximum torque achieved prior to step onset was chosen. Step onset was defined as the instant when the foot was lifted from the floor and was detected using the ground reaction force.

The active phase of ankle torque development (torque generation rate) was determined as the rate of ankle joint moment generation calculated during a period of 60 ms following the onset of GA muscle activity [75]. If a stepping event took place during this time period, this rate of ankle joint moment generation was not calculated [75]. The peak COP displacement was calculated as the largest distance between the COP and ankle during each trial (or until step onset).

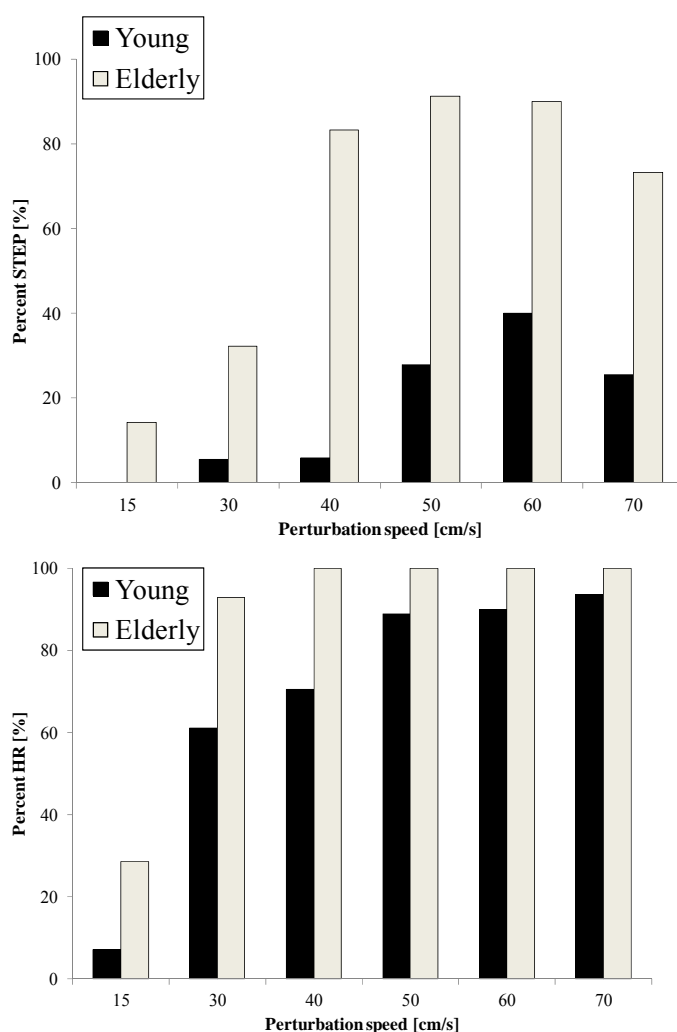
An independent *t*-test was performed to examine group differences in ankle muscle strength, FBOS measures, muscle onset latency, ankle torque generation rate, peak ankle joint moment, and peak COP displacement. Linear regression analyses were performed to examine the relationship between ankle muscle strength and FBOS measures as well as threshold perturbation acceleration for STEP and HR responses (TAccSTEP and TAccHR, respectively). Threshold perturbation acceleration was the lowest magnitude where STEP or HR responses were obtained for each subject. Significance level was set at  $\alpha=.05$ . Pairwise comparisons in muscle onset latency, ankle torque generation rate, peak ankle joint moment, and peak COP displacement were analyzed with adjustments for multiple comparisons using Bonferroni procedure ( $\alpha_{PC} = .05/6 = .0083$ ).

## Results

As the perturbation speed increased, the number of STEP and HR responses increased, except for the fastest perturbation condition for STEP response (70cm/s) (Fig.3.2). Overall, 19.7% and 65.5% of the overall number of trials were STEP responses, and 75.2% and 88.5% were HR responses for young and elderly subjects, respectively. In addition, 26.2% and 74.0% of HR responses were followed by STEP responses for young and elderly subjects, respectively.

Elderly subjects showed a significantly smaller DF strength than young subjects ( $p<.001$ ), but no significant group difference was found in the PF strength. They also showed significantly smaller FFBOS, BFBOS, and TFBOS ( $p\leq.006$ , Table 3.1).





**Figure 3.2.** Number of STEP and HR responses in percentage by group and perturbation condition.

**Table 3.1.**

Ankle muscle strength (PF: Plantarflexor, DF: Dorsiflexor) and FBOS measures normalized by foot length (FFBOS: Forward FBOS, BFBOS: Backward FBOS, TFBOS: Total FBOS). A negative value for BFBOS for Young subjects indicates a larger FBOS in the backward direction since it was calculated from the ankle joint with the forward direction as positive.

Variables	Group		<i>p</i>
	Young	Elderly	
PF[Nm/kg]	1.10±0.30	1.06±0.40	.761
DF[Nm/kg]	0.47±0.11	0.31±0.10	<.001*
FFBOS[%FL]	65.3±5.1	58.6±9.0	.006*
BFBOS[%FL]	-3.7±5.0	2.5±4.0	.001*
TFBOS[%FL]	69.0±4.0	56.0±9.7	<.001*

\**p* < .05.

For all subjects, DF strength was found to be significantly correlated with FF BOS, BF BOS, TF BOS, and TAccHR with a moderate effect size ( $R^2 = .24 \sim .41$ ; Table 3.2). Significant correlations were also found between all the FBOS measures and threshold accelerations for HR and STEP, except for BF BOS and threshold acceleration for STEP response. Within each subject group, significant correlation was detected only between DF strength and TAccHR for the young group.

No significant differences were detected in either muscle onset latency or ankle torque generation rate between young and elderly subjects for all perturbation conditions (Fig.3.3a, 3.3b). However, elderly subjects showed significantly smaller peak ankle joint moment than young subjects in the 60cm/s condition ( $p=.008$ , Fig.3.3c). The peak COP displacement for elderly subjects was also significantly reduced compared to young subjects in this condition ( $p=.004$ , Fig.3.3d).

### Discussion

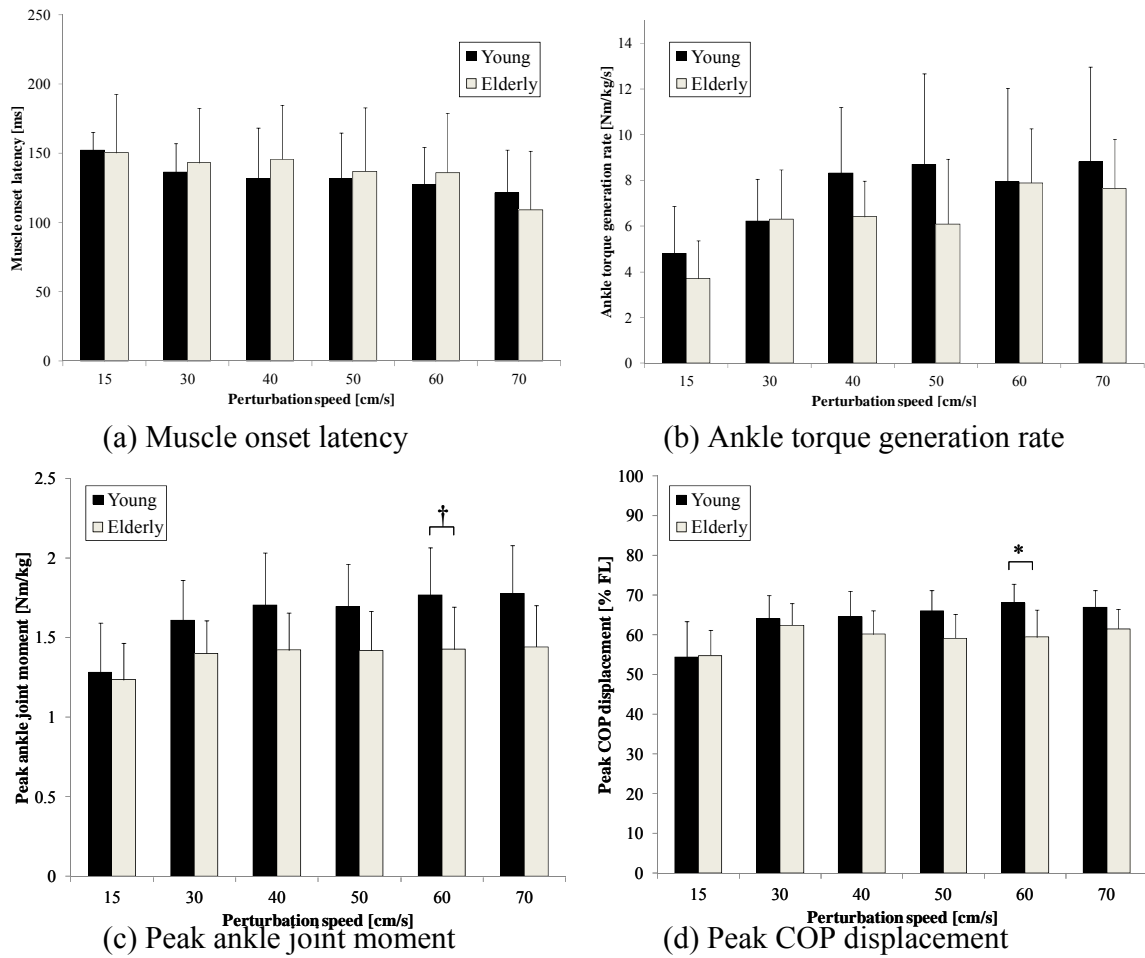
The objective of this study was to examine the association of ankle muscle strength, FBOS, or perturbation acceleration with the ability to recover balance. We hypothesized that FBOS would be a sensitive measure to predict threshold perturbation acceleration, and ankle plantarflexor strength would be a significant predictor of FBOS as well as threshold perturbation acceleration. Our results, with young and elderly subjects combined, demonstrated significant correlations between all the FBOS measures and threshold accelerations for HR and STEP responses, except for BF BOS and threshold acceleration for the STEP response. However, instead of ankle plantarflexor strength,

**Table 3.2.**

Results of linear regression analyses (PF and DF: Plantarflexor and Dorsiflexor strengths, respectively, FFBOS: Forward FBOS, BFBOS: Backward FBOS, TFBOS: Total FBOS, TAccSTEP and TAccHR: Threshold perturbation accelerations for STEP and HR responses, respectively).

Predicted variables	Young		Elderly		Overall	
	Regression equation	R <sup>2</sup>	Regression equation	R <sup>2</sup>	Regression equation	R <sup>2</sup>
FFBOS[m]	-0.001PF+0.164	0.00	0.011PF+0.131	0.04	0.009PF+0.144	0.02
	0.041DF+0.144	0.10	0.067DF+0.122	0.07	0.083DF*+0.120	0.24*
BFBOS[m]	-0.014PF+0.01	0.10	0.007PF	0.08	-0.002PF	0.00
	-0.032DF+0.005	0.07	-0.033DF+0.017	0.10	-0.060DF*+0.022	0.29*
TFBOS[m]	0.013PF+0.159	0.05	0.004PF+0.132	0.00	0.010PF+0.144	0.02
	0.074DF+0.138	0.19	0.100DF+0.105	0.14	0.143DF*+0.099	0.41*
TAccSTEP [m/s <sup>2</sup> ]	-1.35PF+7.97	0.13	0.55PF+4.30	0.02	0.42PF+4.89	0.01
	-11.3DF+10.8	0.24	1.34DF+4.45	0.01	3.48DF+4.17	0.04
	33.0FFBOS+1.3	0.20	25.6FFBOS+1.2	0.17	34.0FFBOS*+0.4	0.23*
	35.2BFBOS+6.7	0.14	-35.9BFBOS+5.1	0.06	-39.2BFBOS+5.4	0.09
	18.3TFBOS+3.5	0.04	27.6TFBOS+1.1	0.22	35.0TFBOS*+0.3	0.32*
TAccHR [m/s <sup>2</sup> ]	-1.22PF+6.00	0.01	-0.064PF+3.885	0.00	-0.31PF+4.55	0.01
	6.09DF*+1.78	0.27*	2.78DF+2.94	0.09	4.90DF*+2.31	0.29*
	23.3FFBOS+0.8	0.07	9.11FFBOS+2.52	0.06	19.1FFBOS*+1.3	0.13*
	-8.86BFBOS+4.55	0.01	-34.6BFBOS+4.0	0.16	-29.3BFBOS*+4.2	0.13*
	18.6TFBOS+1.4	0.07	13.3TFBOS+2.0	0.15	18.2TFBOS*+1.4	0.20*

\* $p < .05$ .



**Figure 3.3.** Group comparison for (a) muscle onset latency, (b) ankle torque generation rate, (c) peak ankle joint moment, and (d) peak COP displacement normalized by foot length (†, \* $p < .0083$ )

dorsiflexor strength was found to be significantly correlated with FBOS measures and threshold perturbation acceleration for heel-rise.

Overall, higher rates of STEP and HR responses in elderly compared to young subjects were observed for each of the 7 perturbation speeds. We also observed that most elderly subjects took a step once they raised their heels (74.0% of HR responses), while most young subjects were able to restore their balance after heel-rise (73.8% of HR responses). As the perturbation speed increased, the number of STEP and HR responses

increased for both young and elderly subjects, except for the fastest perturbation condition for STEP responses. This is because platform perturbations were presented in a pseudo-randomized manner whereby slower translations occurred first. Some subjects who took a step during these earlier trials did not experience the fastest perturbation for safety purposes, which resulted in a decrease in percent STEP responses in the fastest perturbation condition.

No significant differences between the young and elderly were detected in either muscle onset latency or ankle torque generation rate in response to perturbations, which is in agreement with previous findings [75]. Peak ankle joint moment was found to be smaller for faster perturbation speeds in the elderly, with a significant difference at 60cm/s. These results confirmed that the initial ankle muscle torque response may not be a contributing factor to increased incidence of stepping in the elderly, but suggested that maximum ankle torque production during stance perturbations, which is likely to be related to FBOS, could still be a limiting factor.

Elderly adults showed a significantly smaller FBOS, which indicates a constriction of limits of balance. Consistent with this result, their peak COP displacement was smaller compared to young subjects for faster perturbation speeds, with a significant difference found at 60cm/s. These results could explain the smaller peak ankle joint moments observed in the elderly since the ankle joint moment is proportional to the distance between the COP and the ankle joint. Moreover, significant correlations were detected between FBOS measures and threshold perturbation accelerations. FBOS measures seem to be a sensitive measure to predict threshold perturbation accelerations for taking a step or heel-rise. The increased rate of stepping in the elderly during

backward stance perturbations could be due to a reduced FBOS, which limits the range of COP movement to control the COM to achieve upright standing.

A possible reason for a reduced FBOS in elderly adults could be their significantly weaker dorsiflexor strength as compared to young adults. The ankle dorsiflexor strength significantly correlated with all FBOS measures, where individuals with weaker dorsiflexor strength showed smaller FBOS measures. Those individuals also raised their heels during stance perturbations with smaller perturbation speeds, while plantarflexor strength did not correlate with any of these measures. Although no significant differences were detected in plantarflexor strength between young and elderly subjects, elderly subjects showed significantly smaller peak ankle plantarflexor moment during stance perturbations. This suggests that plantarflexor strength measured in a static condition does not reflect the maximum torque production during dynamic tasks.

Age-related declines in the ankle dorsiflexor strength have been demonstrated previously, where significant differences were detected between healthy older adults and fallers [78]. Backward translations were reported to induce muscle activities in tibialis anterior while individuals standing on their toes, suggesting that balance recovery after heel-rise would require dorsiflexor activation [86]. In fact, most elderly subjects in the present study took a step once they raised their heels, while most young subjects were able to restore their balance after heel-rise. It has also been shown that increased muscle activation on the anterior aspect of the body was observed as the velocity of backward translations increased [87]. These findings, taken together, suggest that responses from the ankle dorsiflexors could also be important for balance maintenance during backward stance disturbances. Elderly adults may not be able to control the COP and maintain

balance while standing on their toes as effectively as young adults due to weakness in ankle dorsiflexors, which would limit their ability to restore balance during stance perturbations.

In conclusion, ankle dorsiflexor strength was found to be significantly associated with FBOS measures as well as threshold perturbation speed for heel-rise during backward platform translations. Elderly subjects had a decreased FBOS, which would reduce the ability to restore balance during stance perturbations. FBOS measures and ankle dorsiflexor strength could be sensitive measures to detect elderly individuals with a decline in balance control.

### Bridge

This chapter examined the relationship between ankle muscle strength and functional BOS as well as perturbation acceleration threshold required to maintain balance. The results suggested that weakness in ankle dorsiflexors could limit the ability of elderly adults to restore balance and that FBOS and ankle dorsiflexor strength could be sensitive measures to detect elderly individuals with declined balance control.

In the next chapter, we established a biomechanical model which predicts perturbation acceleration threshold to maintain balance and compared its predictive capability with that of a statistical model.

CHAPTER IV  
ACCELERATION THRESHOLD PREDICTION DURING A BACKWARD STANCE  
PERTURBATION

The study described in this chapter was developed by a number of individuals, including Dr.Marjorie Woollacott, and Dr.Li-Shan Chou. Dr.Marjorie Woollacott, and Dr.Li-Shan Chou contributed substantially to the work by providing critiques about data analysis and development of methodologies. I was the primary contributor to the data analysis and interpretation, model development, and did all the writing.

Introduction

Falls are the leading cause of injury-related deaths and hospital admissions among older adults [33]. Early identification of individuals with a high risk for falling is needed to facilitate the provision of rehabilitation treatment to reduce future fall risks [6]. Impaired balance control is one of the contributing factors to the increased likelihood of falls in the elderly [24]. Stepping has been reported as a commonly executed protective response for balance recovery in response to unexpected perturbations during standing, and the frequency of using a stepping response to restore balance increases in older individuals [28, 30, 88-92].

Steps could occur in anticipation of an impending collision or fall, or in reaction to imposed horizontal movement of the whole body center of mass (COM) with respect to the base of support (BOS) [30]. A protective step would be engaged when the COM travels beyond the boundaries of the BOS. However, this COM displacement-dependent



view only presents a static model of standing balance, and it does not guarantee that postural stability during a movement could be maintained [30]. Even with a location inside the BOS, a standing posture might not be maintained if the COM possesses a sufficiently large horizontal velocity to carry it outside the BOS [27, 29, 30, 43].

Pai and Patton (1997) developed a biomechanical model to determine a set of feasible COM velocity-position combinations that guarantee stable upright posture. They predicted the limiting COM velocities in a given COM position that would still permit successful termination of the movement within the BOS. This dynamic model showed a much better predictive capability than the traditionally used static model for estimating stepping reactions [30].

Hof and colleagues (2005) also suggested the importance of the COM velocity to assess dynamic stability and derived the extrapolated center of mass (XcoM) to quantify standing stability using a simple inverted pendulum model, where the condition of dynamic stability was that the XcoM should be confined within the BOS [48]. Simoneau and Corbail (2005) took into account the effect of speed of ankle torque development for counteracting a loss of balance and determined the set of balance stability boundaries that lead to a successful balance recovery and demonstrated a better capability for predicting unsuccessful balance recovery than the statistical model [62].

Although these models could “differentiate” standing perturbations that require stepping responses based on the COM velocity-position relationship derived from experimental data, it does not allow us to “predict” whether a person would lose balance prior to the occurrence of perturbation. Velocity, which reflects momentum, only describes an instantaneous state of motion. If we were to predict the outcome of balance

recovery in response to unexpected perturbations, it is necessary to examine how an external perturbation alters the body momentum, which would lead to initiation of a protective step.

Acceleration induced by the external perturbation would produce a sudden change in the motion state of the whole body COM and its momentum. Previous studies have shown that platform acceleration provides the initial destabilizing input to postural response [80-85]. Welch and Ting (2008) demonstrated that the initial burst of muscle activity is primarily due to perturbation acceleration in response to support-surface translations. Postural responses have been shown to be scaled with the magnitude of perturbation acceleration in neck muscles of seated subjects [81, 82] and in perturbations induced by arm movements [83]. Perturbation acceleration also affected muscle onset latency and total ankle moment during standing [80, 81, 84]. Perturbation acceleration appears to provide the initial destabilizing input in the postural response. Thus, if we could estimate the acceleration profile that an individual would experience during a stance perturbation, it is feasible to predict the induced changes in momentum and thus its effect on COM movement.

During stance perturbations, actively controlling the center of pressure (COP) position is required to regulate the COM momentum and keep its position within a desired region, as the distance between the COP and COM correlates with the horizontal COM acceleration [76]. The BOS confines the limits for COM control since it provides a possible range for COP movement. The distance between the COM position and boundaries of the BOS would indicate the ability of an individual to generate acceleration for controlling COM momentum. However, it has been shown that the functional base of

support (FBOS), defined as the effective limits of COP movement, decreases with aging [26]. A decrease in FBOS indicates a constriction of the limits of stability for COM control, which would reduce an individual's ability to restore balance during perturbed stance. Increased incidence of stepping in elderly adults could be due to an inability to control the horizontal COM momentum resulting from a decrease in FBOS. Thus, the initial COM position in relation to the FBOS would reflect an individual's functional capability to control COM momentum and restore balance.

These observations, taken together, suggest that it is possible to predict whether an individual could successfully restore balance from an unexpected perturbation based on where the COM is initially located within the BOS/FBOS and COM acceleration profile during the initial destabilizing phase of the perturbation. Such prediction of balance loss would allow us to determine the perturbation threshold that could be sustained by each individual, which is needed to facilitate more effective rehabilitation treatment and prevent falls in the elderly.

Therefore, the objectives of this study were to build a biomechanical model that predicts the threshold perturbation acceleration required to maintain balance, and to compare the predictive capability of this biomechanical model to that of a statistical model.

## Methods

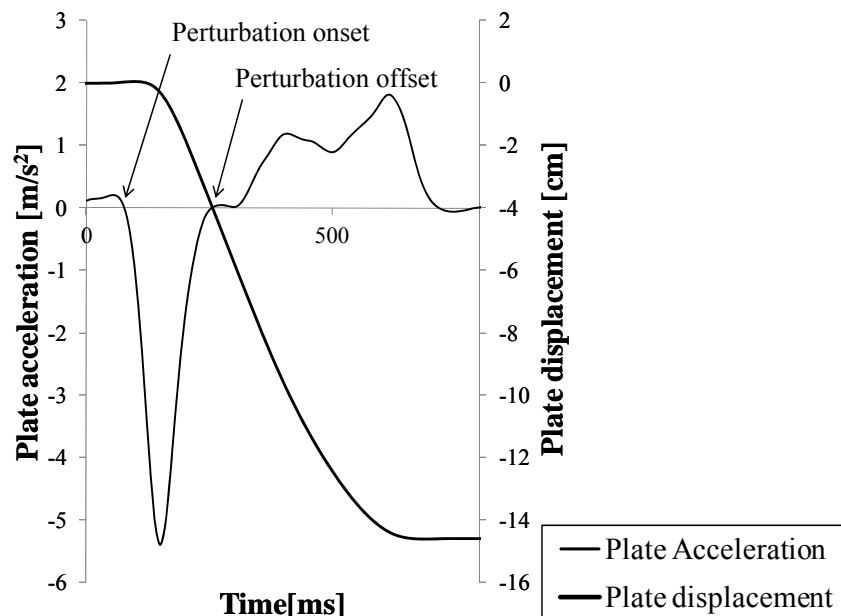
### *Experiment*

Standing posture of 16 healthy young adults [Young: 8 women; mean age 25.4±4.3 years, mean height 171.3±8.7 cm, mean mass 67.7±12.5 kg] and 16 healthy

elderly adults [Elderly: 9 women; mean age  $74.9 \pm 6.2$  years, mean height  $163.1 \pm 9.7$  cm, mean mass  $67.5 \pm 10.4$  kg] was perturbed with a backward support surface translation. Subjects stood with one foot on each of two electronically synchronized force platforms (Institute of Neuroscience, University of Oregon) with their arms folded on their chest [80]. The subject wore a lightweight harness attached to an overhead trolley to ensure safety, and an examiner remained by the side of the subject to prevent a fall. Subjects were asked to try not to bend their trunk and to maintain their balance without moving their feet or taking a step. Six standing platform perturbations (nonlinear ramp-to-parabola acceleration waveforms), including one for each of the platform velocities 15, 30, 40, 50, 60, and 70 cm/s with a total backward translation of 15 cm, were conducted. Perturbations were presented in a pseudo-randomized manner whereby less severe disturbances occurred within earlier trials [32]. This testing order served as a precautionary measure to ensure that subjects would not be exposed to the more severe disturbances early in the test session. The experimental protocol was approved by the Institutional Review Board. No participants had a history or clinical evidence of neurological, musculoskeletal or other medical conditions.

Whole body motion data were captured with an eight-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA) at 120Hz. A total of 29 reflective markers were placed on each subject's bony landmarks [37]. Additionally, 4 markers were placed on the corners of the moving platform to record platform movement. Three-dimensional marker trajectory data were then low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 8Hz. Perturbation acceleration was used to describe the magnitude of balance disturbance since previous studies have shown that

platform acceleration provides the initial destabilizing input in the postural response [80-85]. Perturbation acceleration for each speed condition was calculated from the 2<sup>nd</sup> derivative of plate marker displacement. The onset and offset of platform motion were defined as first and second zero crossing events, respectively, in the backward acceleration phase of the platform movement (Fig.4.1) [80]. Ground reaction forces were collected at 960 Hz. Responses to the platform perturbation were classified as in-place, step (STEP), or heel-rise (HR). An in-place response occurred when the feet remained in contact with the supporting surface throughout the trial. A STEP response occurred if a stepping was observed. A HR response occurred if the heel was raised more than 1% of body height. Threshold perturbation acceleration for STEP and HR responses (TAccSTEP and TAccHR, respectively) were determined as the lowest magnitude where STEP or HR responses were observed for each subject.



**Figure 4.1.** Representative plate displacement and acceleration profiles for the 40 cm/s condition.

COP positions in the antero-posterior (AP) direction during the sustained maximal forward and backward leaning were measured to determine forward FBOS (FFBOS), backward FBOS (BFBOS), and total FBOS (TFBOS) [26]. FFBOS and BFBOS were measured from the ankle joint center with the forward direction as positive values, and TFBOS was the difference between FFBOS and BFBOS. Ankle plantar- and dorsi-flexor strengths (PF and DF, respectively) of the dominant leg were measured during isometric maximum voluntary contraction in a seated position at a neutral ankle position with a BIODEX dynamometer (Biodex Medical Systems, NY).

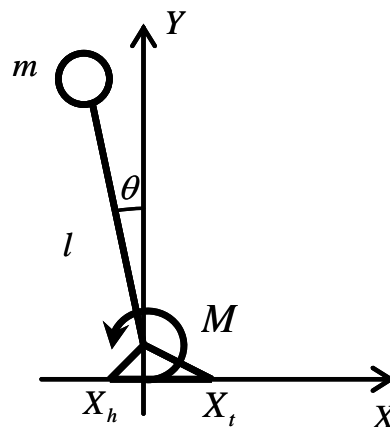
#### *Biomechanical (BIOM) Model*

Threshold perturbation acceleration for balance maintenance was determined using a simple inverted pendulum model (Fig.4.2) with the following assumptions: (1) COM passively moves in the opposite direction of plate movement during the initial backward acceleration phase of perturbation, (2) the perturbation-induced COM acceleration profile can be modeled as a triangle-shape with the peak reached in the middle, (3) the COM velocity needs to be reduced to zero at the movement termination by the COM acceleration resulting from ankle joint moment. Based on these assumptions, the following formula can be obtained (detailed derivation is presented in Appendix B):

$$\ddot{X}_p = 4(Xi - BOSf) \left\{ \frac{a}{a\Delta t^2 + \sqrt{2}\Delta t} \right\} \quad (4-1)$$

where  $a = \sqrt{g/l}$ . This formula defined the threshold acceleration of support surface translation ( $\ddot{X}_p$ ) as a function of the initial COM position ( $Xi$ ), perturbation duration  $\Delta t$  (defined by the time between perturbation onset and offset), and  $BOSf$  (distance from the heel to the forward boundary of the BOS). FFBOS was used as  $BOSf$  to predict threshold

perturbation for HR response (TAccHR), since it was measured as the range of COP excursion during the sustained maximal forward leaning without raising the heels. Foot length was used as  $BOS_f$  to predict threshold perturbation for STEP response (TAccSTEP). A two-way (2 groups, 2 responses) mixed design ANOVA with repeated measures was used to detect differences in the predicted threshold perturbations for STEP and HR responses. Group and response were between-subject and within-subject factors, respectively. Pairwise comparisons in the group-by-response interactions were analyzed with adjustments for multiple comparisons ( $\alpha = 0.05/4 = 0.0125$ ).



**Figure 4.2.** A single-link-plus-foot inverted pendulum model in the sagittal plane.  $X$  indicates the COM position in the antero-posterior direction.  $X_h$  and  $X_t$  indicate the heel and toe positions.  $m$ ,  $l$  and  $M$  are whole body mass, pendulum length (distance from the ankle to the COM), and ankle joint moment.

#### *Statistical (STATS) Model*

In addition to the biomechanical model, we used a statistical model to estimate the threshold perturbation acceleration. Linear regression equations were used to predict threshold accelerations for STEP (TAccSTEP) and HR (TAccHR). Based on our previous findings (in Chapter III) on the linear regression analyses, TFBOS (total FBOS) and DF

(ankle dorsiflexor) strength were used to predict threshold accelerations since both variables demonstrated the highest effect size for predicting TAccSTEP and TAccHR (0.32 and 0.29, respectively):

$$TAccSTEP = -35.0 \times TFBOS - 0.3 \quad (4-2)$$

$$TAccHR = -4.90 \times DF - 2.31 \quad (4-3)$$

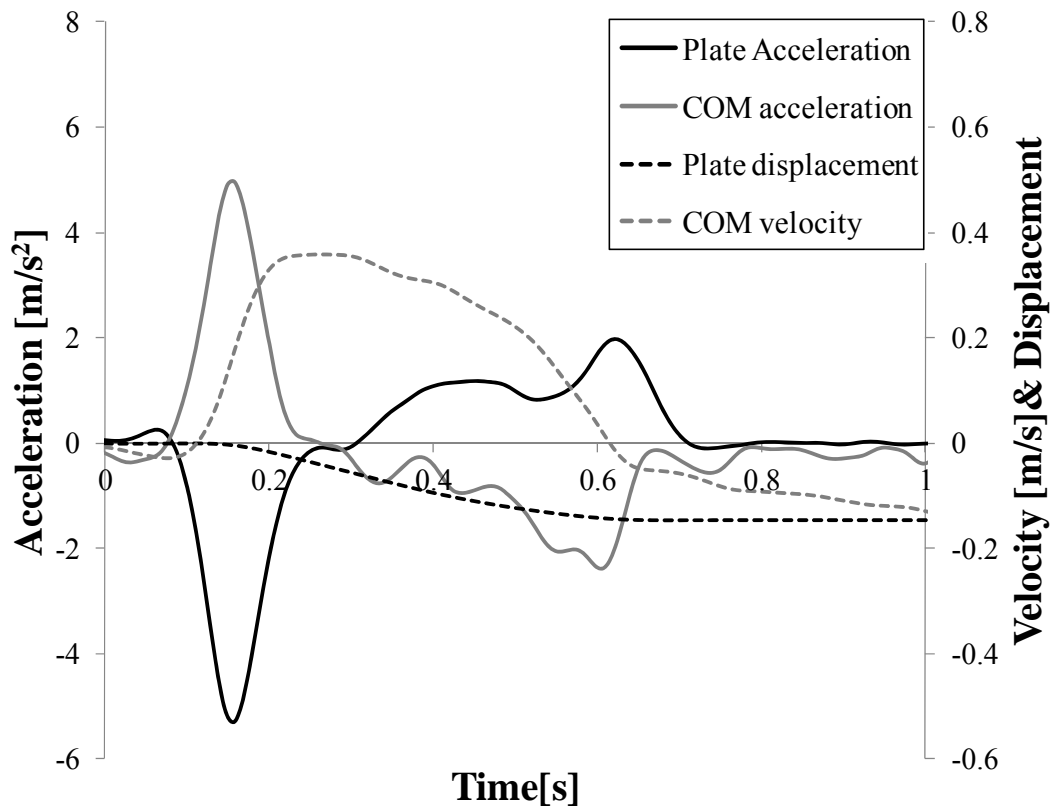
### *Predictive Performance*

Predictive performance of the biomechanical (BIOM) and statistical (STATS) models was assessed. If the measured perturbation acceleration of a trial exceeded the predicted threshold acceleration derived by these models, the trial was categorized as an “unstable” response. Otherwise, the trial was categorized as a “stable” (in-place) response. STEP or HR responses were further used to classify “unstable” trials. To assess its predictive performance, values for sensitivity and specificity were calculated. Sensitivity was defined as the proportion of true-positives (trials appropriately categorized as “unstable”, when their responses were “HR” or “STEP” (unstable trials)). Specificity was defined as the proportion of True-negatives (trials appropriately categorized as “stable”, when their responses were “in-place”).

### Results

The COM acceleration followed a similar profile to the plate acceleration but in the opposite direction (Fig.4.3). Backward acceleration of the plate induced a forward COM acceleration, generating a forward COM velocity. The COM velocity reached its peak at the end of forward COM acceleration, and then gradually decreased.





**Figure 4.3.** Representative plate displacement and acceleration, COM velocity and acceleration profiles for the 40 cm/s condition.

As the perturbation speed increased, the number of STEP and HR responses increased, except for the fastest perturbation condition for the STEP response (70cm/s) (Fig.4.4). The BIOM model predicted smaller TAccHR and TAccSTEP than the STAS model for both subject groups. Number of HR response increased when the perturbation acceleration exceeded the TAccHR predicted by the BIOM or STAS models. Increased incidence of STEP responses was observed when the perturbation acceleration exceeded the TAccSTEP predicted by the BIOM model.

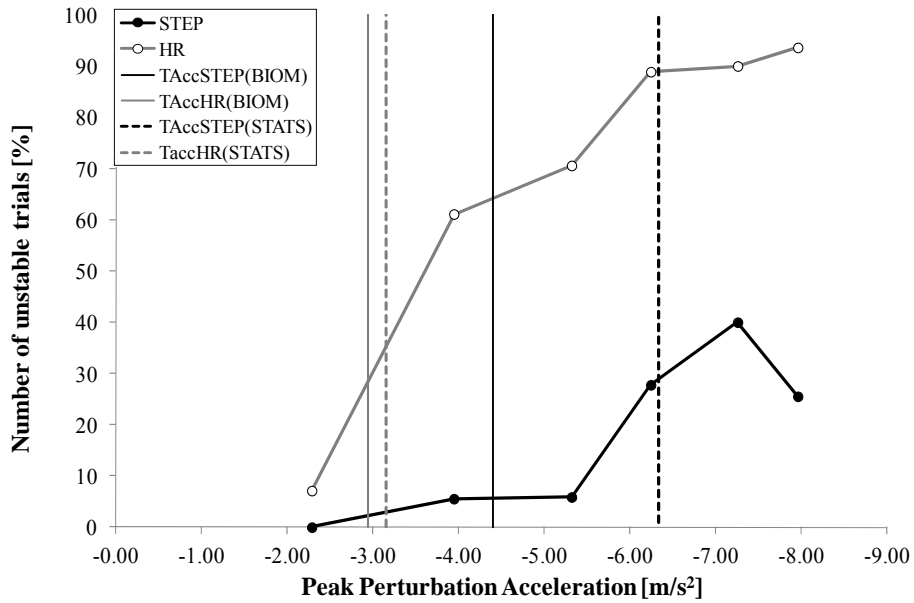
Fig.4.5 shows predicted TAccHR and TAccSTEP boundaries, which were used to categorize STEP or HR responses, as a function of the initial COM position. The

TAccHR predicted by the BIOM model was significantly smaller than the TAccSTEP for both subject groups (Fig.4.6;  $p < .001$ ; Young:  $-2.98 \pm 0.69$  v.s.  $-4.44 \pm 0.76$ ; Elderly:  $-2.30 \pm 0.67$  v.s.  $-4.20 \pm 0.49$ ), which lowered the stable threshold boundary for HR response (Fig.4.5). This reduction was more pronounced in Elderly subjects (45%) than Young subjects (33%), which resulted in a significantly smaller predicted TAccHR for elderly subjects than young subjects (Fig.4.6;  $p = .008$ ). Young subjects used a 69% greater range (0.25-0.69) for their initial COM positions as compared to elderly subjects (0.34-0.60) (Fig.4.5).

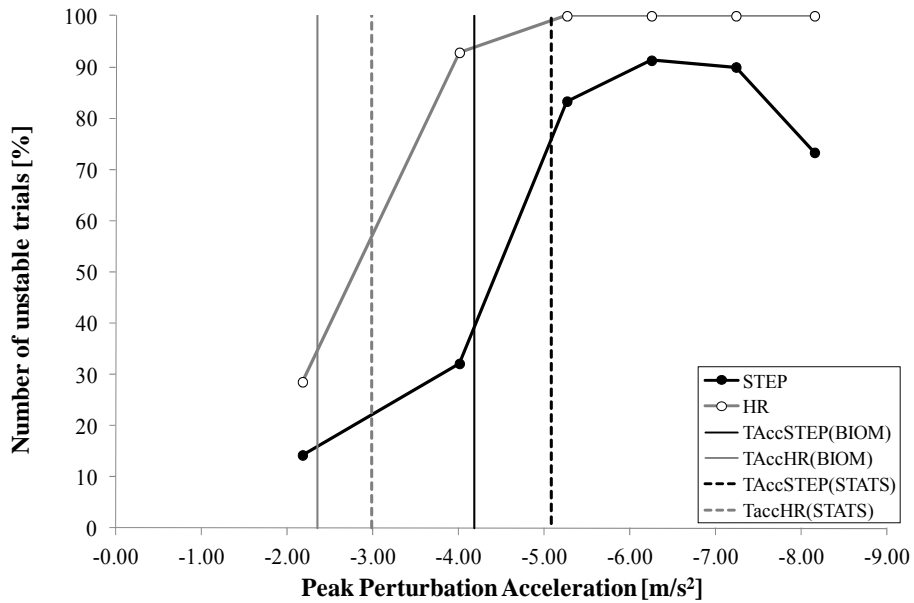
The BIOM model demonstrated higher sensitivity ( $> 89\%$ ) to predict STEP or HR response trials for both subject groups, while the STATS model showed higher specificity ( $> 54\%$ ) detecting Non-STEP and Non-HR trials, except Non-HR trials for Young subjects (Table 4.1).

### Discussion

The objectives of this study were to establish a biomechanical model that predicts threshold perturbation acceleration to maintain balance, and to compare the predictive capability of the biomechanical model with that of a statistical model. As expected, the COM acceleration induced by a backward platform perturbation mirrored the plate acceleration. Our biomechanical model showed better predictive capability than the statistical model for predicting unstable balance recovery, while the statistical model better predicted stable trials.

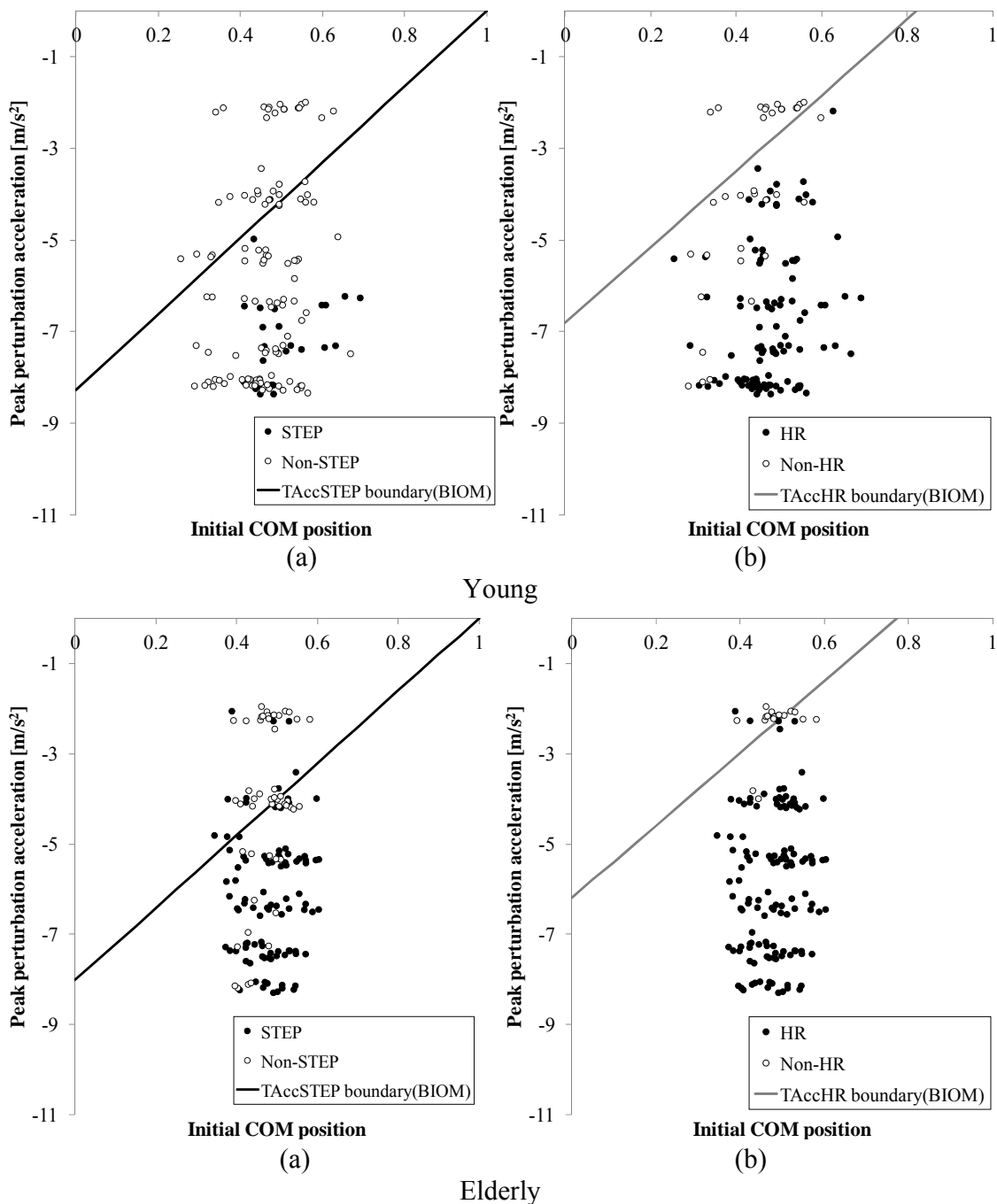


(a) Young

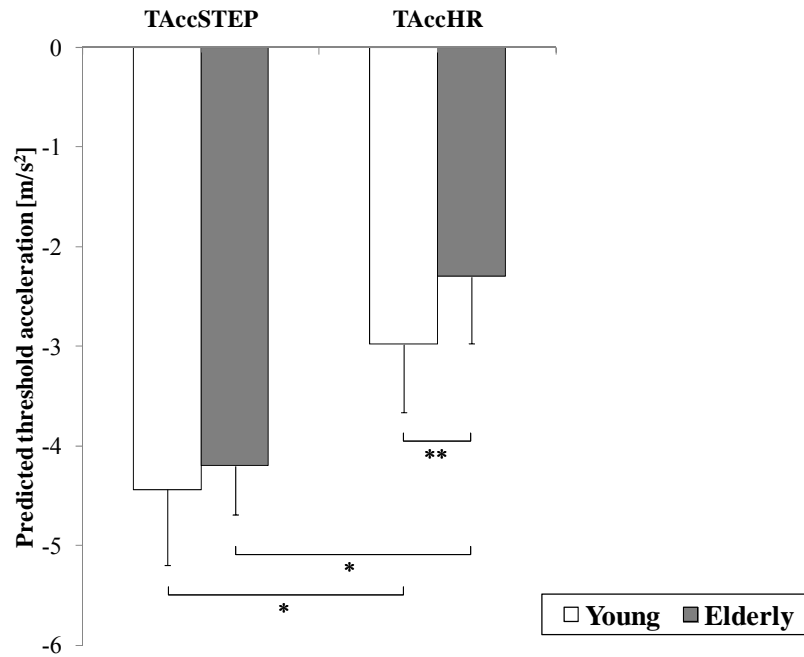


(b) Elderly

**Figure 4.4.** Number of STEP and HR responses with respect to peak backward perturbation acceleration for (a) Young and (b) Elderly subjects. Solid and dashed lines indicate mean TAccSTEP (black) and TAccHR (gray) based on the BIOM and STATS models, respectively.



**Figure 4.5.** Empirically measured perturbation accelerations with respect to the initial COM position and predicted mean threshold acceleration boundaries. Solid black and gray lines indicate mean TAccSTEP and TAccHR boundaries, respectively, predicted by the BIOM model as a function of the initial COM position. If the magnitude of measured perturbation acceleration exceeded the predicted threshold, the trial was categorized as “unstable”, i.e. STEP or HR responses. Filled and open circles indicate empirically observed (a) STEP and Non-STEP, (b) HR and Non-HR responses, respectively.



**Figure 4.6.** Threshold accelerations for STEP (TAccSTEP) and HR (TAccHR) responses predicted by the BIOM model for Young and Elderly groups. Mean value with standard deviation (error bars) for each group is also indicated. (\* $p < .001$ , \*\* $p = .008$ )

**Table 4.1**

Sensitivity and Specificity in percentage when using the biomechanical (BIOM) or statistical (STATS) model for (a) Young and (b) Elderly subjects

(a) Young

Model	Sensitivity [%]		Specificity [%]	
	STEP	HR	Non-STEP	Non-HR
BIOM	96.3	100.0	29.1	50.0
STATS	85.2	99.0	54.5	44.1

(b) Elderly

Model	Sensitivity [%]		Specificity [%]	
	STEP	HR	Non-STEP	Non-HR
BIOM	89.7	97.7	64.7	58.8
STATS	79.4	95.4	80.4	88.2

Our biomechanical model predicted the stable balance threshold based on the initial COM position and functional capability (FBOS or Foot length) of an individual. The use of FBOS for predicting HR responses demonstrated a reduction in the threshold acceleration, which was more pronounced in Elderly subjects, suggesting their reduced functional capability to regulate momentum by acceleration. Elderly subjects also demonstrated a narrow range of their initial COM positions during standing, which could also be a result of their reduced FBOS. A significant reduction in FFBOS was in fact seen in Elderly subjects ( $p=.006$ ;  $58.6\pm 9.0$  v.s.  $65.3\pm 5.1$ , %Foot length)

FFBOS was used as an allowable range for the COP movement to predict trials required a HR response. Since the FFBOS was measured as the range of COP excursion during the sustained maximal forward leaning, it is reasonable to expect the model would better predict HR responses, while using the entire foot length could better predict STEP responses. This was confirmed by an increase in overall percentage of correct predictions for STEP and HR responses when using Foot length (FL) and FBOS, respectively (STEP responses: 81%(FL) and 73%(FBOS); HR responses: 93% (FBOS) and 81% (FL), for Elderly subjects).

Our biomechanical model demonstrated less predictive performance for Non-STEP or Non-HR responses in Elderly subjects, and relatively lower specificity detecting Non-STEP response in Young subjects as compared to the statistical model. While most elderly subjects took a step once they raised their heels (74.0% of HR responses), most young subjects were able to restore their balance after heel-rise (73.8% of HR responses). The low predictive capability for Non-STEP responses in Young subjects could be due to

its limitation of not including a toe segment, where most young subjects restored their balance after HR while standing on toes.

However, our biomechanical model demonstrated a better predictive capability for predicting unstable trials than the statistical model. The consequences of a false detection in identifying an unstable trial are certainly worse than being unable to detect stable trials for safety reasons [62]. The worst scenario when predicting balance recovery responses would be that individuals lose their balance although the model predicted they would be stable, which is a false negative (1-sensitivity). While the statistical model better predicted stable trials, our biomechanical model was able to better predict unstable trials, minimizing false negatives.

Several other limitations could have affected the predictive performance of the biomechanical model. The perturbation acceleration threshold was derived from several simple assumptions. A single segment was assumed above the ankle joint and the pendulum motion was restricted to the sagittal plane [62]. The inverted pendulum model has known limitations, but it provides a good approximation of upright standing behavior in the sagittal plane, even during large body sway [93-95]. Although we asked subjects not to bend their trunk to minimize the use of a hip strategy to recover balance, it is possible that they used this kind of strategy to some extent. The model also did not have a toe segment, which could have affected its predictive capability for STEP trials especially for Young subjects. A multi-link segment model taking into account the hip joint and the toe segment would be more applicable.

In conclusion, this study developed a biomechanical model, which allowed us to predict the outcome of balance recovery based on their initial condition and functional

ability. This biomechanical model demonstrated a better predictive capability than the statistical model to identify unstable balance recovery trials. Our previous study demonstrated significant correlations between ankle dorsiflexor strength and FBOS measures (Chapter III), suggesting a potential use of dorsiflexor strength to predict perturbation speed threshold for individuals with different balance control abilities. The present biomechanical model with FBOS and ankle muscle strength measurements could serve as a valuable assessment tool to examine the dynamic balance control ability of older adults in response to unexpected perturbations and the effect of rehabilitation treatments on their balance control.

### Bridge

This chapter proposed a biomechanical model that predicts threshold perturbation acceleration to maintain upright standing, and compared its predictive capability with that of a statistical model. A single-link-plus-foot inverted pendulum model was used to determine a set of balance stability boundaries, using initial COM position in relation to the BOS. Linear regression equations were used to predict threshold accelerations for the statistical model. Our biomechanical model showed a better predictive capability than the statistical model for identifying unstable balance recovery trials, while the statistical model better predicted stable recovery trials. This biomechanical model could serve as an assessment tool for the examination of dynamic balance control ability in response to unexpected perturbations and the effect of rehabilitation treatments on balance control.

Finally, most falls resulting in fatal physical injuries occur while walking, which requires fine momentum control. The next chapter proposed a novel method to identify



dynamic limits of balance control during walking using COM acceleration, and examined differences in control of dynamic stability during walking for healthy young and elderly adults, and elderly fallers.

CHAPTER V  
CONTROL OF DYNAMIC STABILITY DURING WALKING IN ELDERLY  
FALLERS

The study described in this chapter was developed with Dr. Li-Shan Chou. Dr. Li-Shan Chou contributed substantially to the work by providing critiques about data analysis and development of methodologies. I was the primary contributor to the data collections, data analysis, model development, implementation of the procedure, and did all the writing.

Introduction

Most falls resulting in fatal physical injuries in the elderly occur while walking [33]. A better understanding of the mechanisms underlying gait imbalance is fundamental in designing risk assessments for fall prevention. The occurrence of imbalance has been traditionally regarded as a consequence of the whole body center of mass (COM) movement beyond the boundaries of the base of support (BOS). Since the BOS is defined as the possible range of the COP movement, the relationship between COM and BOS, as limits of COP control, has been investigated to assess dynamic stability. The distance between COM and COP during single limb stance has been used to quantify gait instability [41, 96, 97].

The velocity of the COM could also be a cause for imbalance as recent works have shown that compensatory steps to avoid falling depend on the interaction between the COM position and its velocity in relation to the BOS. Time-to-Contact, which uses

the instantaneous kinematics of the COM to predict when the COM will contact the BOS boundary, has been reported to characterize dynamic stability [49]. Pai and Patton (1997) used a biomechanical model to determine the allowable COM velocities at a given COM position that would permit a successful termination of COM movement within the BOS in the sagittal plane [43-45]. Yang et al. (2009) extended this concept to the frontal plane motion and described medio-lateral momentum during gait [45]. Hof and colleagues (2005) also suggested the importance of COM velocity in the examination of gait stability and derived the extrapolated center of mass (XcoM) using a simple inverted pendulum model, where the condition of dynamic stability was that the XcoM should be confined within the BOS [47, 48].

However, COM velocity, which reflects momentum, only describes an instantaneous state of motion, not providing information on how the balance is maintained by skeletal muscles. As muscle forces produce joint torques and accelerations, the COM acceleration would be directly regulated and reflect the active control of COM momentum. Elderly adults may exhibit difficulties in the regulation of COM momentum due to age-related declines in muscle function, which could lead to gait imbalance. Balance control during gait has been assessed using COM momentum, and an excessive lateral momentum was identified in balance-impaired elderly [38]. Such excessive momentum could be due to poor momentum control resulting from an inappropriate COM acceleration generation. An examination of the COM acceleration, in addition to its velocity, could help us understand how balance is controlled during locomotion, which would lead us to a better understanding of the mechanisms underlying falls during walking.

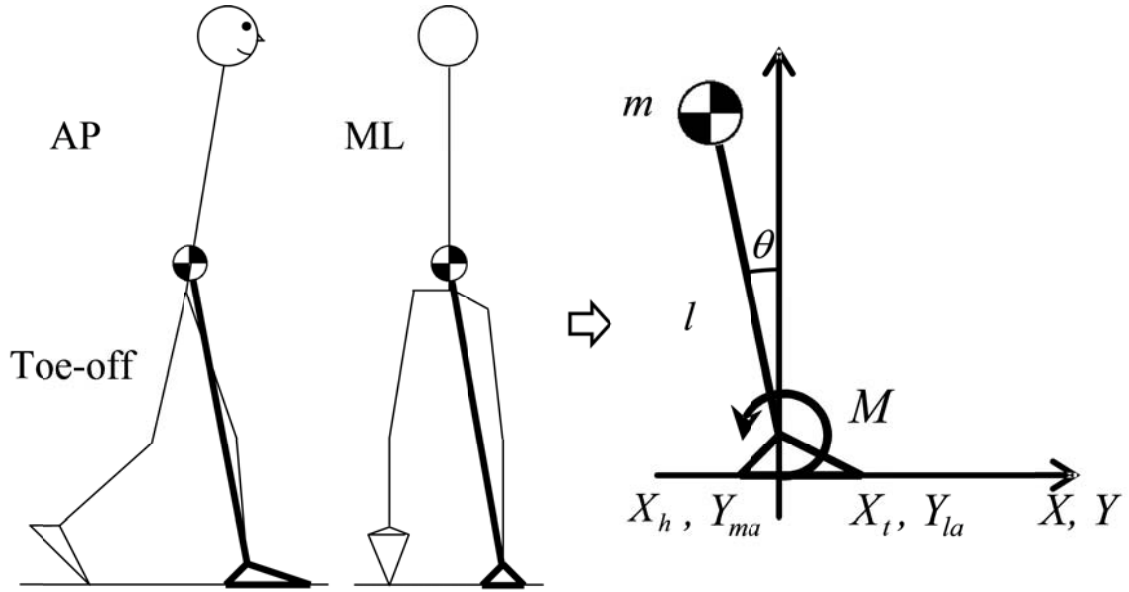
We have recently proposed that examining COM acceleration could provide further insights into balance control during sit-to-stand (STS) movement and established the region of stability using COM position and its acceleration[98]. Therefore, the objectives of this study were to expand our investigation to the COM acceleration during walking and establish regions of stability using COM velocity and acceleration for walking and to examine differences in dynamic momentum control in the antero-posterior and medio-lateral directions during walking among healthy young and elderly adults, and elderly fallers.

## Methods

### *Regions of Stability*

The regions of stability in the antero-posterior (AP) and medio-lateral (ML) directions were derived in two ways: one using COM velocity (ROS<sub>v</sub>), and the other using COM acceleration (ROS<sub>a</sub>). A single-link-plus-foot inverted pendulum model in the sagittal and frontal planes, with a mass point at the whole body COM and a center of rotation at the ankle joint, was used for analysis (Fig.5.1).

Falls most likely occur during walking due to a misplacement of the foot either in the AP or ML direction at heel-strike (HS), which could restrict an individual's ability to regulate the COM momentum. A forward/lateral balance loss would occur if the momentum is not properly controlled or an excessive momentum is generated. A balance loss would also occur after toe-off (TO), the beginning of single stance phase, if an



**Figure 5.1.** A single-link-plus-foot inverted pendulum model in the sagittal and frontal planes.  $X$  and  $Y$  indicate the COM position in the antero-posterior (AP) and medio-lateral (ML) directions.  $X_h$  and  $X_t$  indicate the heel and toe positions.  $X_{ma}$  and  $X_{la}$  indicate the medial and lateral ankles.  $m$ ,  $l$  and  $M$  are whole body mass, pendulum length (distance from the ankle to the COM), and ankle joint moment.

insufficient momentum is generated to bring the COM inside the BOS. Previous studies on the recovery step during a gait-slip in the AP direction have shown that the COM motion state in the AP direction at TO could predict recovery stepping at an accuracy of  $\sim 100\%$ , implying that outcomes at heel-strike would be highly predictable from the COM motion state at TO [45, 46, 99]. Thus, we focused on the COM motion state at/prior to the TO instant. Detailed derivation is presented in Appendix C.

### ROSV

The boundaries of the ROSv in the AP direction were defined using the following equation derived based on the concept of XcoM presented by Hof et al. [48].

$$-\tilde{X}_{TO} \leq \tilde{X}_{TO} \leq 1 - \tilde{X}_{TO} \quad (5-1)$$

where  $\tilde{X}_{TO}$  and  $\tilde{\dot{X}}_{TO}$  are normalized COM position and velocity at TO in AP direction, defined as  $\tilde{X}_{TO} = (X_{TO} - X_h) / L_f$ ,  $\tilde{\dot{X}}_{TO} = \dot{X}_{TO} / (L_f \omega_0)$  ( $\omega_0 = \sqrt{g/l}$ ,  $L_f = X_t - X_h$  : foot length).

Similarly, the ROSv in the ML direction could be obtained:

$$-\tilde{Y}_{TO} \leq \tilde{\dot{Y}}_{TO} \leq 1 - \tilde{Y}_{TO} \quad (5-2)$$

where  $\tilde{Y}_{TO}$  and  $\tilde{\dot{Y}}_{TO}$  are normalized COM position and velocity at TO in the ML direction, defined as  $\tilde{Y}_{TO} = (Y_{TO} - Y_{ma}) / Lw$ ,  $\tilde{\dot{Y}}_{TO} = \dot{Y}_{TO} / (Lw \omega_0)$  ( $\omega_0 = \sqrt{g/l}$ ,  $Lw = Y_{la} - Y_{ma}$  : ankle width,  $Y_{ma}$  and  $Y_{la}$  indicate the medial and lateral ankles).

These relationships represent the region confined by the limiting maximum and minimum COM velocities in a given COM position at TO that would permit successful termination of the movement within the BOS.

### ROSa

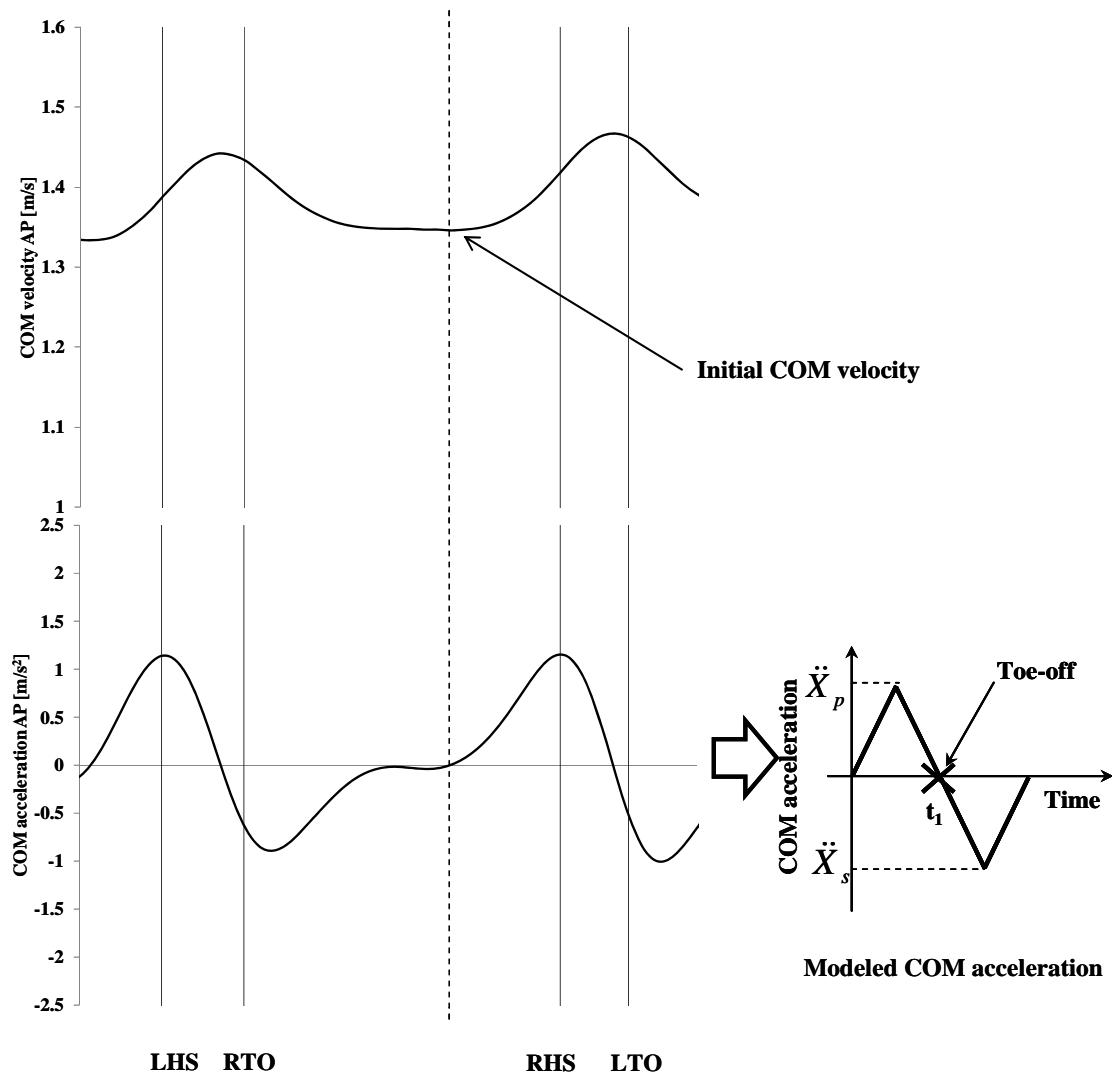
We further defined the ROSa as the region confined by peak COM acceleration that needed to be generated prior to toe-off. COM acceleration was modeled as a triangle-shape with the peak in the middle prior to toe-off. We used COM position and velocity when the COM velocity reaches its minimum prior to TO as initial conditions in the AP direction (Fig.5.2). The boundaries of the ROSa in the AP direction were derived using the following equation:

$$\frac{(\tilde{X}_{TO} + \dot{X}_i / (\omega_0 L_f))(\tilde{X}_{TO} - \dot{X}_i / (\omega_0 L_f))}{\tilde{X}_{TO} - \tilde{X}_i} < \tilde{\ddot{X}}_p < \frac{(1 - \tilde{X}_{TO} + \dot{X}_i / (\omega_0 L_f))(1 - \tilde{X}_{TO} - \dot{X}_i / (\omega_0 L_f))}{\tilde{X}_{TO} - \tilde{X}_i} \quad (5-3)$$

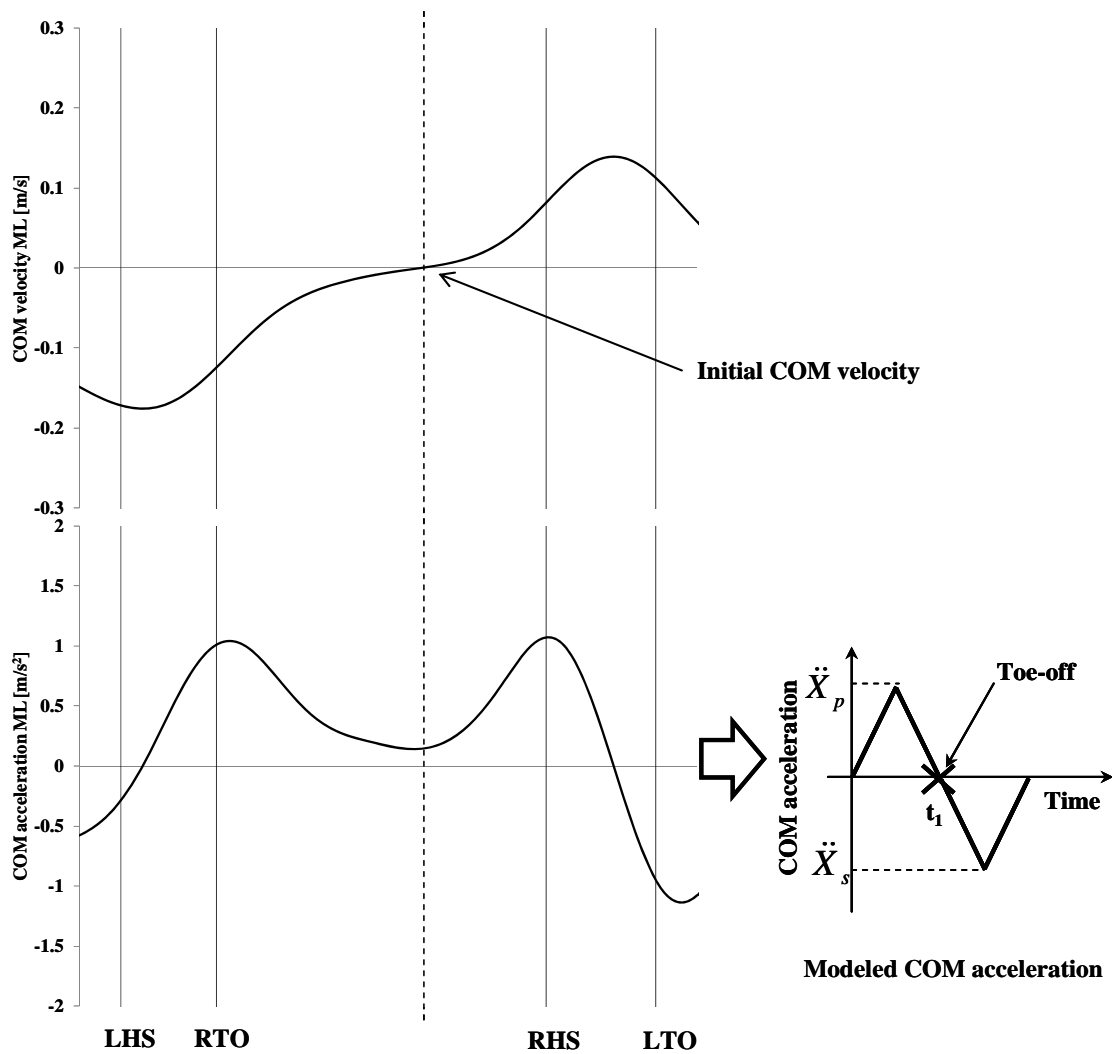
where  $\tilde{X}_{TO}$  and  $\tilde{X}_i$  are the normalized COM position at TO and initial COM position defined as  $\tilde{X}_{TO} = X_{TO} / L_f$  and  $\tilde{X}_i = X_i / L_f$ , respectively.  $\tilde{\ddot{X}}_p$  is the normalized peak COM acceleration prior to TO defined as  $\tilde{\ddot{X}}_p = \ddot{X}_p / \omega_0^2 L_f$ .

Similarly, we used COM position when the COM velocity becomes zero prior to TO as initial COM position ( $Y_i$ ) in the ML direction (Fig.5.3). The boundaries of the ROSa in the ML direction were defined using the following equation:

$$-\frac{Y_{ma} - Y_{TO}}{Y_{TO} - Y_i} \tilde{Y}_{TO} \leq \tilde{Y}_p \leq \frac{Y_{la} - Y_{TO}}{Y_{TO} - Y_i} (1 - \tilde{Y}_{TO}) \quad (5-4)$$



**Figure 5.2.** Representative time-history plot of COM velocity and acceleration in the antero-posterior (AP) direction and modeled COM acceleration profile. LHS/RHS and LTO/RTO indicate left/right heel-strike and toe-off instants.



**Figure 5.3.** Representative time-history plot of COM velocity and acceleration in the medio-lateral (ML) direction and modeled COM acceleration profile. LHS/RHS and LTO/RTO indicate left/right heel-strike and toe-off instants.

where  $\tilde{y}_p$  is the normalized peak COM acceleration defined as  $\tilde{y}_p = \ddot{Y}_p / \omega_0^2 L_w$ ,  $Y_{ma}$  and  $Y_{la}$  indicate the medial and lateral ankles, which are considered as the medial and lateral boundaries of the BOS, respectively.



### *Experimental Protocol*

Fifteen healthy young adults [Young: 7 men/8 women; mean age  $22.1 \pm 1.9$  years, mean height  $170.4 \pm 11.0$  cm, mean mass  $68.2 \pm 14.6$  kg], 15 healthy elderly adults [Elderly: 6 men/9 women; mean age  $70.0 \pm 3.2$  years, mean height  $170.1 \pm 8.7$  cm, mean mass  $79.1 \pm 18.3$  kg], and 15 elderly adults with a history of falls [Fallers: 3 men/12 women; mean age  $71.9 \pm 4.3$  years, mean height  $164.2 \pm 8.6$  cm, mean mass  $83.1 \pm 20.1$  kg] participated in this study. The criterion for inclusion in the faller category was a self-report of two or more falls within the year prior to the study [100]. A fall was defined as any event that led to an unplanned, unexpected contact with a supporting surface. All participants did not have a history or clinical evidence of neurological, musculoskeletal or other medical conditions. All participants reported no history of neurological pathology, head trauma, cerebrovascular accident, vestibular dysfunction, or visual impairment uncorrectable by lenses. The experimental protocol was approved by the Institutional Review Board. Written and verbal instructions of testing procedures were provided, and written consent was obtained from each subject prior to testing.

Subjects were instructed to walk barefoot at a self-selected comfortable pace along a 10-m unobstructed walkway. Walking trials with data collection were started after subjects became familiar with the laboratory setting by performing a few practice trials. Data from 4-6 trials were collected for each subject. Whole body motion was captured with an eight-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA). A total of 29 reflective markers were placed on the subject [37]. Three-dimensional marker trajectories were collected at 60Hz and smoothed using a fourth-order Butterworth filter with a cut-off frequency of 8Hz. The whole-body COM was

calculated as the weighted sum of 13 body segments, including head and neck, trunk, pelvis, 2 upper arms, 2 forearms with hands, 2 thighs, 2 shanks, and 2 feet [37]. Anthropometric reference data were adopted from the initial work of Dempster [34]. COM velocity and acceleration were calculated using Woltring's cross validated spline algorithm from the COM position [101]. TO (Toe-off) and HS (Heel-strike) instants were detected based on the vertical velocity of the midfoot [102, 103].

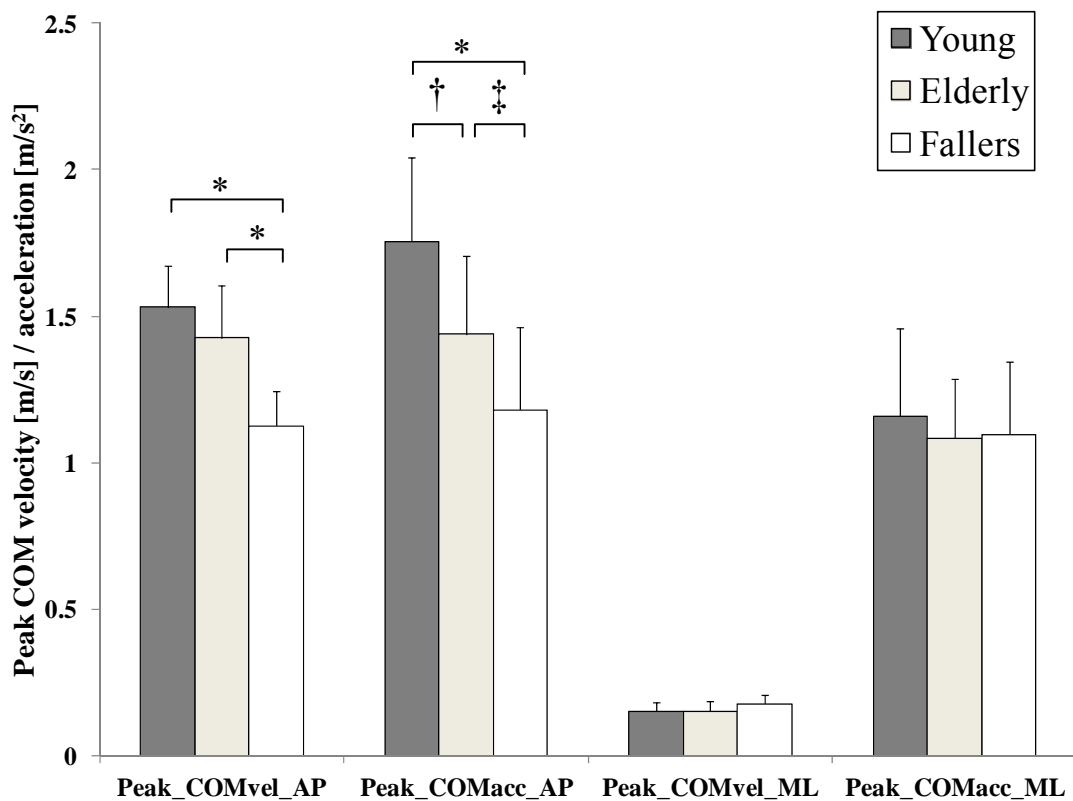
The COM position was referenced to the heel and medial ankle positions and normalized to the foot length and width in AP and ML directions, respectively. Peak COM velocity and acceleration of the gait cycle in AP and ML directions were obtained. Normalized COM position and velocity at TO, and normalized peak COM acceleration prior to TO were identified to construct the ROSv and ROSa. The magnitude of stability was defined as the shortest distance from the experimental data to the forward and lateral boundary of the ROSv and ROSa in AP and ML directions, respectively (Stability margins) [45].

One-way ANOVA was used to detect group differences among Young, Elderly and Fallers groups. Post-hoc analyses used *t*-tests based on Tukey's HSD method to determine the sources of significance. Statistical analysis was performed using SPSS (Chicago, IL). Significance level was set at  $\alpha=0.05$ .

## Results

No significant differences were found in the body height, weight, and foot length and width for both sides among Young, Elderly, and Fallers groups. Fallers demonstrated significantly smaller peak COM velocity than both Young and Elderly groups in the AP

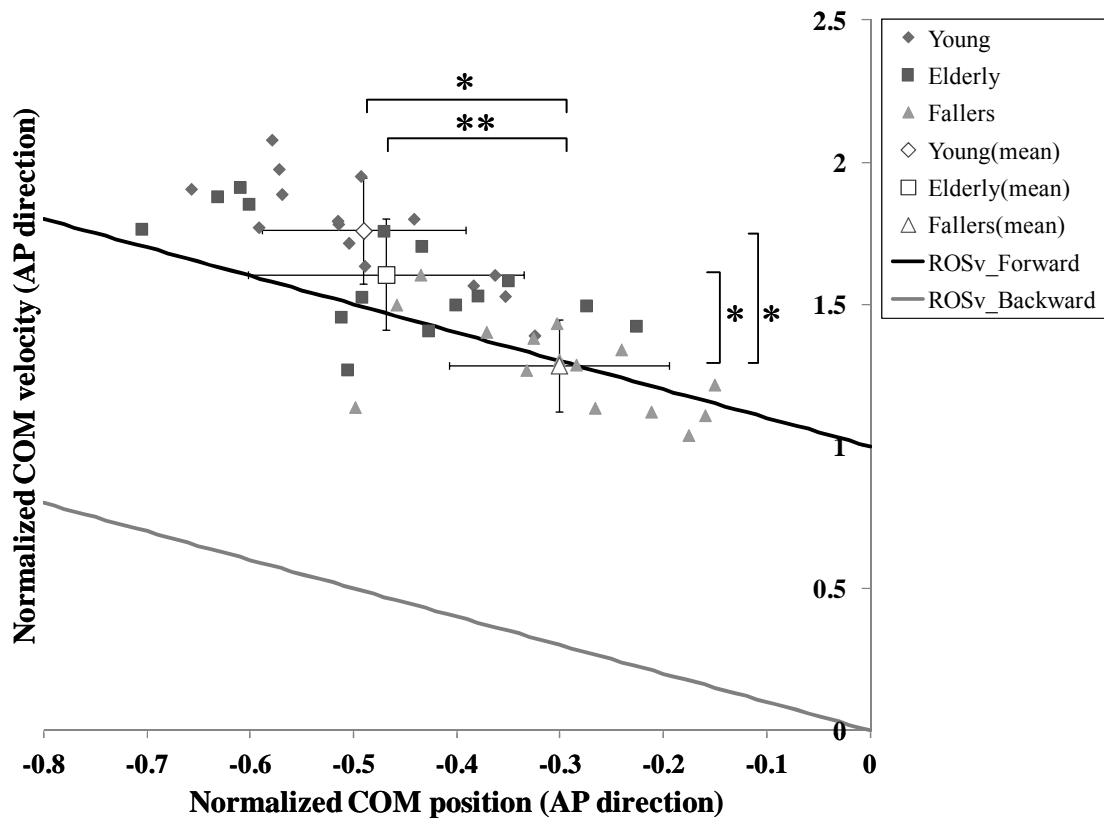
direction ( $p \leq .001$ ; Fig.5.4). The same trend was seen in the average COM velocity (i.e., gait velocity:  $p \leq .001$ ; Young:  $1.39 \pm 0.12$  m/s; Elderly:  $1.29 \pm 0.17$  m/s; Fallers:  $1.03 \pm 0.11$  m/s). Peak COM acceleration in the AP direction differed significantly among all three groups ( $p \leq .042$ ). No significant group differences were found in the ML direction, although Fallers showed larger peak COM velocity than Young ( $p = .068$ ) and Elderly ( $p = .057$ ) groups in the ML direction.



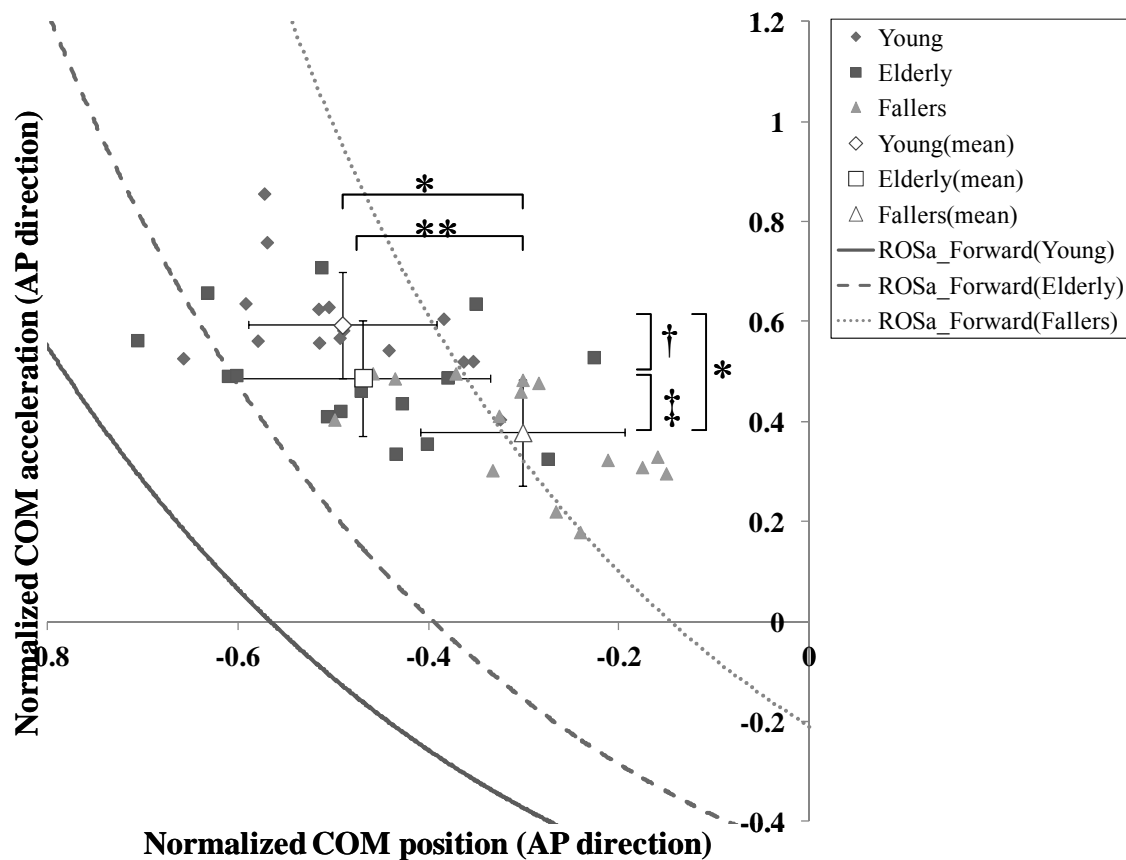
**Figure 5.4.** Peak COM velocity (COMvel) and acceleration (COMacc) in AP and ML directions for Young, Elderly, and Fallers. Values are mean  $\pm$  SD. (\* $p < .001$ , † $p = .011$ , ‡ $p = .042$ )

Normalized COM velocity at TO, normalized peak COM acceleration prior to TO with respect to normalized COM position at TO and boundaries of the ROSv and ROSa, which serve as balance stability boundaries, were plotted in Fig.5.5 and Fig.5.6 for the

AP direction and Fig.5.7 and Fig.5.8 for the ML direction, respectively. Data from all trials of Young group (84 trials) were located outside the forward boundary of the ROSv, so were 78% (65 out of 83 trials) and 52% (46 out of 88 trials) from Elderly and Fallers groups (Fig.5.5). None of the data were located outside the lateral boundary of the ROSv, but 4% (3 out of 84), 5% (4 out of 83), and 10% (9 out of 88) from Young, Elderly, and Fallers groups were located outside the medial boundary, respectively (Fig.5.7). Similarly,

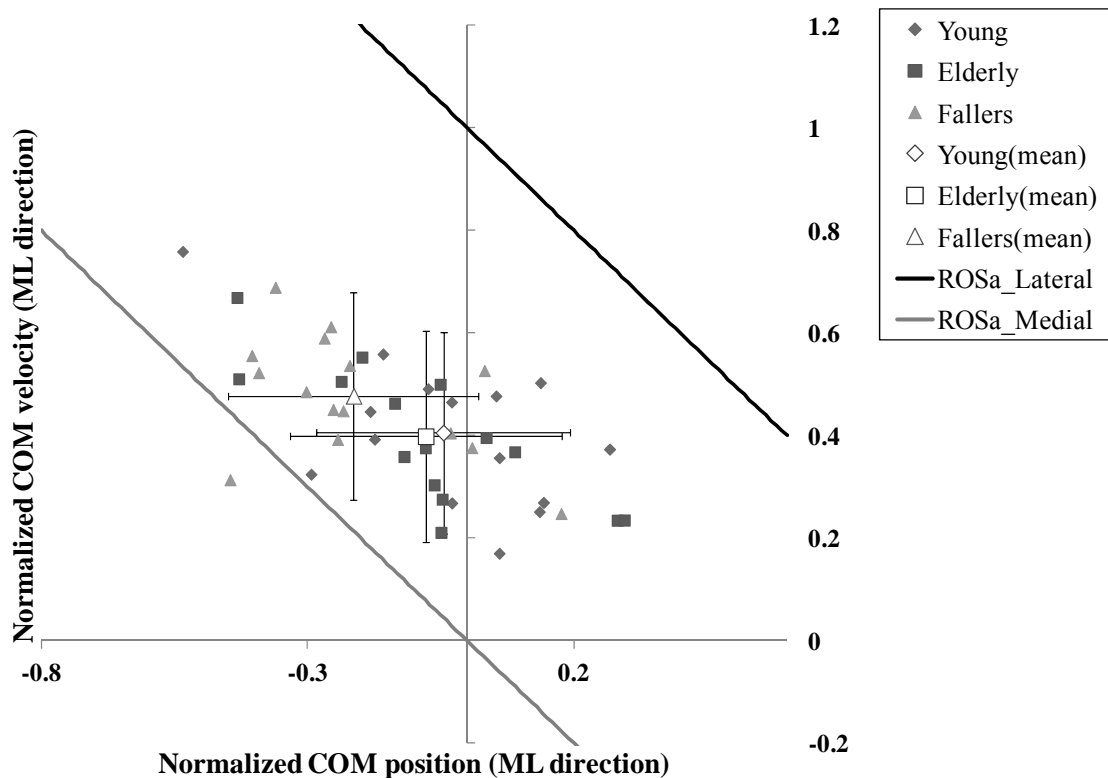


**Figure 5.5.** Normalized COM velocity at LTO (left toe-off) with respect to normalized COM position at LTO in AP direction for Young, Elderly, and Fallers. Each plot indicates the mean data for each subject. Mean  $\pm$  SD for each group is also indicated. Black and gray solid lines indicate the forward and backward boundaries of the ROSv, respectively. (\* $p < .001$ , \*\*  $p = .001$ )



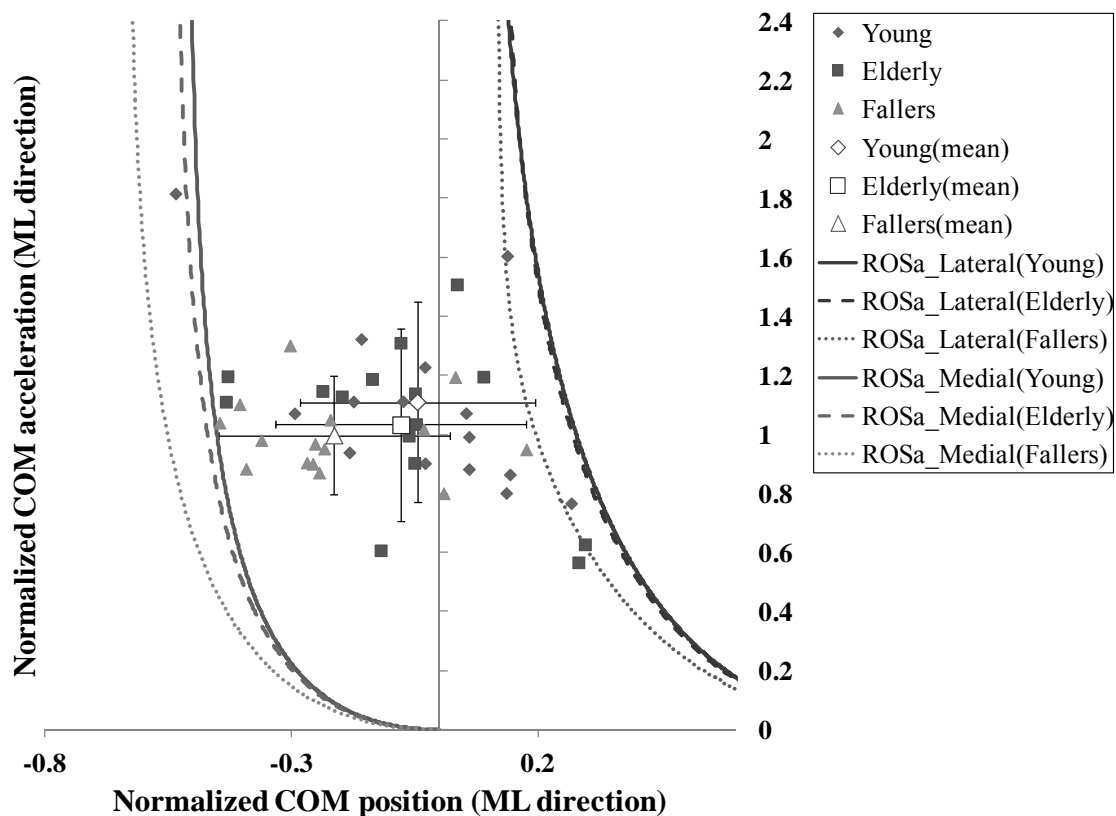
**Figure 5.6.** Normalized peak COM acceleration prior to LTO with respect to normalized COM position at LTO in AP direction for Young, Elderly, and Fallers. Each plot indicates the mean data for each subject. Mean  $\pm$  SD for each group is also indicated. Solid and dashed curves indicate the forward boundaries of the ROSa for each subject group. Since the boundaries are dependent on the initial COM position varied between subjects and trials, those boundaries were averaged for each group. (\* $p < .001$ , \*\*  $p = .001$ , † $p = .030$ , ‡ $p = .024$ )

data from all trials of Young group (84 trials) were located outside the forward boundary of the ROSa, so were 87% (72 out of 83 trials) and 66% (58 out of 88 trials) from Elderly and Fallers groups (Fig.5.6). None of the data was located outside the boundaries of the ROSa, except 5% (4 out of 88) from Fallers group, which was located outside the medial boundary (Fig.5.8).



**Figure 5.7.** Normalized COM velocity at LTO with respect to normalized COM position at LTO in ML direction for Young, Elderly, and Fallers. Each plot indicates the mean data for each subject. Mean  $\pm$  SD for each group is also indicated. Black and gray solid lines indicate the forward and backward boundaries of the ROSv, respectively.

Mean normalized COM position at TO for Fallers group ( $-0.30 \pm 0.11$ ) was found to be significantly anterior to those of Young and Elderly groups ( $p \leq .001$ ; Young:  $-0.49 \pm 0.10$ ; Elderly:  $-0.47 \pm 0.13$ ). No significant difference was detected between Young and Elderly groups. Mean normalized COM velocity at TO for Fallers ( $1.29 \pm 0.16$ ) was also significantly smaller than both Young and Elderly groups ( $p < .001$ ; Young:  $1.76 \pm 0.19$ ; Elderly:  $1.61 \pm 0.20$ ). Mean normalized peak COM acceleration prior to TO in the AP direction differed significantly among all three groups ( $p \leq .030$ ; Young:  $0.59 \pm 0.11$ ; Elderly:  $0.49 \pm 0.12$ ; Fallers:  $0.38 \pm 0.11$ ).

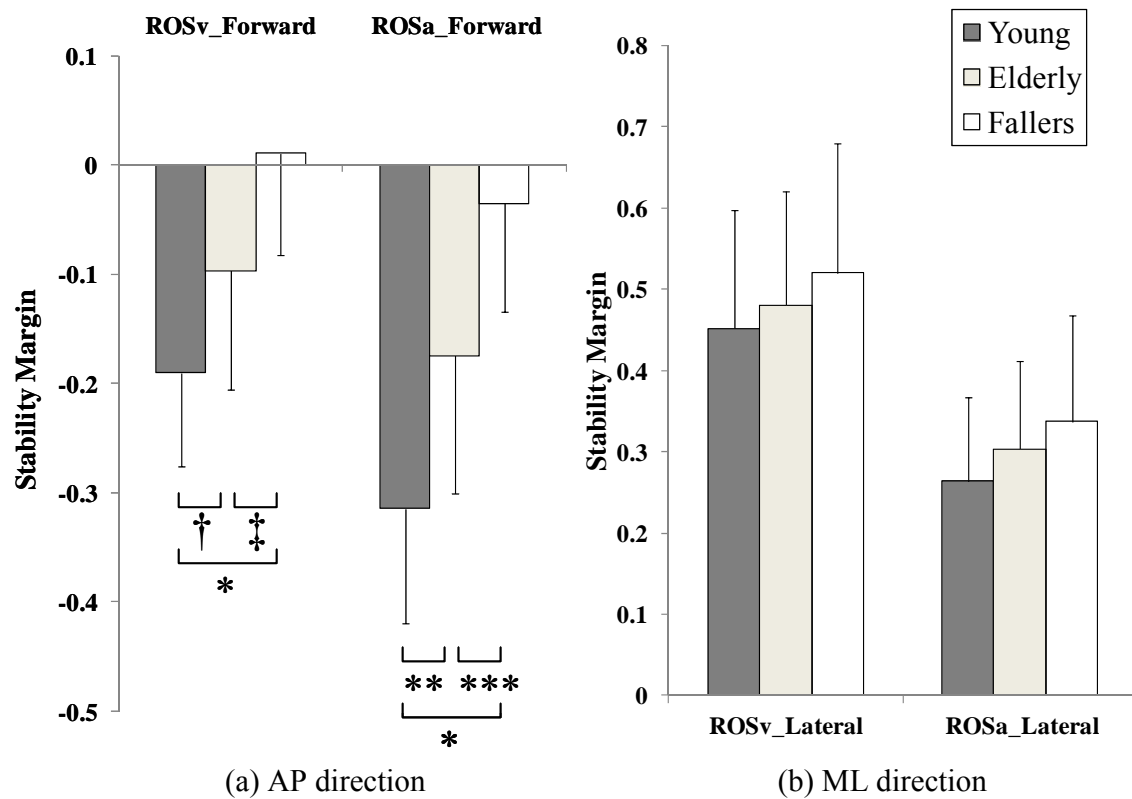


**Figure 5.8.** Normalized peak COM acceleration prior to LTO with respect to normalized COM position at LTO in ML direction for Young, Elderly, and Fallers. Each plot indicates the mean data for each subject. Mean  $\pm$  SD for each group is also indicated. Black and gray curves indicate the lateral and medial boundaries of the ROSa for each subject group. Since the boundaries are dependent on the initial COM position varied between subjects and trials, those boundaries were averaged for each group.

In the ML direction, no significant group differences were found in mean normalized COM position (Young:  $-0.05 \pm 0.24$ ; Elderly:  $-0.08 \pm 0.21$ ; Fallers:  $-0.21 \pm 0.18$ ), mean normalized COM velocity (Young:  $0.41 \pm 0.20$ ; Elderly:  $0.40 \pm 0.13$ ; Fallers:  $0.48 \pm 0.12$ ), or mean normalized peak COM acceleration (Young:  $1.10 \pm 0.29$ ; Elderly:  $1.04 \pm 0.27$ ; Fallers:  $0.99 \pm 0.13$ ).

Stability margins based on the ROSv and ROSa both differed significantly among all three groups in the AP direction ( $p \leq 0.028$  for ROSv;  $p \leq 0.004$  for ROSa; Fig.5.9). No

significant group differences were detected in Stability margins in the ML direction, although Fallers showed a larger stability margin than Young group ( $p=.065$ ).



**Figure 5.9.** Stability margins based on the ROSv and ROSa for Young, Elderly, and Fallers in (a) AP and (b) ML directions. Stability margin was defined as the shortest distance from the experimental data to the forward and lateral boundary of the ROSv and ROSa in AP and ML directions, respectively. Values are mean  $\pm$  SD. (\* $p < .001$ , \*\* $p = .003$ , \*\*\* $p = .004$ , † $p = .028$ , ‡ $p = .011$ )

### Discussion

The purpose of this study was to examine the control of COM momentum during the single support phase of walking, with the use of regions of stability derived by COM velocity and acceleration (ROSv and ROSa). Our results revealed that there was a significant difference in peak forward COM acceleration between healthy young and



elderly subjects although no significant difference was found in forward COM velocity. Elderly fallers used a more conservative strategy demonstrating significantly decreased forward COM velocity and acceleration, and shortened distance between the COM and BOS at toe-off in the AP direction.

Elderly fallers showed significantly smaller average and peak forward COM velocities than both healthy young and elderly subjects, while no significant differences were detected between healthy groups, in agreement with previous reports [37, 41, 103]. However, the peak forward COM acceleration differed significantly between healthy young and elderly groups, suggesting that even though similar momentum was observed during walking, the momentum could be controlled differently between individuals with different balance control abilities.

Demonstrating larger stability margins, elderly fallers showed a more conservative gait strategy than healthy subjects in the AP direction. The COM positions of elderly fallers were located significantly anterior to those for healthy young and elderly adults, with significantly smaller forward COM velocity at toe-off, which placed their mean data point inside the boundaries of the ROSv. In contrast, the mean data points for healthy subjects were located outside the forward boundary of the ROSv (all the data and 78% of the data for healthy young and elderly subjects, respectively). These results imply that healthy subjects, on average, were dynamically unstable at the toe-off instant during walking, indicating they would fall forward unless they take a step forward at this instant. They utilized a forward momentum to maintain a forward progression of the whole body. On the other hand, more than half of the trials (52%) for elderly fallers were located inside the boundaries of the ROSv, suggesting that they could terminate walking

without taking a step forward at this instant. Elderly fallers appeared to use a more conservative strategy in their sagittal plane momentum control.

Similar results were obtained in the ROSa in the AP direction, where the data for elderly fallers were significantly closer to the forward boundary of the ROSa, indicated by significantly larger stability margins. In agreement with our findings for peak COM velocity and acceleration, normalized peak COM acceleration prior to toe-off differed significantly among all three groups, although no significant difference was found between healthy groups in normalized COM velocity at toe-off as seen in the ROSv. Healthy young and elderly subjects both utilized a similar momentum to propel their body forward at toe-off, but controlled the momentum differently prior to toe-off, possibly due to age-related decline in muscle strength. Healthy elderly adults are known to have decreased muscular strength and difficulties in generating muscle torques at a higher rate [18, 50-55, 104], indicating less ability to regulate momentum by acceleration produced by skeletal muscles. Our muscle strength assessment (maximum isometric voluntary contraction) in fact revealed significantly weaker knee extensor strength in Elderly ( $0.98 \pm 0.38$  Nm/kg) and Fallers ( $1.06 \pm 0.23$  Nm/kg) groups as compared to Young group ( $p \leq .002$ :  $1.55 \pm 0.42$  Nm/kg). Significantly smaller peak COM acceleration demonstrated by our elderly subjects could be indicative of their poor momentum control ability, which was more pronounced in elderly fallers, resulting in significantly smaller COM velocity than healthy subject groups. COM acceleration could sensitively reflect their momentum control ability, which would allow us to better distinguish differences in balance control and strategies used for age-related gait adaptations.

Significantly decreased forward COM velocity, acceleration and separation between the COM and BOS at toe-off in the AP direction observed in elderly fallers could be indicative of their reduced momentum control ability and a protective strategy for potential falls as well as reduced muscular strength. Tripping over obstacles during gait has been reported as a common cause of falls in the elderly [3, 105, 106], and healthy elderly adults have also demonstrated an increased risk for obstacle contact [107, 108]. Slower COM velocity could serve to reduce the severity of a forward fall due to a trip, decreasing the forward momentum of the body. Slips were also found to be a common triggering event for falls [109], and increased step length has been shown to result in increased probability of hazardous slips [110]. Shorter separations between the COM and BOS could reduce slip severity. Shorter COM-BOS separations with a smaller COM velocity would also decrease mechanical demands of gait, reducing moment arms for the body weight about the joints of the supporting limb and thereby muscular efforts [37]. However, significantly smaller COM acceleration could indicate elderly fallers' inability to properly control momentum, which could result in imbalance in response to a sudden change in momentum induced by external perturbations, such as trips or slips. This inability might predispose them to a greater risk of falling.

Stability margins in the AP direction differed significantly among all three groups with the difference more emphasized when the ROSa was used, although a similar COM position in relation to the BOS and its velocity were observed at toe-off in healthy young and elderly adults. These results suggest a potential use of the regions of stability to differentiate individuals with different balance control abilities.

No significant differences were detected in COM position at TO, COM velocity (peak/at TO), peak COM acceleration, or Stability margins in the ML direction, in contrast to the previous reports where elderly adults with balance impairments demonstrated a significantly greater and faster COM motion in the ML direction than healthy controls [35, 38, 41]. However, elderly fallers still showed larger peak COM velocity than healthy individuals in the ML direction although the differences were not significant ( $p \geq .057$ ). Mean normalized COM position at heel-strike instant for Fallers group ( $-0.46 \pm 0.23$ ) was also found to be significantly medial to those of Young and Elderly groups ( $p = .035$  with Young:  $-0.23 \pm 0.25$ ;  $p = .049$  with Elderly:  $-0.24 \pm 0.26$ ). It should be noted that similar COM position and its velocity at TO were observed in healthy young and elderly adults, whereas the COM was medially located with larger velocity for elderly fallers (Fig.5.7). Increased laterally-directed COM velocity appeared to compensate for an increased medial deviation of the COM from the BOS, confining the COM within the region of dynamic stability (ROSV), which resulted in the same stability margins as healthy individuals. These findings could suggest that elderly fallers might not have any balance problems in the ML direction.

One possible limitation of the study is that the boundaries of regions of stability were derived from several simple assumptions, which could limit the predictive capacity of the boundaries. We used a single segment connecting the COM from the ankle to represent the human body. Hip flexors/extensors and abductors/adductors play an important role of dynamic balance of the upper body segment, including the head, arms, and trunk, in the AP and ML directions, respectively, during walking [77]. The inverted pendulum model with the ankle joint only may not be sufficient enough to account for

dynamic balance control during walking. This would contribute to the predictive capability of the boundaries, which could explain our findings of no significant differences in stability margins in the ML direction. However, despite using a simplified pendulum motion from the ankle, the empirically measured COM positions and velocities during walking were adequately illustrated by the feasible stability region (FSR), which is similar to our ROS<sub>v</sub>, in both AP and ML directions and the model predictions from the FSR were consistent with experimental observations in response to slips induced during walking [43, 45, 74, 111]. The model used to define the ROS<sub>a</sub> also simply assumed the COM only undergoes forward/lateral acceleration prior to toe-off. This assumption would provide conservative estimation of the boundaries, where the subjects could have generated backward/medial acceleration to reduce the momentum prior to toe-off.

In conclusion, this study examined control of dynamic stability during walking, using regions of stability derived by COM velocity and acceleration. Healthy young and elderly subjects utilized similar momentum to propel the body forward, but controlled the momentum differently. Our results revealed significant differences in control of dynamic stability for elderly fallers in the AP direction, but there were no indications of instability in the ML direction. Elderly fallers adapted a more conservative strategy with significantly smaller forward COM velocity and acceleration, and smaller COM-BOS separation at toe-off in the AP direction. This conservative strategy may be indicative of their reduced momentum control ability and a protective strategy for potential falls as well as reduced muscular strength. Examining COM acceleration in addition to its velocity would provide a greater understanding of a person's momentum control, which would allow us to better understand the mechanisms underlying imbalance or falls.

## CHAPTER VI

### CONCLUDING SUMMARY

#### Main Findings

The overall goal of this dissertation study was to develop biomechanical models focusing on COM acceleration that identify dynamic limits of balance control in daily functional activities associated with falling, including sit-to-stand (STS) movement, standing (stance perturbation), and walking. Those models would allow us to identify individuals with a higher potential risk of falling.

Poor performance of STS movement has been identified as one of the risk factors of falls among elderly individuals. The first study established regions of stability during STS based on the COM position at seat-off and its instantaneous velocity (ROSV) or its peak acceleration (ROSA), and assessed feasibility of these regions to differentiate individuals with difficulty in STS movement from healthy individuals. Differences among individuals were more clearly distinguished with COM acceleration. Healthy young and elderly adults were found to demonstrate similar strategies but with different COM accelerations, so were elderly adults with difficulty in STS and healthy young adults with exaggerated trunk flexion. Although similar momentum was observed, the momentum was controlled differently during STS movement. Elderly adults with difficulty in STS showed significantly smaller stability margin than the healthy young and elderly groups when the ROSA was used, even though there were no detectable differences in the ROSV. These findings suggested that the ROSA could provide further insights into how the momentum is controlled prior to seat-off, which allows us to

differentiate individuals with difficulty in STS movement, who are most likely at a risk for imbalance or falls.

Poor postural control ability is also one of the risk factors in the increased incidence of falls in the elderly. The second study examined postural recovery responses to backward support surface translations during quiet standing for healthy young and elderly adults. To identify biomechanical factors affecting postural recovery responses, we first investigated the relationship between ankle muscle strength and functional BOS (FBOS: the effective area for COP movement) as well as perturbation acceleration threshold that required a heel-rise (HR) or step (STEP) to maintain balance. Ankle dorsiflexor strength was found to be significantly correlated with all the FBOS measures and threshold acceleration for HR. Significant correlations were also found between all FBOS measures and threshold accelerations for HR and STEP, except for the backward FBOS and threshold acceleration for STEP. The results suggested that weakness in ankle dorsiflexors could limit the ability of elderly adults to restore balance and that FBOS and ankle dorsiflexor strength could be sensitive measures to detect elderly individuals with declined balance control. Accordingly, we established a biomechanical model that predicts threshold perturbation acceleration to maintain upright standing, and compared its predictive capability with that of a statistical model. A single-link-plus-foot inverted pendulum model was used to determine a set of balance stability boundaries, using initial COM position in relation to the BOS. Linear regression equations were used to predict threshold accelerations for the statistical model. The results of the biomechanical models demonstrated that the use of FBOS for predicting HR responses decreased the balance stability boundaries, which was more pronounced in elderly subjects, indicating their

reduced functional ability to recover balance. Our biomechanical model showed a better predictive capability than the statistical model for identifying unstable balance recovery trials, while the statistical model better predicted stable recovery trials. This biomechanical model could serve as an assessment tool to examine one's dynamic balance control ability in response to unexpected perturbations and the effect of rehabilitation treatments on their balance control.

Lastly, walking requires a fine momentum control where COM acceleration could play an important role. The third study established regions of stability during walking based on the COM position at toe-off and its instantaneous velocity (ROS<sub>v</sub>) or its peak acceleration (ROS<sub>a</sub>), and examined differences in control of dynamic stability in the antero-posterior (AP) and medio-lateral (ML) directions during walking for healthy young and elderly adults, and elderly fallers. Although no significant difference in forward COM velocity was detected between healthy young and elderly subjects, the peak forward COM acceleration differed significantly, suggesting age-related differences in momentum control during walking. Elderly fallers adapted a more conservative gait strategy, demonstrating significantly smaller forward COM velocity and acceleration with their COM significantly closer to the base of support at toe-off, which placed their mean data point inside the boundaries of the ROS<sub>v</sub>. Similar results were obtained in the ROS<sub>a</sub>, indicated by the data located significantly closer to the forward boundary. No significant differences were detected in any of the COM measures in the ML direction. Significantly smaller peak COM acceleration could be indicative of poor momentum control ability, which was pronounced in elderly fallers. These results suggested that examining COM acceleration in addition to its velocity would provide a greater



understanding of a person's momentum control, which would allow us to better understand the mechanisms underlying imbalance or falls.

Overall, the most appealing discovery through this dissertation work was the importance of COM acceleration as a potential measure to better differentiate individuals with different balance control abilities. Dynamic balance control has been traditionally assessed in terms of the interaction between COM position and its velocity, where insufficient momentum could be a trigger for imbalance. This dissertation study demonstrated that COM acceleration would provide further information on their momentum control, which could differentiate individuals with functional limitations from healthy individuals. An examination of COM acceleration, in addition to its velocity, could help us understand how balance is controlled during the movement, which could better reveal underlying mechanisms causing imbalance and provide an insightful evaluation of balance dysfunction.

#### Limitations of the Study

Several limitations exist in this study. Sample size would be a concern for any human testing. The number of subjects in the study may have limited the statistical power to detect group differences. For the second and third studies, more than 15 subjects were involved for each subject group, which was thought to be a reasonable sample size for studies of this kind. However, our first study regarding STS movement had only 10 subjects for each subject group, which was partially due to difficulty to gather a large portion of elderly subjects who fell into the category of having difficulty in STS without having neurological, musculoskeletal or other medical conditions. Nevertheless, we

believe our sample size was enough to clearly represent the population since it produced statistical significance, although a larger number of subjects might have revealed undetected differences between elderly with difficulty and young adults with exaggerated trunk flexion. Future studies should include an increased sample size.

Our estimation of the whole body COM of subjects could have potential errors due to inappropriate marker placements and a lack of anthropometric information specific for each subject. The markers were placed on bony landmarks of the body to quantify body segment parameters. Anthropometric data obtained in previous cadaveric studies were used to estimate the COM for each body segment, and then the whole body COM was calculated as the weighted sum of the body segments. Skin motion artifact and inaccurate marker placement resulting from greater adipose tissue in obese subjects would have affected our estimation. Human variations due to gender, age, and body composition, would have also affected our estimation of anthropometric measures. Although those limitations are common to human movement analyses using any marker based system, estimation of the COM with anthropometric reference data adopted from the literature has been validated and commonly accepted in the researcher community.

Most of the limitations lie on the way we modeled the whole body movement with the use of a single-link-plus-foot inverted pendulum. We used a single segment connecting the COM from the ankle to represent whole human body, and assumed the length of the segment doesn't change over time. For our first study, the use of a constant length inverted pendulum model to describe the multi-segmental STS movement could be a limitation. However, we confirmed that this multi-segmental motion mainly affected the vertical COM movement (the maximum differences in COM position between the

experiment and that estimated from the IPM, using the average pendulum length between seat-off and standing: 1.3 cm (SD 0.4) and 14.9 cm (SD 2.9) in the horizontal and vertical directions, respectively). Using different pendulum lengths (length at seat-off or standing) did not change our findings on the ROS<sub>v</sub> and ROS<sub>a</sub>, either. It has also been shown that the empirically measured COM positions and velocities during STS were adequately illustrated by the FSR [43], which is similar to our ROS<sub>v</sub>, and the model predictions from the FSR were consistent with experimental observations for STS followed by a volitional fall and balance recovery from induced slips during STS [43, 45, 60, 61, 74]. Therefore, the constant length IPM seems to adequately describe the horizontal COM movement during STS, although models taking into account changes in pendulum length, such as telescopic inverted pendulum [58, 59], would be more applicable.

The perturbation acceleration threshold was derived from several simple assumptions in our second study. A single segment was assumed above the ankle joint and the pendulum motion was restricted to the sagittal plane [62]. The inverted pendulum model has known limitations, but it provides a good approximation of upright standing behavior in the sagittal plane, even during large body sway [93-95]. Although we asked subjects not to bend their trunk to minimize the use of hip strategy to recover balance, it is possible that they used this kind of strategy to some extent. The model also did not have a toe segment, which could have affected its predictive capability for STEP trials especially for Young subjects. Multi-link segment model taking into account the hip joint and the toe segment would be more applicable.

The boundaries of regions of stability for walking were also derived from several simple assumptions, which could limit the predictive capacity of the boundaries in our

third study. We used a single segment connecting the COM from the ankle to represent human body, excluding the effect of the upper body motion. However, despite using a simplified pendulum motion, the empirically measured COM positions and velocities during walking were adequately illustrated by the feasible stability region (FSR), which is similar to our ROS<sub>v</sub>, and the model predictions from the FSR were consistent with experimental observations in response to slips induced during walking [43, 45, 74, 111]. The model used to define the ROS<sub>a</sub> also simply assumed the COM only undergoes forward/lateral acceleration prior to toe-off. This assumption would provide conservative estimation of the boundaries, where the subjects could have generated backward/medial acceleration to reduce the momentum prior to toe-off.

#### Future Research

Dynamic balance control has been traditionally assessed in terms of the interaction between COM position and its velocity, where insufficient momentum could be a trigger for imbalance. This dissertation study demonstrated that COM acceleration would provide further information on their momentum control, which could differentiate individuals with functional limitations from healthy individuals. Regions of stability established in this dissertation could serve as a quantitative assessment tool for abnormal control of posture and movement in the individuals who are prone to losses of balance and falls. In continuation of the studies included in this dissertation, suggestions for future research can be made as follows.

We established regions of stability during STS movement based on the COM position at seat-off and its instantaneous velocity (ROS<sub>v</sub>) or its peak acceleration (ROS<sub>a</sub>),

and demonstrated that the ROSa could better differentiate elderly individuals with difficulty in STS movement. Difficulty in performing STS movement is commonly observed and associated with falls in older adults [11-15], [18-21]. In fact, 7 out of 10 elderly subjects with difficulty in STS in this study had two or more falls in the year either prior to or after the testing, based on follow-up check-ups. Although identifying older adults with difficulty in STS using regions of stability could enhance our identification of elderly fallers, future study should include elderly fallers to assess its capability to differentiate them from healthy individuals.

We have also shown that FBOS and ankle dorsiflexor strength could be sensitive measures to detect elderly individuals with declined balance control. Our biomechanical model with FBOS demonstrated better predictive capability than the statistical model for identifying failed balance recovery trials. However, the study included healthy individuals only. Age-related declines in the ankle dorsiflexor strength have been demonstrated previously, where significant differences were detected between healthy older adults and fallers [78]. Future research should examine postural responses to stance perturbations for elderly fallers to see if those biomechanical measures and models could differentiate them from healthy individuals.

Our third study demonstrated age-related differences in momentum control during walking. Elderly fallers adapted a more conservative gait strategy, demonstrating significantly smaller forward COM velocity and acceleration with their COM significantly closer to the base of support at toe-off. Significantly smaller peak COM acceleration could be indicative of poor momentum control ability, which was pronounced in elderly fallers. A fall is induced by a sudden change in their momentum,

such as slipping or stumbling over an obstacle, where we need to regulate momentum by acceleration to maintain balance. Our results suggested that fallers might have less ability to regulate momentum by acceleration, which would result in imbalance in response to such external perturbations. Future study should include the testing where their walking is perturbed, inducing a fall, to see the relationship between their unperturbed stability and likelihood of falling.

Since slips were found to be the most common triggering event for falls [109], dynamic balance control during a slip has been assessed in terms of the interaction between position and velocity of the COM in relation to the BOS, providing feasible combinations of COM velocity and position for which loss of balance can be avoided [43, 46, 60, 112]. A study has shown that pre- and post-slip onset stability was highly correlated with the incidence of balance loss, and proactive adjustments in posture and gait pattern resulted in improved pre-slip stability [74]. Also, adaptive improvement in post-slip onset stability after repeated slip exposure predicted subsequent reduction of balance loss [74, 113]. Unperturbed gait stability tended to be lower in the young individuals who failed to recover after slip exposure, indicating that initial gait conditions would affect post-slip stability [111]. These studies imply that postural control deficits could be improved by motor training, resulting from improvement of one's dynamic stability prior to or after slip onset. However, these findings were limited to healthy young adults, and whether such improvement in dynamic stability could be possible for elderly individuals, especially for those who are prone to losses of balance, such as elderly adults with a history of falls, is still unknown. Our third study demonstrated that the elderly fallers used a more conservative strategy in their dynamic balance control

during normal walking, indicating reduced balance control ability. Control of dynamic stability in elderly fallers in response to a slip during walking would be different from healthy individuals. Difference in balance control ability may not be detectable when focusing only on COM velocity as we demonstrated in this dissertation study. Examining COM acceleration would provide further insights into dynamic balance control, allowing for insightful evaluation of balance dysfunction. Understanding proactive and reactive control of dynamic stability in elderly fallers would allow us to develop effective risk assessment tools and training strategies for reducing incidence of falls.

Finally, regions of stability established in this dissertation work could serve as a new quantitative assessment tool for abnormal control of posture and movement in the individuals with functional limitations who are prone to losses of balance and falls, including patients with neuromotor disorders. Classifying patients into certain categories based on the quantitative assessment of dynamic stability would facilitate effective individual-specific rehabilitation treatments. Development of a new assessment tool for dynamic balance control ability and treatment strategies to improve balance function in such individuals should be included in future research.

## APPENDIX A

### DERIVATION OF THE BOUNDARIES OF THE ROS<sub>v</sub> AND ROS<sub>a</sub> DURING SIT-TO-STAND MOVEMENT

#### Derivation of Boundaries of the ROS<sub>v</sub>

Boundaries of the ROS<sub>v</sub> were defined based on the work by Hof et al. (2005). When the COM has an initial velocity of  $\dot{X}$ , the condition for stability stated as  $X + \dot{X} / \omega_0$  (XcoM) should be within the BOS, where  $\omega_0 = \sqrt{g/l}$ . Since the BOS is confined within the foot length ( $X_h \leq X + \dot{X} / \omega_0 \leq X_t$ , where  $X_h$  and  $X_t$  are the heel and toe positions), the following formula can be obtained:

$$-\tilde{X}_s \leq \tilde{\dot{X}}_s \leq 1 - \tilde{X}_s \quad (\text{A.1})$$

where  $\tilde{X}_s$  and  $\tilde{\dot{X}}_s$  are normalized COM position and velocity at seat-off, defined as

$$\tilde{X}_s = (X_s - X_h) / L_f, \quad \tilde{\dot{X}}_s = \dot{X}_s / (L_f \omega_0) \quad (L_f = X_t - X_h : \text{foot length}).$$

#### Derivation of Boundaries of the ROS<sub>a</sub>

Using the inverted pendulum model shown in Fig.1, if we assume  $\theta$  is small and  $M \approx mgP$  ( $P$ : COP position), the equation of motion ( $ml^2\ddot{\theta} = mgl \sin \theta + M$ ) could be simplified as  $P = X - A\ddot{X}$ . Since the range of  $P$  is limited within the BOS ( $X_h \leq P \leq X_t$ ), the following relationship can be obtained:

$$\frac{L_f}{A} (\tilde{X}_s - 1) \leq \ddot{X}_s \leq \frac{L_f}{A} \tilde{X}_s \quad (\text{A.2})$$

where  $A = l/g$ , and  $\ddot{X}_s$  is the possible range of COM acceleration after seat-off.



A successful STS movement termination is to maintain the COM position directly over the BOS as its velocity vanishes. The COM velocity, generated by its acceleration prior to and at the seat-off, is required to be reduced to zero before the COM reaches the BOS boundary. This condition allows for an estimation of the required peak COM acceleration prior to seat-off to achieve a successful movement termination.

With the modeled COM acceleration profile with seat-off scenario (a) (Fig.2.1a), we can derive the relationship among the COM position, velocity and acceleration as followings. From the initial position to seat-off:

$$(1/4)\ddot{X}_p t_1^2 = X_s - X_i \quad (\text{A.3})$$

where  $t_1$ : time from initial position to seat-off,  $X_s$ : COM position at seat-off,  $X_i$ : initial COM position, and  $\ddot{X}_p$ : peak COM acceleration prior to seat-off.

Similarly, from seat-off to the instant when the COM reaches the BOS boundary,

$$\int_0^{t_2} (V_s + \frac{1}{2}\ddot{X}_s t) dt = X_{BOS} - X_s \quad (\text{A.4})$$

where  $V_s$ : COM velocity at seat-off,  $t_2$ : time from seat-off to the instant when the COM reaches the BOS boundary,  $X_{BOS}$ : position of the BOS boundary,  $\ddot{X}_s$ : peak COM acceleration after seat-off.

Since the COM velocity needs to be reduced to zero when the COM reaches the BOS boundary ( $V_s + \frac{1}{2}\ddot{X}_s t_2 = 0$ , where  $V_s = \frac{1}{2}\ddot{X}_p t_1$ ), Eq. (A.4) would be written as:

$$\ddot{X}_p = -\frac{X_{BOS} - X_s}{X_s - X_i} \ddot{X}_s \quad (\text{A.5})$$

The following equation can be obtained after substituting  $\ddot{X}_p$  in Eq. (A.2) with Eq.(A.5):

$$-\frac{X_h - X_s}{X_s - X_i} \tilde{X}_s \leq \tilde{X}_p \leq \frac{X_t - X_s}{X_s - X_i} (1 - \tilde{X}_s) \quad (\text{A.6})$$

where  $\tilde{\ddot{X}}_p$  is the normalized peak COM acceleration defined as  $\tilde{\ddot{X}}_p = A\ddot{X}_p / L_f$ ,  $X_h$  and  $X_t$  indicate the heel and toe positions, which are considered as the backward and forward boundaries of the BOS, respectively. Those boundaries can be also obtained from the boundaries of the ROSv, using Eq. (A.1). Eq.(A.6) can also be derived using Eq. (A.1),  $(1/4)\ddot{X}_p t_1^2 = X_s - X_i$ , and  $V_s = \frac{1}{2} \ddot{X}_p t_1$ .

Similarly, when using the COM acceleration profile with seat-off scenario (b) (Fig. 2.1b), boundaries of the ROSa were defined as below. From the initial position to seat-off, the following relationship can be obtained:

$$\begin{aligned} \frac{1}{(2+a)^2} \ddot{X}_p t_1^2 + \int_0^{\frac{a}{2+a} t_1} \left( \frac{1}{2+a} \ddot{X}_p t - \frac{2+a}{2t_1} \ddot{X}_p t_1^2 \right) dt &= X_s - X_i \\ \therefore \frac{6+a(6-a^2)}{6(2+a)^2} \ddot{X}_p t_1^2 &= X_s - X_i \end{aligned} \quad (A.7)$$

where  $t_1$ : time from the initial position to seat-off,  $X_s$ : COM position at seat-off,  $X_i$ : initial COM position,  $\ddot{X}_p$ : peak COM acceleration prior to seat-off,  $a$ : the ratio of the peak negative to the positive peak acceleration prior to seat-off.

From seat-off to the instant when the COM reaches the boundary of the BOS,

$$\int_0^{t_2} \left\{ V_s + \left( \ddot{X}_s t - \frac{\ddot{X}_s}{2t_2} t^2 \right) \right\} dt = X_{BOS} - X_s \quad (A.8)$$

where  $V_s$ : COM velocity at seat-off,  $t_2$ : time from seat-off to the instant when the COM reaches the boundary of the BOS,  $X_{BOS}$ : boundary of the BOS,  $\ddot{X}_s$ : peak COM

acceleration after seat-off. When substituting  $V_s = \frac{1}{2} \left( \frac{2-a^2}{2+a} \right) \ddot{X}_p t_1$  to the above equation, the

following relationship can be obtained:

$$\frac{1}{2} \left( \frac{2-a^2}{2+a} \right) \ddot{X}_p t_1 t_2 + \frac{1}{3} \ddot{X}_s t_2^2 = X_{BOS} - X_s \quad (\text{A.9})$$

Considering the generated COM velocity needs to be reduced to zero when the COM reaches the BOS boundary ( $V_s + \frac{1}{2} \ddot{X}_s t_2 = 0$ ) and with substituting Eq. (A.7) to it Eq. (A.9),

we can derive

$$\ddot{X}_p = - \frac{6+a(6-a^2)}{(2-a^2)^2} \frac{X_{BOS} - X_s}{X_s - X_i} \ddot{X}_s \quad (\text{A.10})$$

The following formula can be obtained by substituting  $\ddot{X}_s$  in Eq. (A.2) with Eq. (A.10):

$$-G \frac{X_h - X_s}{X_s - X_i} \tilde{X}_s \leq \tilde{X}_p \leq G \frac{X_i - X_s}{X_s - X_i} (1 - \tilde{X}_s) \quad (\text{A.11})$$

where  $G = \{6 + a(6 - a^2)\} / (2 - a^2)^2$ .

Eq. (A.6) and (A.11), therefore, provide us the upper and lower boundaries of peak COM acceleration to be generated prior to seat-off that would allow a successful termination of the STS movement.

## APPENDIX B

### DERIVATION OF THE THRESHOLD ACCELERATION OF BACKWARD SUPPORT SURFACE TRANSLATION

This section describes the derivation of the threshold acceleration of backward support surface translation. A single-link-plus-foot inverted pendulum model in the sagittal plane, which has a mass point at the COM and a center of rotation at the ankle joint, as shown in Fig.4.2, was used, where  $x$  indicates the COM position in the antero-posterior direction, and  $x_h$  and  $x_t$  indicate the heel position and toe position, respectively.

The derivation of the threshold acceleration involved two steps. First, the range of resultant COM acceleration in a given COM position was estimated. Then, the threshold acceleration of backward support surface translation in which a person can maintain their balance was determined.

#### Range of Resultant COM Acceleration in a Given COM Position

Using the inverted pendulum model shown in Fig.4.2, if we assume  $\theta$  is small and  $M \approx mgP$  ( $P$ : COP position), the equation of motion ( $ml^2\ddot{\theta} = mgl \sin \theta + M$ ) could be simplified as  $P = X - A\ddot{X}$ . Since the range of  $P$  is limited within the BOS ( $x_h \leq P \leq x_t$ ), the following relationship can be obtained:

$$\frac{L_f}{A}(\tilde{X} - 1) \leq \ddot{X} \leq \frac{L_f}{A} \tilde{X} \quad (\text{B.1})$$

where  $A = l/g$ , and  $\ddot{X}$  is the possible range of COM acceleration in a given COM position.

### Threshold Acceleration of the Backward Support Surface Translation

COM velocity induced by backward support surface translation needs to be reduced to zero before the COM reaches the boundary of the BOS to maintain one's balance without stepping. According to the pilot data, we made the following assumptions: (1) COM passively moves in the opposite direction of the direction of plate movement during initial backward acceleration phase of the perturbation, (2) this COM acceleration profile can be assumed as a triangle-shape with the peak in the middle, (3) COM velocity generated by this acceleration needs to be reduced to zero at the movement termination by the COM acceleration resulting from ankle joint moment. According to the assumptions, COM velocity generated by perturbation  $V_p$  would be  $V_p = -1/2 \ddot{X}_p \Delta t$  where  $\ddot{X}_p$  indicates peak acceleration of the perturbation plate,  $\Delta t$  indicates perturbation duration. This generated COM velocity needs to be reduced to zero before the COM reaches the forward boundary of the BOS. That is,

$$\int_0^t (V_p + \ddot{X}_m t) dt = BOSf - \left\{ Xi - \frac{1}{4} \ddot{X}_p (\Delta t)^2 \right\} \quad (B.2)$$

where

$BOSf$  : Forward boundary of the BOS

$t$  : Time until the COM reaches the  $BOSf$  from the end of the backward perturbation acceleration

$V_p$  : COM velocity at the end of the backward perturbation acceleration, passively generated by perturbation

$\ddot{X}_m$  : Possible maximum COM acceleration at the end of the backward perturbation acceleration

$X_i$  : Initial COM position before the backward perturbation

$\frac{1}{4} \ddot{X}_p (\Delta t)^2$  : Passive COM excursion caused by backward perturbation acceleration

$BOSf - \left\{ X_i - \frac{1}{4} \ddot{X}_p (\Delta t)^2 \right\}$  : Remaining distance to the forward boundary of the BOS at the end of the backward perturbation acceleration.

Since  $V_p = -1/2 \ddot{X}_p \Delta t$ , eq. (B.2) would be:

$$-\frac{1}{2} \ddot{X}_p \Delta t(t) + \frac{1}{2} \ddot{X}_p t^2 = BOSf - X_i + \frac{1}{4} \ddot{X}_p (\Delta t)^2 \quad (B.3)$$

In this regard, initiated COM velocity  $V_p$  needs to be reduced to zero at the movement termination:  $V_p + \ddot{X}_p t = 0$ ,  $\therefore t = (\ddot{X}_p \Delta t) / 2 \ddot{X}_p$ . Eq. (B.3) becomes

$$T \ddot{X}_p^2 + 2T \ddot{X}_p \ddot{X}_p + 8B \ddot{X}_p = 0 \quad (B.4)$$

where  $T = (\Delta t)^2$ ,  $B = BOSf - X_i$ . From eq.(B.1), a person can generate  $\ddot{X} = L_f (\tilde{X} - 1) / A$  in a given normalized COM position of  $\tilde{X}$  to counter-balance the backward perturbation, preventing the COM from going forward. Since the normalized COM position ( $\tilde{X}$ ) at the end of the perturbation is  $\tilde{X} = \frac{1}{L_f} \left\{ X_i - \frac{1}{4} \ddot{X}_p (\Delta t)^2 - X_h \right\}$ ,  $\ddot{X}_m$  would be

$\ddot{X}_m = -\frac{T}{4A} \ddot{X}_p + \frac{1}{A} (X_i - BOSf)$ . By assigning this relationship to (B.4), the following

equation would be finally obtained:

$$\ddot{X}_p = 4(X_i - BOSf) \left\{ \frac{a}{a \Delta t^2 + \sqrt{2} \Delta t} \right\} \quad (B.5)$$

where  $a = \sqrt{l/A} = \sqrt{g/l}$ . This indicates threshold acceleration of backward support surface translation, in which a person can maintain balance without taking a step, in a given initial COM position ( $X_i$ ), perturbation duration  $\Delta t$ , and  $BOSf$ .

## APPENDIX C

### DERIVATION OF THE BOUNDARIES OF THE ROS<sub>v</sub> AND ROS<sub>a</sub> DURING WALKING

#### Derivation of the Boundaries of the ROS<sub>v</sub> in the AP and ML Directions

##### *AP Direction*

The boundaries of the ROS<sub>v</sub> were derived based on the work presented by Hof et al. (2005). When the COM has an initial velocity of  $\dot{X}$ , the condition for stability stated as  $X + \dot{X} / \omega_0$  (XcoM) should be within the BOS, where  $\omega_0 = \sqrt{g/l}$ . Since the BOS is confined within the feet area ( $X_h \leq X + \dot{X} / \omega_0 \leq X_t$ ), the following formula can be obtained:

$$-\tilde{X}_{TO} \leq \tilde{\dot{X}}_{TO} \leq 1 - \tilde{X}_{TO} \quad (C.1)$$

where  $\tilde{X}_{TO}$  and  $\tilde{\dot{X}}_{TO}$  are normalized COM position and velocity at toe-off instant (TO) in AP direction, defined as  $\tilde{X}_{TO} = (X_{TO} - X_h) / L_f$ ,  $\tilde{\dot{X}}_{TO} = \dot{X}_{TO} / (L_f \omega_0)$  ( $\omega_0 = \sqrt{g/l}$ ,  $L_f = X_t - X_h$ : foot length).

##### *ML Direction*

Similarly, the ROS<sub>v</sub> in the ML direction can be obtained:

$$-\tilde{Y}_{TO} \leq \tilde{\dot{Y}}_{TO} \leq 1 - \tilde{Y}_{TO} \quad (C.2)$$

where  $\tilde{Y}_{TO}$  and  $\tilde{\dot{Y}}_{TO}$  are normalized COM position and velocity at TO in the ML direction, defined as  $\tilde{Y}_{TO} = (Y_{TO} - Y_{ma}) / Lw$ ,  $\tilde{\dot{Y}}_{TO} = \dot{Y}_{TO} / (Lw \omega_0)$  ( $\omega_0 = \sqrt{g/l}$ ,  $Lw = Y_{la} - Y_{ma}$ : ankle width,  $Y_{ma}$  and  $Y_{la}$  indicate the medial and lateral ankles).

## C.2. Derivation of the Boundaries of the ROSa in the AP and ML Directions

COM must remain at a position over the BOS as the velocity vanishes if they intend to stop walking. The COM velocity at TO, generated by its acceleration prior to TO, needs to be reduced to zero before the COM reaches the boundary of the BOS. This condition allows us to estimate the peak COM acceleration required to achieve a successful movement termination.

### *AP Direction*

The COM acceleration profile was modeled as a triangle-shape with the peak in the middle prior to TO. We used COM position and velocity when the COM velocity became its minimum prior to TO as initial COM position and velocity.

From initial position to toe-off,

$$\dot{X}_i t + 1/4 \ddot{X}_p t^2 = X_{TO} - X_i \quad (C.3)$$

where  $t$ : time from initial position to toe-off,  $X_{TO}$ : COM position at toe-off,  $X_i$ : initial COM position,  $\ddot{X}_p$ : peak COM acceleration prior to toe-off. COM velocity at the instant of toe-off would be:  $\dot{X}_{TO} = \dot{X}_i + 1/2 \ddot{X}_p t$ .

From eq. (C.1),  $-\omega_0 L_f \tilde{X}_{TO} \leq \dot{X}_i + 1/2 \ddot{X}_p t \leq \omega_0 L_f (1 - \tilde{X}_{TO})$ . By assigning this relationship to eq. (C.3), the following formula would be obtained:

$$\frac{(\tilde{X}_{TO} + \dot{X}_i / (\omega_0 L_f))(\tilde{X}_{TO} - \dot{X}_i / (\omega_0 L_f))}{\tilde{X}_{TO} - \tilde{X}_i} < \tilde{\ddot{X}}_p < \frac{(1 - \tilde{X}_{TO} + \dot{X}_i / (\omega_0 L_f))(1 - \tilde{X}_{TO} - \dot{X}_i / (\omega_0 L_f))}{\tilde{X}_{TO} - \tilde{X}_i} \quad (C.4)$$

where  $\tilde{X}_{TO}$  and  $\tilde{X}_i$  are the normalized COM position at TO and initial COM position defined as  $\tilde{X}_{TO} = X_{TO} / L_f$  and  $\tilde{X}_i = X_i / L_f$ , respectively.  $\tilde{\ddot{X}}_p$  is the normalized peak COM acceleration prior to TO defined as  $\tilde{\ddot{X}}_p = \ddot{X}_p / \omega_0^2 L_f$ . This relationship provides us the peak



COM acceleration needed to be generated prior to TO that would allow a successful termination of the movement with a given COM position at TO.

### *ML Direction*

The COM acceleration profile was modeled as a triangle-shape with the peak in the middle prior to TO. We used COM position when the COM velocity became zero prior to TO as initial COM position ( $Y_i$ ) in the ML direction. The ROSa derived for STS movement in Appendix A can be applied. From eq. (A.5),

$$-\frac{Y_{ma} - Y_{TO}}{Y_{TO} - Y_i} \tilde{Y}_{TO} \leq \tilde{Y}_p \leq \frac{Y_{la} - Y_{TO}}{Y_{TO} - Y_i} (1 - \tilde{Y}_{TO}) \quad (C.5)$$

where  $\tilde{Y}_p$  is the normalized peak COM acceleration defined as  $\tilde{Y}_p = \ddot{Y}_p / \omega_0^2 L_w$ ,  $Y_{ma}$  and  $Y_{la}$  indicate the medial and lateral ankles, which are considered as the backward and forward boundaries of the BOS, respectively. This relationship provides us the peak COM acceleration needed to be generated prior to toe-off that would allow a successful termination of the movement with a given COM position at toe-off.

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