The Influence of Age on Compensatory Stepping Thresholds

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THESIS

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This thesis is dedicated to my wife. Thank you, Brooke.

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LIST OF ABBREVIATIONS

COM	Center of mass
$d_{threshold}$	Stepping threshold displacement
d	The anteroposterior distance between the COM of the body and the toe marker (anterior disturbances) or heel marker (posterior disturbances)
d_{min}	Minimum d
d_{step}	d at step completion
L_{step}	Compensatory step length
L_{low}	Lowering step length
MOS	Margin of stability
MOS_{DO}	Margin of stability at disturbance onset
MOS_{min}	Minimum margin of stability
MOS_{step}	Margin of stability at step completion
TFA	Trunk flexion angle at step completion
TFV	Trunk flexion velocity at step completion
t_{low}	Lowering step time
t_{step}	Time after disturbance onset of compensatory step completion
t_{step} - t_{belt}	Time of compensatory step completion relative to when the treadmill belts stopped moving
v_{belt}	Peak absolute value of the velocity of the treadmill belts
θ_{hip_ext}	Peak hip extension angle
θ_{hip_flex}	Peak hip flexion angle
θ_{knee_flex}	Peak knee flexion angle

LIST OF ABBREVIATIONS (continued)

$\omega_{shoulder\_ext}$	Average of the left and right initial peak shoulder extension angular velocities
$\omega_{shoulder\_flex}$	Average of the left and right initial peak shoulder flexion angular velocities
$\omega_{SLshoulder}$	Peak shoulder extension (anterior disturbances) or flexion (posterior disturbances) angular velocity of the non-stepping-limb side.
$\omega_{NSLshoulder}$	Peak shoulder extension (anterior disturbances) or flexion (posterior disturbances) angular velocity of the non-stepping-limb side.

SUMMARY

The purpose of the following studies was to investigate the effects of increasing age on anterior and posterior compensatory stepping thresholds. Young (18-35 years of age), middle-aged (55-64 years), and older (65 years and older) adults responded to anterior and posterior disturbances from surface translations. Stepping thresholds were defined as the disturbance displacements beyond which a single or second compensatory step was elicited. Subjects were given disturbances under three conditions. In one condition, disturbances were delivered as subjects were standing and subjects were instructed to "try not to step." In a second condition, disturbances were delivered as subjects were standing and subjects were instructed to "try to take only one step." In a third condition, disturbances were delivered as subjects walked and subjects were instructed to "try to take only one step." It was hypothesized that stepping thresholds would be reduced with increasing age. As an exception, it was hypothesized that posterior, single-stepping thresholds would not be reduced with increasing age. These hypotheses were partially supported. Measures that quantified the relationship between the body center of mass and the edge of the base of support were evaluated as indicators of stability. Kinematics of the step, lower extremity joints, trunk, and shoulder were measured to reflect modifiable aspects of the compensatory response.

When given disturbances from an initial standing position and instructed to "try not to step," young adults demonstrated larger anterior single-stepping thresholds than that of middle-aged and older adults. No influence of increasing age was observed for posterior single-stepping thresholds. These results supported our hypotheses. For middle-aged and older adults, the inability to avoid anterior compensatory steps was associated with becoming dynamically unstable following smaller displacements than young adults.

When given disturbances from an initial standing position and instructed to “try to take only one step,” young adults demonstrated larger anterior multiple-stepping thresholds than middle-aged and older adults. Young adults demonstrated larger posterior, multiple-stepping thresholds than older adults, but not middle-aged adults. These results partially supported our hypotheses. Age-related reductions in posterior multiple-stepping thresholds were associated with shorter compensatory steps.

When disturbances were given during walking and subjects were instructed to “try to take only one step,” declines in stepping thresholds were not strongly associated with age or were not consistent across all disturbance velocities. These results partially supported our hypotheses. An important factor to successfully recovering from an anterior or posterior disturbance was compensatory step length. A longer step length allowed for a greater distance between the center of mass of the body and the edge of the base of support at step completion.

The methods and results of these studies may be used to identify targets for fall-prevention interventions. These targets include enhancing the muscular response of the lower extremities so that a disturbance imparts less dynamic instability and trunk rotation, as well as increasing step length when a step is needed. Exercise

interventions that incorporate aspects of specificity have potential for affecting these targets.

1. INTRODUCTION

1.1 **Background**

With more than 2,200,000 injuries and nearly 20,000 deaths reported annually in the United States, falls are the leading cause of unintentional injury and death for adults age 65 and older (data from 2008; CDC, 2011). Yearly direct medical costs of fatal and non-fatal, fall-related injuries have totaled $0.2 billion and $19 billion, respectively (Stevens, 2006). From 2001 to 2009, the rate of fall injuries for adults age 65 and over has increased an average of 2.4% per year (CDC, 2011). From 1999 to 2008, the rate of fall-related mortality for adults age 65 and over has increased an average of 6.4% per year (CDC, 2011). These escalating rates of injury and mortality are magnified by the growing population of older adults, which is predicted to double by the year 2030 to a total of 70 million people (Knickman and Snell, 2002). Clearly, fall-related mortality and injuries are common and costly problems that are rapidly becoming more prevalent.

The yearly fall incidence of older adults does not appear to have been reduced in the last 30 years, as suggested by reported incidences of 28% (Prudham and Evans, 1981), 32% (Tinetti et al., 1988), 39% (Hausdorff et al., 2001), and 42% (Fox et al., 2010). The increasing prevalence of fall-related injuries and mortality calls for improved assessments of fall risk and interventions for fall prevention that can be administered on a large scale. The rates of fall incidence, fall-related injuries, and fall-related mortality increase with age (Campbell et al., 1990; Sattin, 1992). Therefore, studies that investigate the influence of age on factors related to fall risk have the potential to reduce the fall incidence of older adults by identifying specific targets for assessments and interventions.

Slips and trips account for 30-77% of falls in community-living older adults (Berg et al., 1997; Gabell et al., 1985; Hill et al., 1999; Lord et al., 1993). Both fall causes are due to an external disturbance (e.g. a slippery floor, a curb) that necessitates a compensatory stepping response, that is, steps that occur in response to a disturbance and that stabilize upright stance (McIlroy and Maki, 1993), thereby preventing a fall.

Commonly reported methods for delivering controlled disturbances in the laboratory include waist pulls and surface translations. Waist pulls are delivered by applying a force at the subject's waist through a cord. Surface translations are delivered by the motion of a platform or treadmill belt upon which the subject stands. In response to such external disturbances, older adults step more frequently and step in response to smaller disturbance magnitudes than young adults (Hall et al., 1999; Jensen et al., 2001; Mille et al., 2003; Pai et al., 1998; Schulz et al., 2005). Older adult fallers, that is, those who had experienced a fall within one year prior to participation, have demonstrated a decreased ability to avoid stepping in response to a forward waist pull than their age-matched counterparts (Pai et al., 1998). These results suggest that the inability to avoid stepping may reflect fall risk outside of the laboratory.

If a disturbance has a sufficiently large magnitude, all individuals, regardless of age, require a compensatory step to avoid a fall. In these circumstances, age-related differences in compensatory stepping persist. Compared to young adults, older adults take multiple compensatory steps more frequently and in response to smaller disturbances (Luchies et al., 1994; McIlroy and Maki, 1996; Schulz et al., 2005).

A relevant aspect of performing a successful, one-step response to a disturbance is maintaining stability. Here, stability is a continuous, quantitative value that, in the absence of the support limbs giving way, is indicative of if an individual is falling. Positive stability implies that an individual is not falling. Negative stability suggests that an individual is falling, and therefore must enact a compensatory response to regain stability. The magnitude of stability reflects how “close” a person is to changing stability states.

In *quasi-static* situations, the distance (*d*) between the vertical projection of the body center of mass (COM) and the boundary of the base of support reflects direction-specific stability (Shumway-Cook and Woollacott, 1995; Winter., 1995). The margin of stability (*MOS*) is a measure of *dynamic* stability that considers both the horizontal position and velocity of the COM relative to the base of support in its calculation (Hof et al., 2005). The *MOS* is directly related to the impulse that would be required to change the state of stability. Here, “dynamic stability” is a term that specifically refers to *MOS*. “Stability” is a more general term that applies to both *MOS* and *d*. By including stability measures in an investigation of compensatory stepping, the influence of age on the ability to maintain and restore dynamic stability can be assessed.

Important aspects to the successful maintenance of stability in response to a disturbance include the control of body-segment rotation and, when a step is taken, the control of foot placement. When attempting to avoid a step in response to an external disturbance, segment rotation can be minimized by producing ankle moments that move the body center of mass away from the edge of the base of support (i.e. “ankle strategy”; Horak and Nashner, 1986). On the other hand, segment rotations about the COM of the body can be utilized to avoid a step (Hof et al., 2005; Hof, 2007). Examples of segment rotations include hip flexion or extension (“hip strategy”, Horak and Nashner, 1986), knee flexion or extension (Henry et al., 1998), and shoulder flexion or extension. For compensatory stepping responses, completing a sufficiently long step in the available amount of time, minimizing trunk rotation, and effectively rotating the upper extremities are important contributors to successfully recovering from a postural disturbance in one step (Luchies et al., 1994; Schulz et al., 2005), recovering from postural disturbances requiring multiple steps (Owings et al., 2001), recovering from slips (Troy et al, 2008; Marigold et al., 2003; Tang and Woollacott, 1998; Troy et al., 2009), and recovering from trips (Pavol et al., 2001). By considering kinematic variables pertaining to step placement and body segment rotation, the modifiable aspects of the compensatory stepping response that influence age-related differences in stability maintenance can be identified.

It is evident that aging has a deleterious effect on the ability to limit the number of compensatory steps. A previous study quantified this effect by determining the age-specific disturbance displacements that resulted in forward compensatory steps (Mille et al., 1993). This *anterior single-stepping threshold* was larger for young adults than for older adults. To the best of our knowledge, age-related differences in the *posterior single-stepping threshold* have not been investigated. Furthermore, the effects of age on *anterior and posterior multiple-stepping thresholds* have not been studied. Here, multiple-stepping thresholds are defined as the disturbance displacement beyond which two or more compensatory steps are elicited.

Previous observations of compensatory stepping have been made as subjects responded to disturbances from an initial standing position. Trips and slips, however, occur during locomotion. Therefore, the initial sensory and motor conditions are considerably different than those of upright standing. In addition, differences exist in the neuromuscular and biomechanical consequences of the disturbance. Consequently, the sensory and motor demands of the response to disturbances during upright standing and during locomotion are different. It is unknown if, as a result of the different sensory and motor demands, age-related differences in multiple-stepping thresholds would be observed when disturbances are applied during gait. Quantifying the effects of age on single- and multiple-stepping thresholds would provide a novel understanding of age-related declines in compensatory stepping—a response that is necessary for the successful recovery of common fall causes. Identifying the aspects of the compensatory stepping response associated with age-related declines in stepping thresholds, such as stability (*d* or *MOS*), step kinematics, and body segment rotations, would help identify specific targets for intervention.

The older adults of previously referenced studies had a minimum age of 60 to 65 years (Hall et al., 1999; Jensen et al., 2001; McIlroy and Maki, 1996; Mille et al., 2003; Pai et al., 1998; Schulz et al., 2005). However, the detrimental effects of increasing age on the compensatory stepping response may be manifested at an earlier age. Unpublished data from an ongoing, one-year prospective study (study first reported in Rosenblatt et al., 2010) suggest that the rate of falling by adults age 65 and older (1.32

± 0.40 falls per person) is not significantly different than that of adults age 55 to 64 (1.13 ± 0.40 falls per person, $p = 0.52$). This result suggests that it may be important to identify age-related declines in compensatory stepping that are present in "middle-age."

1.2 **Purpose**

The purpose of the following three studies was to investigate the effects of age on anterior and posterior compensatory stepping thresholds. The studies investigated anterior and posterior single-stepping thresholds from an initial standing position (Chapter 2), anterior and posterior multiple-stepping thresholds from an initial standing position (Chapter 3), and anterior and posterior multiple-stepping thresholds from disturbances applied while walking on a treadmill (Chapter 4). Young (35 and younger), middle-aged (55-54), and older (65 and older) subjects were included in these studies. It was hypothesized that stepping thresholds would be reduced with increasing age. As an exception, it was hypothesized that posterior, single-stepping thresholds would not be reduced with increasing age. The *significance* of these studies is that they may identify modifiable aspects of the compensatory stepping response to moderate disturbances that degrade with age and that may be relevant to the successful recovery from common fall causes. In doing so, the results could inform future assessment and interventions for fall prevention. By improving assessments and interventions, the high fall incidence associated with old age may be reduced, in turn decreasing the number and cost of fall-related injuries and mortality. These studies are *innovative* because they use a novel combination of recently developed techniques and measures to advance our understanding of how age influences factors relevant to fall risk. By reporting single- and multiple-stepping thresholds, these studies will provide a unique, quantified assessment of the effects of increasing age on the compensatory stepping response. By using stability measures in conjunction with other kinematic measures, these studies will provide new insight as to how age affects the ability and strategies to maintain stability with a compensatory response

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PREFACE TO CHAPTERS TWO, THREE, AND FOUR

Chapters two, three, and four are studies on compensatory stepping thresholds. What distinguishes each chapter is the type of threshold that is of focus (single-stepping or multiple-stepping threshold) and the initial condition (standing or walking) during which the disturbance was delivered. A total of 13 young subjects, 12 middle-aged subjects, and 11 older subjects participated in at least one of the studies. Subjects who participated in all three studies did so over the course of two laboratory visits. Each visit was separated by a minimum of 1 day and a maximum of 17 days. An effort was made to evenly distribute the order in which subjects participated in the three studies among groups and sexes. The distribution of participation order was made uneven by subjects who dropped out of the study. One male, middle-aged subject only participated in the study of Chapter 2, ending participation before a second visit due to medical concerns not associated with the study. One male, middle-aged subject only participated in the study of Chapter 3, ending further participation due to scheduling conflicts. One female, middle-aged subject completed the studies of Chapters 2 and 4, but ended participation partway through the study of Chapter 3 due to muscular discomfort. One male, older subject completed the studies of Chapters 2 and 3, but ended participation after the first trial of Chapter 4 due to muscle discomfort.

2. THE INFLUENCE OF INCREASING AGE ON THE THRESHOLDS OF SINGLE COMPENSATORY STEPS.

2.1 <u>Introduction</u>

With increasing age, unintentional falls make up a larger proportion of nonfatal injuries (15.5% for ages 18-35, 33.8% for ages 55-64, 62.4% for ages 65+; data from 2001-2009; CDC, 2011). The escalating prevalence of fall-related injuries may be reduced by identifying age-related factors that are related to fall risk and amenable to intervention. The ability to recover from an external disturbance without taking a compensatory step is reduced with age. Here, a compensatory step is defined as a step that serves to stabilize upright stance (McIlroy and Maki, 1993). Upon delivery of anterior disturbances, that is, disturbances that may require forward compensatory steps, older adults step more frequently and step in response to smaller disturbance magnitudes than young adults (Hall et al., 1999; Jensen et al., 2001; Mille et al., 2003; Pai et al., 1998; Schulz et al., 2005). Such age-related differences are less apparent in the response to posterior disturbances (Hall et al., 1999; Schulz et al., 2005). The ability to avoid forward stepping is further degraded for older adults with a recent fall history, suggesting that the ability to avoid stepping may be indicative of fall risk (Pai et al., 1998).

Preventing a compensatory step requires successful management of the center of mass (COM) of the body relative to the edge of the base of support. The margin of stability (*MOS*) is a measure that quantifies this relationship by considering both the horizontal position and velocity of the COM relative to the base of support in its calculation (Curtze et al., 2010; Hof et al., 2005; Hof et al., 2007; Hof et al., 2010). If the *MOS* is positive, then an individual is considered dynamically stable (i.e. the vertical projection of the COM will return to or be maintained within the base of support). In a standing, dynamically stable condition, a compensatory step can be avoided by producing ankle moments that move the COM away from the edge of the base of support (i.e. "ankle strategy"; Horak and Nashner, 1986). If the *MOS* is negative, then an individual is considered dynamically unstable (i.e. the vertical projection of the COM will not return to or will move outside of the base of support). A compensatory step may be avoided in this condition, and dynamic stability can be restored, by counter-rotating segments about the COM (Hof et al., 2005; Hof, 2007). Previously reported counter-rotation movements include flexion or extension of the hip ("hip strategy", Horak and Nashner, 1986), knee (Henry et al., 1998), or upper extremities (Marigold et al., 2003; Tang and Woollacott, 1998; Troy et al., 2009).

Age-related reductions in anterior single compensatory stepping thresholds have been observed (Mille et al., 2003). Single-stepping thresholds are defined as the disturbance displacement (e.g. the distance the waist is pulled forward by a cord) beyond which a step is elicited. Our current understanding of the influence of age on single compensatory stepping thresholds would benefit from the novel inclusion of the *MOS*. By measuring the *MOS*, the effects of a disturbance on dynamic stability can be measured, the minimum *MOS* that can be maintained without a compensatory step can be noted, and the strategies for maintaining/retaining dynamic stability can be observed for different age groups. Older adults may become dynamically unstable in response to disturbances of smaller magnitudes, or older adults may step unnecessarily despite being dynamically stable (Pai et al., 1998). For fall circumstances in everyday life, the direction of the disturbance is often not anticipated. Therefore, assessments of compensatory stepping thresholds would be more context-specific to falls outside the laboratory by having a degree of directional uncertainty, delivering anterior *and* posterior disturbances intermixed within the same protocol.

The purpose of this study was to investigate the effects of age on anterior and posterior single compensatory stepping thresholds. To do this, young adults (35 and younger), middle-aged adults (55-64), and older adults (65 years and older) were included in this study. It was hypothesized that *anterior* single-stepping thresholds *would be reduced* with increasing age. It was also hypothesized that *posterior* single-stepping thresholds *would not be reduced* with increasing age. *MOS* was measured as an indicator of dynamic stability associated with the response. Peak hip flexion/extension angles, peak knee flexion angles, and peak flexion/extension shoulder angular velocities were evaluated as indicators of counter-rotation strategies.

2.2 **Methods**

This protocol was approved by the University of Illinois at Chicago Institutional Review Board and subjects provided written, informed consent prior to participation. Thirteen young adults, 11 middle-aged adults, and 11 older adults participated in this study (Table I). Subjects were not allowed to participate if they reported neurological, musculoskeletal, or other injuries or disorders that would limit their safe participation. In addition, older and middle-aged adults were screened using DEXA (QDR® 4500 Elite, Hologic®, Inc, Bedford, MA) for a minimum bone mineral density of 0.61 g/cm^2 at the left femoral neck.

TABLE I

MEAN ± STANDARD DEVIATION FOR SUBJECT CHARACTERISTICS

Group	n (male:female)	Age (years)	Height (cm)	Mass (kg)	Avg. Foot Length (cm)
Young	13 (6:7)	31.1 ± 0.8	172.5 ± 8.9	74.4 ± 13.5	26.0 ± 1.7
Middle-Aged	11 (6:5)	57.6 ± 2.5	170.3 ± 10.5	76.4 ± 11.4	26.2 ± 1.8
Older	11 (6:5)	73.8 ± 5.3	169.3 ± 10.3	73.7 ± 13.9	26.3 ± 1.9

[a] No significant main effects of age group (one-way ANOVA) were observed for height ($p = 0.721$), mass ($p = 0.875$), or average foot length ($p = 0.892$).

Subjects wore their own comfortable walking shoes and were outfitted with a safety harness that was instrumented with a load cell (Omega Engineering, Inc., Stamford, CT) to measure the force supported by the harness. Twenty-two retroreflective markers were placed on the arms, legs, and torso of each subject (Kadaba et al., 1990). Subjects then stood on a microprocessor-controlled, stepper motor-driven, dual-belt treadmill (Figure 1) with their arms to the side, feet positioned shoulder-width apart, and toes evenly aligned (Figure 2). Once subjects were in position, the investigator initiated the disturbance through a computer interface. Subjects were not given a verbal or visual cue that the disturbance was initiated. After a one- to three-second delay, the treadmill belts were accelerated in the anterior or posterior directions. Subjects were instructed to “try not to step.”

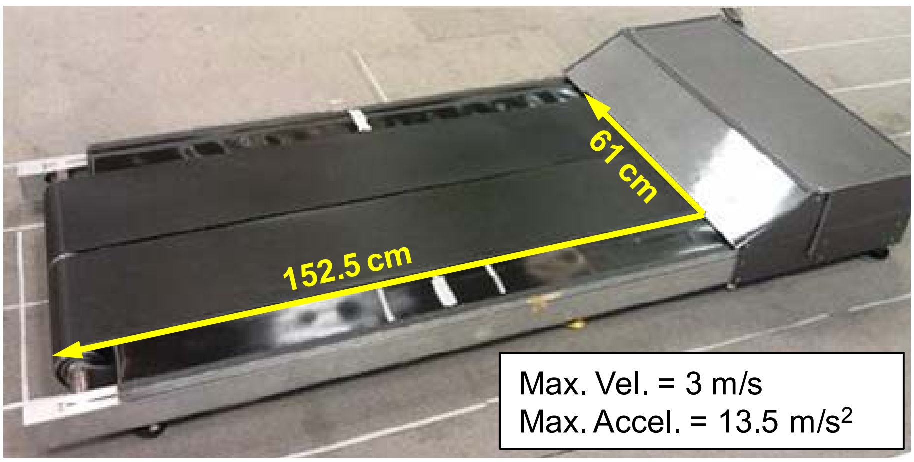
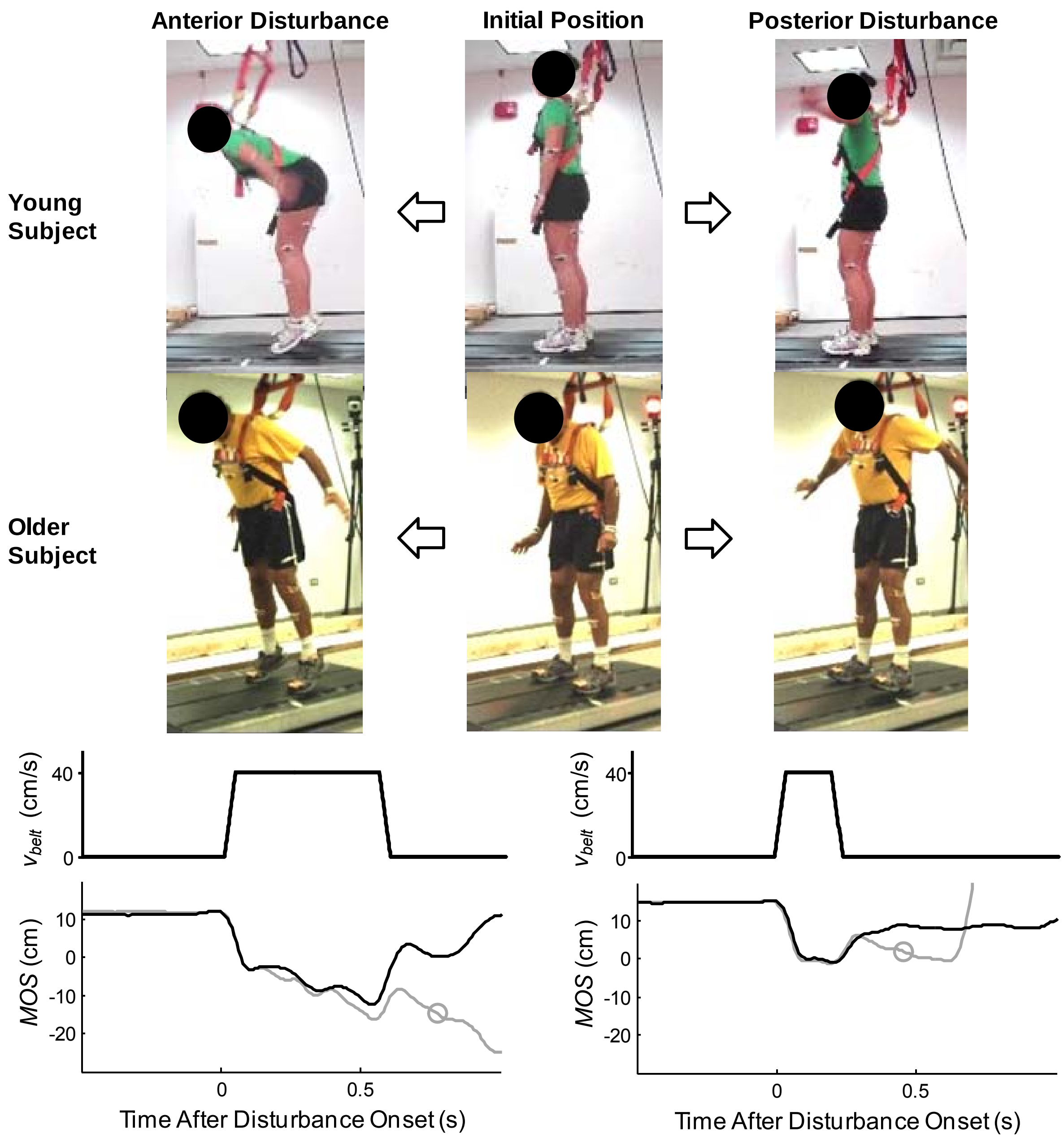


Figure 1. The microprocessor-controlled, stepper motor-driven, dual-belt treadmill (ActiveStep™, Simbex, Lebanon, NH).

The disturbances delivered by the treadmill were defined by “velocity profiles” that were trapezoidal in shape (Figure 2). Each velocity profile had three phases, including an acceleration phase, a constant velocity phase, and a deceleration phase. The available, absolute peak disturbance velocities (v_{belt}) were incremented on an exponential scale (v_{belt} = 1.8 * $1.25^{10,11\ldots16}$ cm; adapted from Mille et al., 2003) and rounded to the nearest even integer. For anterior disturbances, that is, backward belt motion that may necessitate forward steps, v_{belt} ranged from 20 cm/s to 64 cm/s (Figure 3). For posterior disturbances, v_{belt} ranged from 16 cm/s to 52 cm/s. For both



Anterior Disturbance
Initial Position
Posterior Disturbance
Young Subject
Older Subject
v_{belt} (cm/s)
40
0
MOS (cm)
10
0
-10
-20
0
0.5
Time After Disturbance Onset (s)

Figure 2. Time series for absolute treadmill belt velocity (v_{belt}) and margin of stability (MOS) for an older subject (bottom row of pictures and gray *MOS* curves) and young subject (top row of pictures and black *MOS* curves).

[a] For both subjects, the anterior disturbance (left side of figure) had a peak v_{belt} of 40 cm/s and an absolute displacement of 22 cm.

[b] For both subjects, the posterior disturbance (right side of figure) had a peak v_{belt} of 40 cm/s and an absolute displacement of 8 cm.

[c] The older subject stepped with the left limb in response to both anterior and posterior disturbances. The young subject did not step.

[d] *MOS* (Equation 2.1) is shown relative to the right toe (anterior disturbances) or heel (posterior disturbances) markers.

[e] Pictures of the disturbance responses are taken immediately before toe off by the older subject (marked by a gray circle on the *MOS* figures) and at an analogous time for the young subject.

[f] The *MOS* for the older subject in response to the posterior disturbance (lower right figure, gray curve) rapidly becomes positive at the end because of a second step with the right limb.

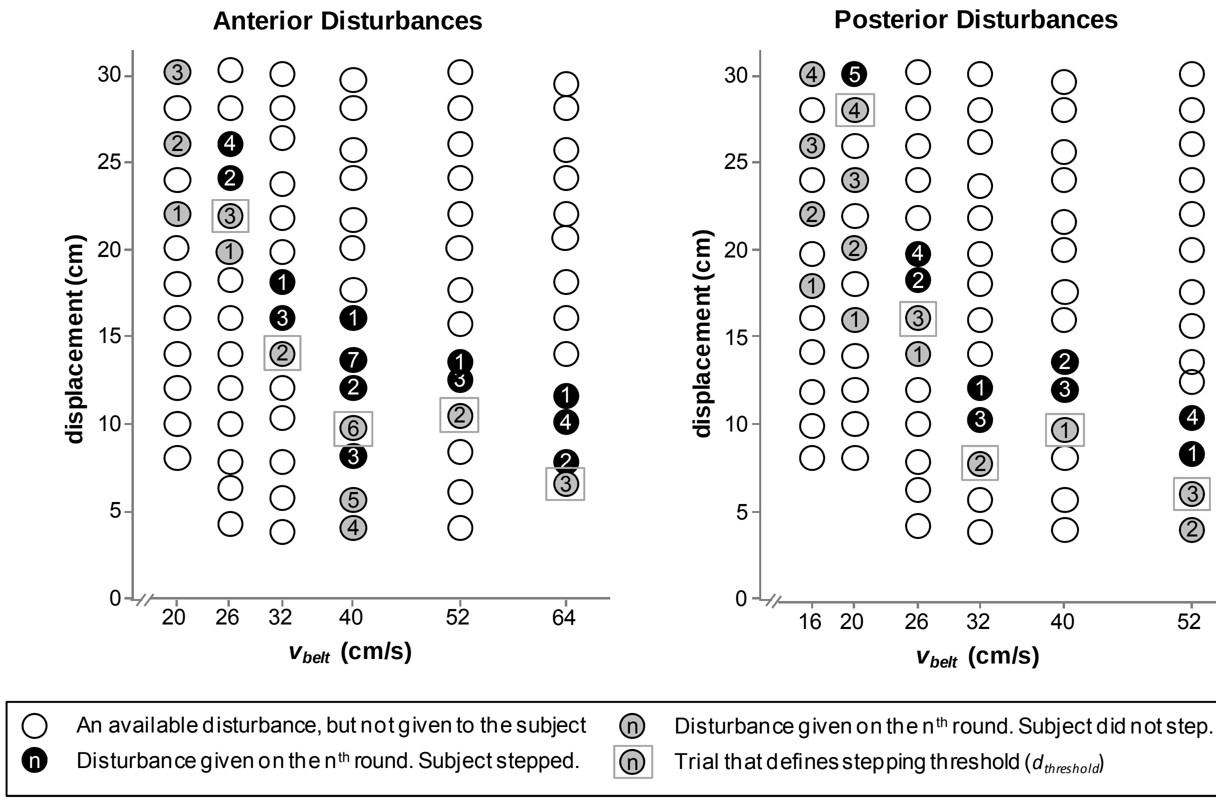


Figure 3. An example of an older subject's progression of disturbances.

[a] The available disturbances are represented by circles positioned at their respective absolute displacement and absolute peak velocity (v_{belt}).

[b] White circles represent disturbances that were not given to the subject. Black circles represent disturbances that resulted in a step. Gray circles represent disturbances that did not result in a step.

[c] A gray box around the circle represents a disturbance that was used to define the stepping threshold displacement ($d_{threshold}$) at a given v_{belt}.

[d] The number within the circle represents the round of disturbances in which the disturbance was delivered.

the forward and backward directions, absolute disturbance displacements ranged from approximately 4 cm to approximately 30 cm. Disturbance displacements were separated by approximately 2 cm increments. The acceleration and deceleration phases were 40 to 60 ms in duration, and ranged in acceleration from ±400 cm/s^2 to ±1300 cm/s^2.

The following progression of disturbances was designed to identify thresholds at each v_{belt} while limiting the total number of disturbances. Initially, subjects were given 12 practice trials. The first six practice trials were anterior disturbances that sequentially increased in v_{belt} and decreased in absolute displacement. The next six practice trials were posterior disturbances that sequentially increased in v_{belt} and decreased in absolute displacement. The following block of 12 trials represented the beginning of the test protocol, and consisted of the same 12 disturbances as the practice round presented in a random order (Figure 3, marked with a "1"). The remainder of the test protocol consisted of blocks of up to 12 disturbances. These disturbances, each having a unique v_{belt}, were presented in a random order. Based on the subject response (no step, step, or use of harness, as observed by the primary investigator), the disturbance displacement at a given v_{belt} was either increased (no step) or decreased (step or use of harness) by two increments (4 cm) for the subsequent block of disturbances. If this displacement was not available or had already been given to the subject, then a one-increment (2 cm) change in the displacement was made for the subsequent block. This progression continued until stepping thresholds were established. Stepping thresholds were defined as the largest displacement ($d_{threshold}$) at a given v_{belt} for which a subject could avoid stepping or engaging the harness (Figure 3, disturbances within grey boxes). Here, $d_{threshold}$ is always expressed as an absolute value. Stepping thresholds were confirmed by the observation that subjects stepped or engaged the harness at displacements of the same v_{belt} that were 2 cm and 4 cm (if available) greater than $d_{threshold}$. Subjects received, on average, 67.9 ± 7.0 total disturbances, and subjects were allowed to rest as requested.

Marker positions were recorded by eight cameras at 120 Hz (Cortex, Motion Analysis, Santa Rosa, CA). The three-dimensional coordinates of each marker were exported to text files from which custom programs (MATLAB, The Mathworks, Inc., Natick, MA) were used to calculate dependent variables. Stepping and non-stepping responses were verified from the motion of the heel and toe markers. All reported steps must have extended the edge of the base of support, which was defined as the most anterior toe (anterior disturbances) or posterior heel (posterior disturbances) marker, by a minimum of 1 cm. The instrumented harness was considered to be engaged if greater than 50% body weight was supported.

MOS was calculated as follows (adapted from Hof et al., 2005):

$$MOS = d + \left(\frac{v}{\sqrt{\frac{g}{l}}} \right) \qquad (2.1)$$

Subjects were positioned facing the $+x$ direction. For anterior disturbances, $d = x_{toe} - x_{COM}$ and $v = v_{toe} - v_{COM}$. x_{COM} and v_{COM} are the anteroposterior COM position and velocity, respectively. The COM position was estimated from the marker positions on the arms, legs, pelvis, and trunk (Dempster, 1955 via Winter, 2005). x_{toe} and v_{toe} are the anteroposterior position and velocity of the toe marker, respectively. For posterior disturbances, $d = x_{COM} - x_{heel}$ and $v = v_{COM} - v_{heel}$. x_{heel} and v_{heel} are the anteroposterior position and velocity of the heel marker, respectively. The gravity term is g = 9.81 m/s^2, and l is the sagittal plane distance between the ankle center and the COM. MOS_{min} was defined as the minimum value of MOS measured after the disturbance onset. The greater value relative to the left or right foot was selected to represent MOS_{min}. In order to determine if proactive adjustments to stability were made before the disturbance, MOS was also calculated at the time of disturbance onset (MOS_{DO}).

The primary dependent variable was $d_{threshold}$, expressed as a percentage of foot length, at each v_{belt}. If the subject recovered from the largest displacement (≈ 30 cm) without taking a step, or if the subject was unable to recover from the smallest displacement (≈ 4 or 6 cm) without avoiding a step, then no $d_{threshold}$ was recorded for that subject at that v_{belt}.

MOS_{min} and joint kinematics were measured for the non-stepping trials that defined $d_{threshold}$. The smaller of the right and left peak hip flexion angles (θ_{hip_flex}) or extension angles (θ_{hip_ext}) was selected for anterior and posterior disturbances, respectively. The smaller of the right and left peak knee flexion angles (θ_{knee_flex}) was selected for posterior disturbances. The average of the left and right initial peak shoulder extension velocities ($\omega_{shoulder\_ext}$) or flexion velocities ($\omega_{shoulder\_flex}$) was selected for anterior and posterior disturbances, respectively.

For $d_{threshold}$, MOS, and joint kinematic variables, stepwise multiple linear regressions were conducted with v_{belt} (as a percentage of foot length), age, and an age*v_{belt} interaction as independent variables. Separate regressions were conducted for anterior and posterior disturbances. The stepping method criterion for entry of a variable into the model was an F-test significance of $p < 0.05$. The stepping method criterion for removal of a variable from the model was an F-test significance of $p < 0.10$. If the final regression model contained an $age*v_{belt}$ interaction, separate one-way analyses of variance (ANOVAs) were conducted at each of the six treadmill belt velocities to determine differences between young, middle-aged, and older adults. If significant main effects were found, Tukey's post hoc tests were used to evaluate specific between-group differences. All statistics were evaluated using SPSS (SPSS Inc., Chicago, IL) with a significance level of $\alpha = 0.05$.

2.3 **Results**

2.3.1 **Anterior Disturbances**

When anterior single-stepping thresholds were frequently established across all age groups, the thresholds of young subjects were longer than the thresholds of middle-aged and older subjects. At low anterior treadmill belt velocities ($v_{belt} \leq 26$ cm/s), thresholds were not always established, especially for young adults (Figure 4). Thresholds were not established at low velocities because subjects recovered from the longest displacements (≈ 30 cm) without stepping. The final linear regression model ($p < 0.001$, $r^2 = 0.400$) for $d_{threshold}$ (% foot length) was the following:

$$d_{threshold} = 92.021 - 0.106*v_{belt} - 0.003*(age*v_{belt}) \qquad (2.2)$$

The $age*v_{belt}$ interaction accounted for 36.1% of the variance in $d_{threshold}$, and v_{belt} accounted for an additional 3.9% of the variance in $d_{threshold}$. Based on separate one-

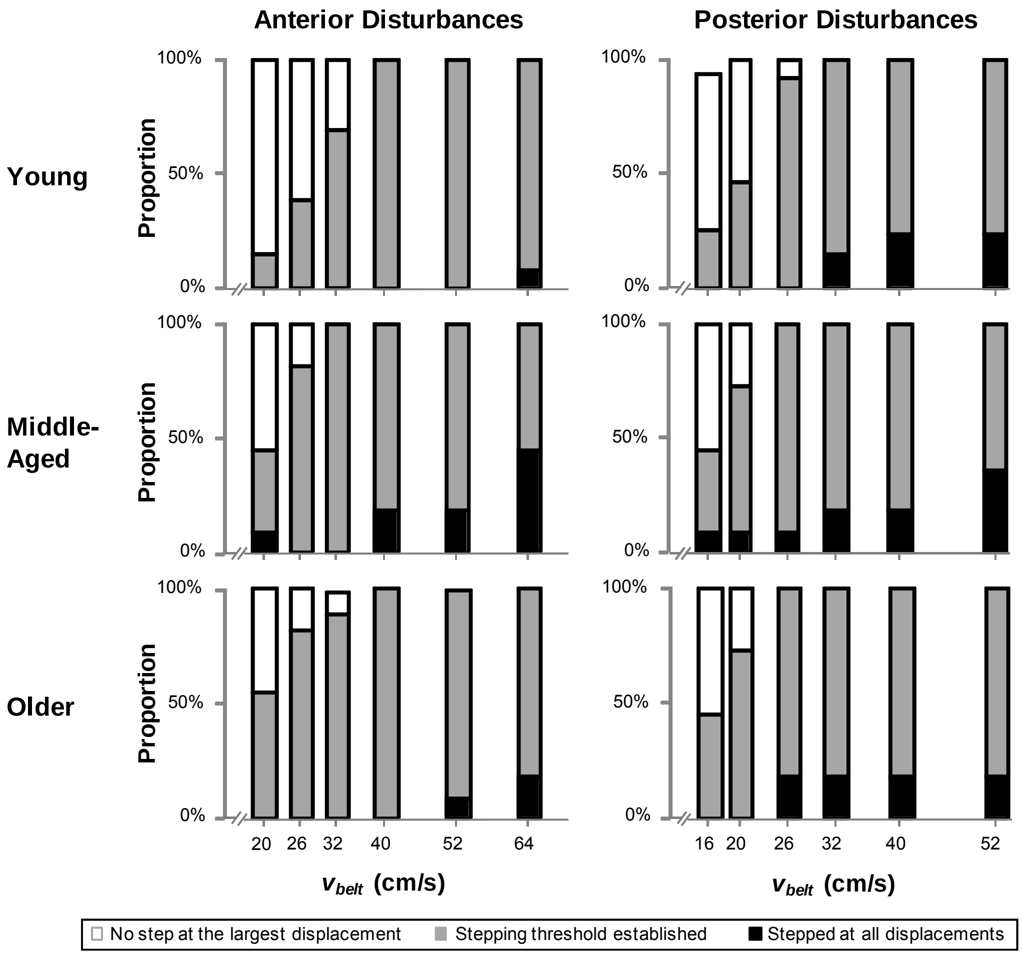


Figure 4. The proportion of subjects who established stepping thresholds, organized by group (Young, Middle-Aged, or Older), direction of disturbance (Anterior or Posterior), and peak disturbance velocity (v_{belt}).

[a] At each v_{belt}, subjects either did not step at the largest displacement (white), established a stepping threshold (gray), or stepped in response to at all displacements (black).

[b] Proportions do not sum to 100% if investigator error resulted in no definitive threshold being established for a subject at a given v_{belt}.

way ANOVAs, significant main effects of age group were observed at treadmill belt velocities of 32 cm/s ($p = 0.038$), 40 cm/s ($p < 0.001$), and 52 cm/s ($p = 0.035$). At a treadmill belt velocity of 32 cm/s, the $d_{threshold}$ of young subjects was significantly longer than that of middle-aged subjects ($p = 0.045$, Figure 5). At a treadmill belt velocity of 40 cm/s, the $d_{threshold}$ of young subjects was significantly longer than that of middle-aged subjects ($p = 0.045$) and older subjects ($p = 0.001$). At a treadmill belt velocity of 52 cm/s, no significant between-group differences were observed.

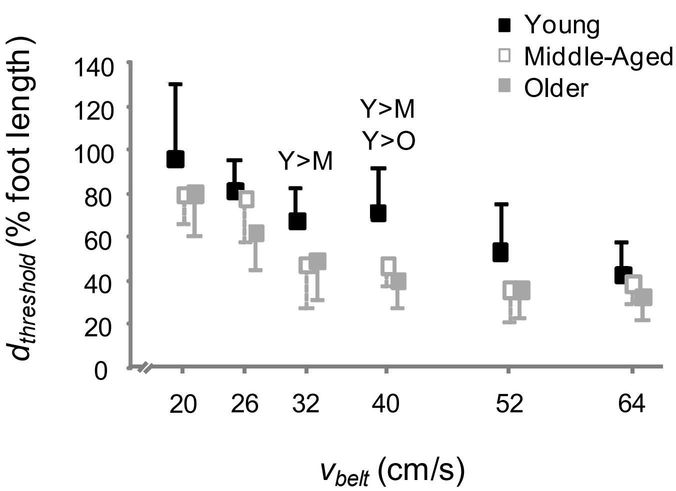


Figure 5. Anterior single-stepping thresholds ($d_{threshold}$) at each peak treadmill belt velocity (v_{belt}).

[a] Young (Y) subjects are denoted by black markers, middle-aged (M) subjects are denoted by white markers, and older (O) subjects are denoted by gray markers.

[b] Significant ($p < 0.05$) differences between groups are noted (e.g. young adults had significantly greater values than older adults, Y>O).

Margin of stability variables were not influenced by increasing age. No variables were entered into the stepwise linear regression for MOS_{DO} (total mean ± s.d. = 11.9 ± 1.8 cm). The final linear regression model ($p < 0.001$, $r^2 = 0.503$) for MOS_{min} (cm) was the following:

$$MOS_{min} = 0.992 - 0.069*v_{belt} - 2.282*10^{-4}*(age*v_{belt}) \qquad (2.3)$$

The v_{belt} accounted for 47.8% of the variance in $d_{threshold}$, and the $age*v_{belt}$ interaction accounted for an additional 2.5% of the variance in $d_{threshold}$. No significant main effects of age group on MOS_{min} (total mean ± s.d. = -8.1 ± 4.6 cm) were observed from separate one-way ANOVAs at each treadmill belt velocity ($p \geq 0.052$). An example of *MOS* throughout the response to an anterior disturbance is provided in Figure 2.

Hip and shoulder kinematic variables were not influenced by increasing age. No variables were entered into the stepwise linear regression for θ_{hip_flex} (total mean ± s.d. = 25 ± 17 deg). The final linear regression model ($p = 0.032$, $r^2 = 0.030$) for $\omega_{shoulder\_ext}$ (deg/s) was the following:

$$\omega_{shoulder\_ext} = 77.117 - 0.350*v_{belt} \qquad (2.4)$$

2.3.2 **Posterior Disturbances**

Posterior single-stepping thresholds were not influenced by age (Figure 6). The final linear regression model ($p < 0.001$; $r^2 = 0.329$) for $d_{threshold}$ (% foot length) was the following:

$$d_{threshold} = 81.366 - 0.319 * v_{belt} \tag{2.5}$$

The excluded coefficient for *age* (-0.096) was not significant ($p = 0.160$).

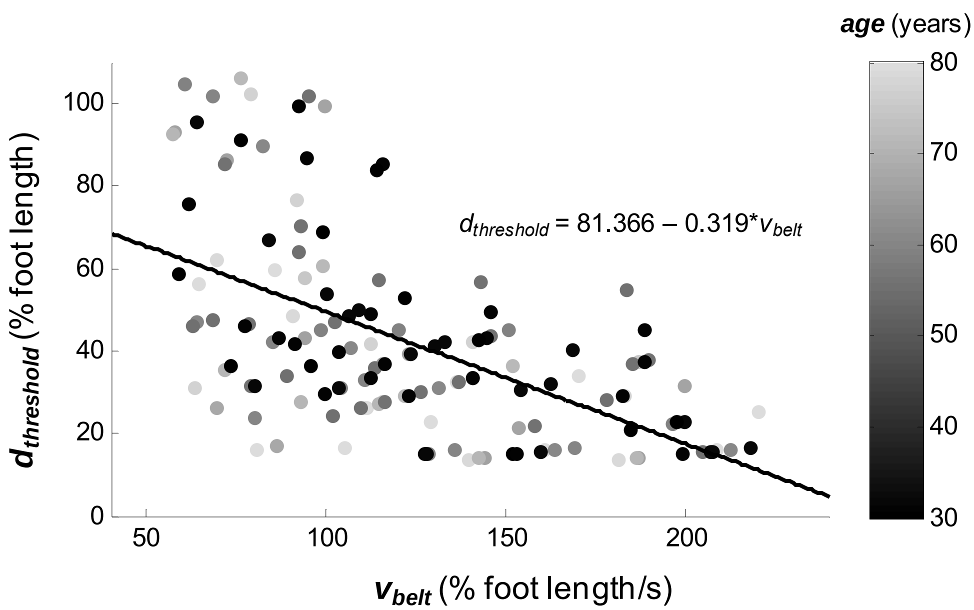


Figure 6. Posterior single-stepping thresholds ($d_{threshold}$) as a function of treadmill belt velocity (v_{belt}).

[a] Increasing age is represented by a gradient transition from black to grey. *age* and the *age**v_{belt} interaction were not chosen by the stepwise regression.

The minimum margin of stability associated with posterior single-stepping thresholds trended towards greater stability with increasing age. The final linear regression model ($p < 0.001$, $r^2 = 0.586$) for MOS_{min} (cm) was the following:

$$MOS_{min} = 6.762 - 0.062 * v_{belt} + 0.031 * age \qquad (2.6)$$

The v_{belt} accounted for 56.1% of the variance in MOS_{min}, and *age* accounted for the additional 2.5% of the variance in MOS_{min}. No variables were entered into the stepwise linear regression for MOS_{DO} (total mean ± s.d. = 16.0 ± 2.1 cm). An example of *MOS* throughout the response to a posterior disturbance is provided in Figure 2.

With increasing age, knee flexion was increased and shoulder flexion velocity was decreased. The final linear regression model ($p = 0.006$, $r^2 = 0.054$) for θ_{knee_flex} (deg; flexion is a positive angle) was the following:

$$\theta_{knee\_flex} = 9.579 + 0.098 * age \qquad (2.7)$$

The final linear regression model ($p = 0.007$, $r^2 = 0.050$) for $\omega_{shoulder\_flex}$ (deg/s) was the following:

$$\omega_{shoulder\_flex} = 245.201 - 1.576 * age \qquad (2.8)$$

No variables were entered into the stepwise linear regression for θ_{hip_ext} (total mean ± s.d. = 1 ± 7 deg).

2.4 **Discussion**

The purpose of this study was to investigate the effects of age on anterior and posterior single compensatory stepping thresholds. It was hypothesized that anterior single-stepping thresholds would be reduced with increasing age. It was also hypothesized that posterior single-stepping thresholds would not be reduced with increasing age. These hypotheses were supported. Observed differences were most likely due to the subject's response to the disturbance, not other influential factors such as body anthropometrics (Rogers et al., 1996) or dynamic stability before disturbance onset.

2.4.1 **Anterior Disturbances**

Young adults were better at avoiding an anterior step than middle-aged or older adults. Age-related reductions in $d_{threshold}$ were observed at treadmill belt velocities of 32 cm/s and 40 cm/s. In response to disturbances of lower velocities, the majority of young subjects did not establish stepping thresholds (Figure 4). Instead, these young subjects recovered from the longest displacements without stepping. Previous research has identified low-velocity stepping thresholds in response to forward waist pulls. These low-velocity thresholds transitioned to shorter, high-velocity thresholds at smaller velocities for older adults (17.8 ± 4.3 cm/s) than for young adults (23.5 ± 5.1 cm/s; Mille et al., 2003). In response to surface translations, however, a low-velocity stepping threshold may not exist for the populations represented in the present study. Instead, there may be an age-related minimum v_{belt} below which a disturbance will not elicit a step regardless of displacement.

The $d_{threshold}$ identified in this study can help explain previous findings. In response to a disturbance velocity of 40 cm/s, older adults have been reported to step more frequently than young adults when the disturbance displacement (≈ 58% foot length for our subjects; Jensen et al., 2001) was above the average $d_{threshold}$ of older adults (40.0 ± 12.1% foot length), but below the average $d_{threshold}$ of young adults (71.5 ± 20.2% foot length). When the disturbance displacements of previous studies were above the $d_{threshold}$ of both young and older adults (displacement ≈ 58% foot length, velocity = 60 cm/s; Jensen et al., 2001) or below the $d_{threshold}$ of both young and older adults (displacement ≈ 38.5% foot length, velocity = 40 cm/s; Hall et al., 1999), no age-related differences were reported. For larger v_{belt} values, the observed anterior $d_{threshold}$ were smaller than high-velocity thresholds in response to waist pulls (72 ± 8% foot length for young adults; 59 ± 8% foot length for older adults; Mille et al., 2003). These differences may be due to waist pulls as a disturbance instead of surface translations, the predictability of the disturbance direction, and/or the methods of determining thresholds.

Even though older and middle-aged subjects responded to smaller disturbance displacements than young subjects, the resulting dynamic instability was not different between age groups. Most likely, an increase in the disturbance displacement is accompanied by an increase in the external impulse applied to the subject. If this is the case, a smaller external impulse had the same effect on the dynamic stability of middle-aged and older subjects as a larger impulse had on young subjects. This result suggests that increasing age diminishes the ability to resist dynamic instability due to an external impulse. This suggestion concurs with previous observations that older adults, in response to anterior disturbances, are deficient in absorbing kinetic energy with the lower extremity muscles and minimizing linear momentum of the COM and proximal segments (Hall and Jensen, 2002; Jensen et al., 2001). The negative MOS_{min} associated with stepping thresholds were not significantly different between groups, suggesting that age does not influence the ability or willingness to recover from dynamic instability without taking a step. As v_{belt} increased, subjects recovered from greater instability without stepping (Equation 2.3). This trend is most likely due to the subjects' abilities to use the third phase of the velocity profile, the deceleration of which became greater with increasing v_{belt}, to help retain dynamic stability (Bothner and Jensen, 2001). Potential age-related differences in the ability to use this deceleration phase may warrant further study. As observed in Figure 2, the *MOS* of the older subject started to become less negative when the treadmill belt decelerated. Despite this trend towards positive stability, the older adult stepped. Therefore, a measure of *MOS* other than a minimum value may better reveal age-related differences in dynamic stability maintenance that lead to declines in stepping thresholds.

Counter-rotation strategies at the hip and shoulders were not influenced by age. In response to surface translations, young adults have demonstrated a hip strategy marked by hip flexion of about 15 - 30 deg (Runge et al., 1999). The subjects of this study demonstrated similar, if not greater θ_{hip_flex}, suggesting that a hip strategy was utilized.

2.4.2 **Posterior Disturbances**

Age did not significantly influence posterior $d_{threshold}$ (Equation 2.5, Figure 5). This result supports previous reports that old and young adults did not differ in the frequency of posterior stepping (Schulz et al., 2005; Hall et al., 1999). In this study, a smaller proportion of young subjects established stepping thresholds at v_{belt} = 20 cm/s than middle-aged or older subjects (Figure 4). This v_{belt} may be close to a minimum velocity, below which young adults will not step regardless of disturbance displacement. The minimum v_{belt} that evokes stepping in middle-aged and older adults could presumably be lower than that of young adults.

Although the MOS_{min} became larger with increasing age, any age-related differences in the ability to resist or recover from dynamic instability did not significantly affect stepping thresholds. As with anterior disturbances, the inverse relationship of MOS_{min} and v_{belt} was most likely a result of the ability to use surface decelerations to restore stability (Bothner and Jensen, 2001).

Counter-rotation strategies in response to posterior disturbances changed with age. With increasing age, less shoulder flexion velocity was incorporated into the response (Equation 2.8) and greater knee flexion was utilized (Equation 2.7). These observations may warrant further study on a potential age-related transition in counter-rotation strategies. Such a transition, however, does not appear to significantly influence posterior stepping thresholds.

2.4.3 **Limitations**

Although the effects of increasing age on stepping thresholds were determined, further study is warranted to better understand what influences stepping thresholds. For anterior disturbances, 60% of the variance in $d_{threshold}$ was not explained by age and

v_{belt}. For posterior disturbances, 67.1% of the variance in $d_{threshold}$ was not explained by v_{belt}. Perhaps greater variance in $d_{threshold}$ could have been explained by including tests of physical function (e.g. strength, reaction time). Furthermore, the within-subject repeatability of stepping thresholds should be investigated before the clinical applicability of this measure can be assessed.

2.4.4 **Conclusions**

To the best of our knowledge, this is the first study to evaluate the influence of age on anterior and posterior stepping thresholds in response to surface translations of unpredictable direction. This study uniquely demonstrated declines of anterior stepping thresholds by adults as young as 55. The older adults of previously referenced studies had a minimum age of 60 years (Mille et al., 2003). Fall-risk assessments and interventions may be better served by expanding the focus to include a middle-aged population. Future study is needed to determine if the age-related declines observed in this study are amenable to intervention and relevant to fall risk outside the laboratory.

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3. THE INFLUENCE OF INCREASING AGE ON THE THRESHOLDS OF MULTIPLE-COMPENSATORY STEPS: DISTURBANCES WHILE STANDING

3.1 Introduction

When the magnitude of a disturbance to static posture exceeds a threshold, a feet-in-place response is insufficient to restore stability. Consequently, a compensatory step is often necessary to avoid a fall (Chapter 1; Mille et al., 2003; Maki and McIlroy, 1997). Compensatory stepping is a common response by older adults to fall-causing disturbances. According to surveillance footage in a geriatric center, 45% of observed falls involved an attempted compensatory step (Holliday et al., 1990). Compared to the one-step response that is commonly exhibited by young adults in response to laboratory-induced disturbances, older adults perform multiple compensatory steps more frequently and in response to smaller disturbances (Luchies et al., 1994; McIlroy and Maki, 1996; Schulz et al., 2005). With perturbation-based training interventions, older adults have demonstrated a reduction in the frequency of multiple stepping responses (Mansfield et al., 2010). The prevalence of compensatory stepping as a recovery strategy, in conjunction with the age-related changes in compensatory stepping and the observation that these changes are amenable to intervention, suggests that the performance of compensatory stepping has potential to be a viable target for interventions to reduce fall incidence.

Single compensatory stepping thresholds are defined here as the disturbance displacements beyond which a single compensatory step results. These *single-*

stepping thresholds decline with age (Mille et al., 2003; Chapter 1). Older adults have demonstrated a higher frequency of multiple-step responses and a larger average total number of compensatory steps than young adults (Brauer et al., 2002; Luchies et al., 1994; McIlroy and Maki, 1996; Schulz et al., 2005). To the best of the author's knowledge, age-specific thresholds for *multiple* compensatory steps, or disturbance displacements beyond which second compensatory steps are used to maintain balance, have not been reported.

The purpose of this study was to investigate the influence of age on anterior and posterior thresholds of multiple compensatory steps. To do this, young adults (18-35 years of age), middle-aged adults (55-64 years of age), and older adults (65 years and older) were included in this study. It was hypothesized that anterior and posterior stepping thresholds would be reduced with increasing age. Anteroposterior margin of stability (*MOS*; Curtze et al., 2010; Hof et al., 2005; Hof et al., 2007; Hof et al., 2010) and the distance (*d*) between the body's center of mass (COM) and edge of the base of support were evaluated as indicators of stability during the compensatory stepping response. Sagittal plane step, trunk, and shoulder kinematics were evaluated to gain insight about potentially modifiable aspects of the stepping response that may influence stepping thresholds and stability.

3.2 **Methods**

This protocol was approved by the University of Illinois at Chicago Institutional Review Board and subjects provided written, informed consent before participation.

Thirteen young adults (male:female = 6:7; age = 31.1 ± 0.8 years; height = 172.5 ± 8.9 cm; mass = 74.4 ± 13.5 kg), 11 middle-aged adults (male:female = 6:5; age = 58.2 ± 2.6 years; height = 169.3 ± 9.3 cm; mass = 77.3 ± 11.1 kg), and 11 older adults (male:female = 6:5; age = 73.8 ± 5.3 years; height = 169.3 ± 10.6 cm; mass = 73.7 ± 13.9 kg) participated in this study. No significant main effects of age group (one-way ANOVA) existed for height ($p = 0.628$) or mass ($p = 0.786$). Subjects were not allowed to participate if they reported neurological or musculoskeletal injuries or disorders that would limit their safe participation. In addition, older and middle-aged adults were screened using DEXA (QDR® 4500 Elite, Hologic®, Inc, Bedford, MA) for a minimum bone mineral density of 0.61 g/cm^2 at the left femoral neck.

Subjects wore comfortable walking shoes and were outfitted with a safety harness. In order to measure the force supported by the harness, the harness was instrumented with a load cell (Omega Engineering, Inc., Stamford, CT). Twenty-two retroreflective markers were placed on the arms, legs, pelvis, and trunk of each subject (Kadaba et al., 1990). Subjects then stood on a microprocessor-controlled, stepper motor-driven, dual-belt treadmill (ActiveStep™, Simbex, Lebanon, NH; Figure 1) with their arms to the side, feet positioned shoulder-width apart, and toes evenly aligned (visible in Figures 7 and 8). Once subjects were in position, the investigator initiated the disturbance through a computer interface. After a one to three second delay, the treadmill belts were accelerated in the anterior or posterior direction. Subjects were instructed to “try to take only one step” in response to belt movements. However, subjects were also informed that a second step was acceptable if it did not extend

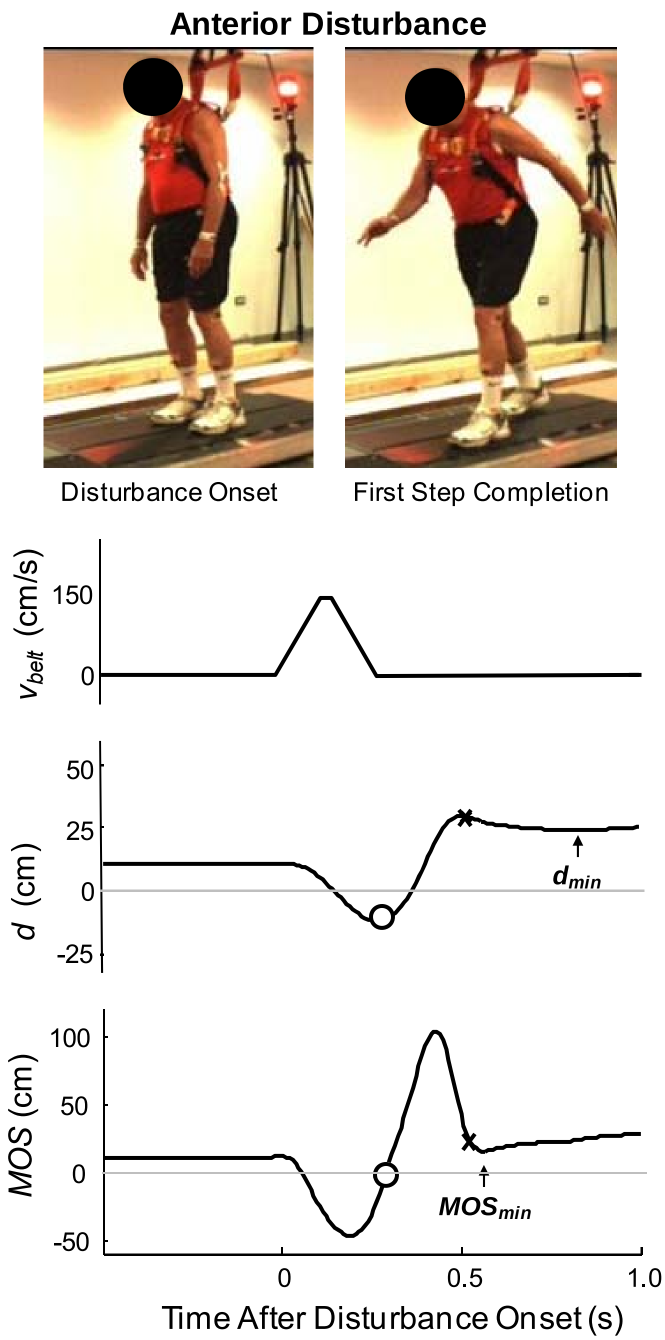

Figure 7. Time series for v_{belt}, d, and MOS of an older subject recovering from an anterior disturbance (v_{belt} = 150 cm/s, displacement = 25 cm) with one step.

[a] The stability measures, *MOS* and *d* (Equation 3.1), are displayed for the stepping (left) limb.

[b] Toe-off is marked by ○. Step completion is marked by X.

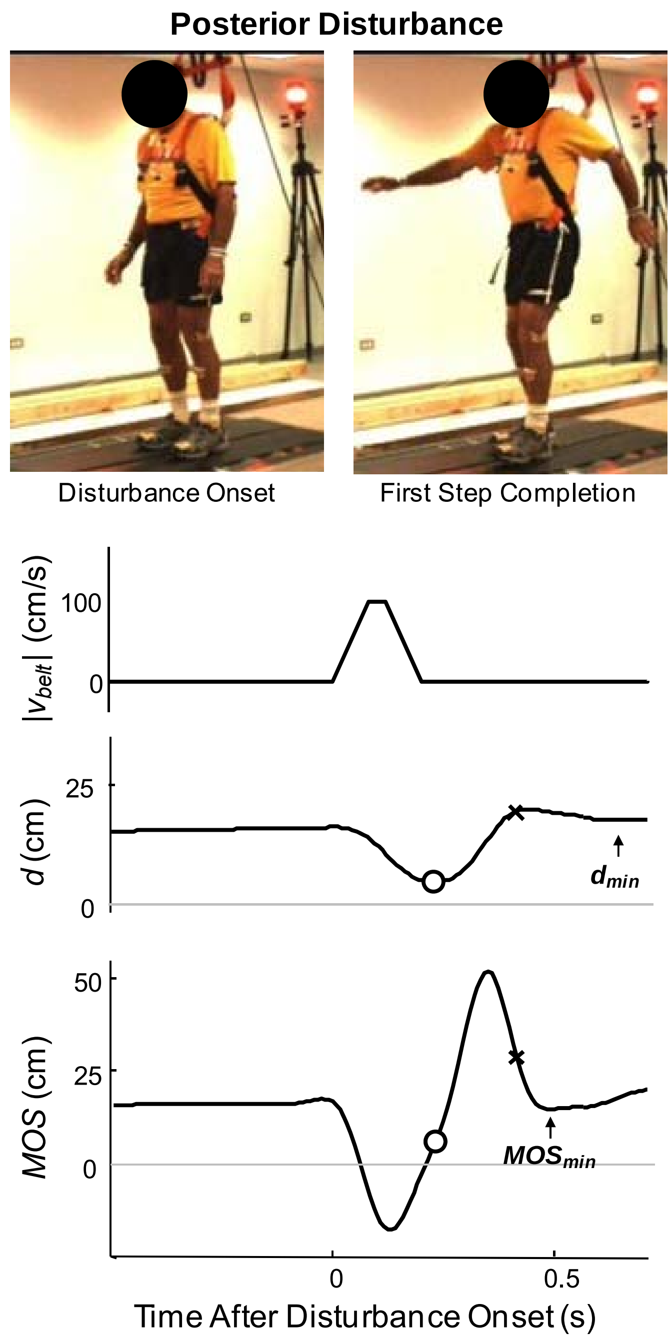

Figure 8. Time series for v_{belt}, d, and MOS of an older subject recovering from a posterior disturbance (v_{belt} = 100 cm/s, displacement = 10 cm) with one step.

[a] The stability measures, *MOS* and *d* (Equation 3.1), are displayed for the stepping (right) limb.

[b] Toe-off is marked by O. Step completion is marked by X.

beyond the initial step, that is, more anterior than the initial forward steps or more posterior than the initial backward steps. Subjects were notified if a second step extended beyond the first step.

The range of disturbance velocities and displacements were chosen with the goal of eliciting single and multiple-stepping responses by all age groups. The disturbances delivered by the treadmill were defined by “velocity profiles” that were trapezoidal in shape (Figures 7 and 8). Each velocity profile had three phases, including an acceleration phase, a constant velocity phase, and a deceleration phase. Disturbance direction was described relative to the direction of the step necessary to avoid a fall. For anterior disturbances, that is, posteriorly-directed belt movements that required forward steps, absolute peak disturbance velocities (v_{belt}) were 100, 126, 150, and 176 cm/s. For posterior disturbances, that is, anteriorly-directed belt movements that required backward steps, v_{belt} was 90, 100, 110, or 120 cm/s. For anterior disturbances, the absolute displacements ranged from approximately 20 to 75 cm. For the posterior disturbances, the absolute displacements ranged from approximately 10 to 65 cm. Disturbance displacements were separated by approximately 5 cm increments (Figure 9, all circles). The acceleration and deceleration phases were 80 to 200 ms in duration, and ranged from 1100 cm/s^2 to 1333 cm/s^2. Minimum displacements were limited by the acceleration capabilities of the treadmill (i.e. with the maximum treadmill acceleration, v_{belt} could not be achieved without surpassing a minimum displacement). Maximum displacements were limited by treadmill length.

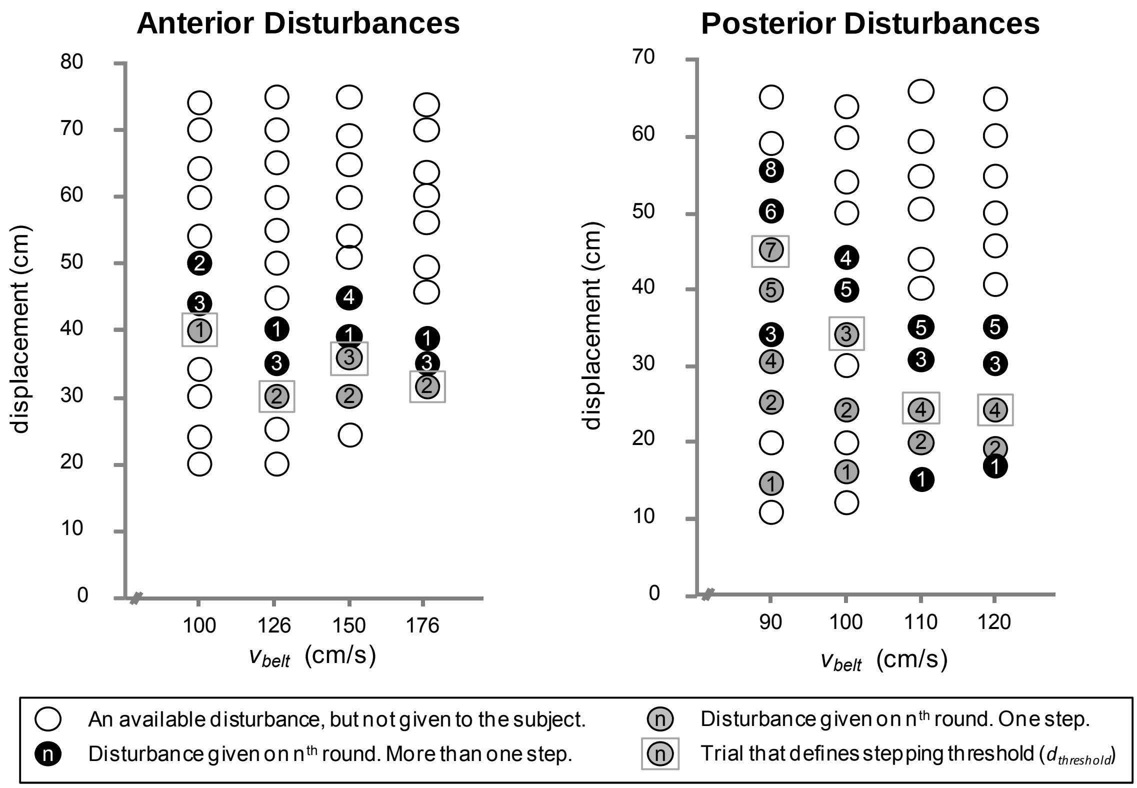

Figure 9. An example of an older subject's progression of disturbances.

[a] The available disturbances are represented by circles positioned at their respective absolute displacement and absolute peak velocity (v_{belt}).

[b] White circles represent disturbances that were not delivered to the subject. Black circles represent disturbances that resulted in failed responses. Gray circles represent disturbances that resulted in only one step.

[c] A gray box around the circle represents a disturbance that was used to define the multiple-stepping threshold displacement ($d_{threshold}$) at a given v_{belt}.

[d] The number within the circle represents the block of disturbances in which the disturbance was delivered.

The following progression of disturbances was designed to identify thresholds at each v_{belt} while limiting the total number of disturbances. Initially, subjects were given eight practice trials, one at each v_{belt}. The first four practice trials were anterior disturbances for which v_{belt} sequentially increased. The next four practice trials were posterior disturbances for which v_{belt} sequentially increased. For young adults, the displacements of these initial disturbances were approximately 65 cm for anterior disturbances and approximately 40 cm for posterior disturbances. For middle-aged and older adults, the initial disturbance displacements were approximately 40 cm for anterior disturbances and approximately 15 cm for posterior disturbances. The following block of eight trials represented the beginning of the test protocol, and consisted of the same eight disturbances presented in the practice block, the order of which was randomized (Figure 9, circles with a “1”). The remainder of the test protocol was comprised of blocks of up to eight disturbances, with each block consisting of a randomized sequence of disturbances of unique v_{belt}. Based on the subject performance (one step or a failed response due to multiple steps or use of the harness), the disturbance absolute displacement at a given v_{belt} was either increased (one step) or decreased (failed response) by two increments (≈ 10 cm) for the subsequent block of disturbances. If this displacement was not available or had already been given to the subject, then a one-increment (≈ 5 cm) change in the displacement was made for the subsequent block. This progression continued until multiple-stepping thresholds were established (Figure 9).

Multiple-stepping thresholds were defined as the largest absolute displacement ($d_{threshold}$) at a given v_{belt} from which a subject could recover with only one step (Figure 9, gray circles within gray boxes). Second steps were disregarded if they did not extend the edge of the base of support in the direction of the disturbance. The value of an established stepping threshold was considered supported by failed responses observed at displacements that were approximately 5 cm and 10 cm (if available) greater than $d_{threshold}$. Subjects were exposed, on average, to 46.4 ± 8.6 total disturbances (maximum = 64, minimum = 31), and subjects were allowed to rest as requested.

Marker positions were recorded at 120 Hz by eight cameras (Cortex, Motion Analysis, Santa Rosa, CA). The three-dimensional coordinates of each marker were exported to text files from which custom programs (MATLAB, The Mathworks, Inc., Natick, MA) were used to calculate dependent variables. Second steps were verified from the positions of the heel or toe markers. The harness was considered to be engaged if greater than 50% body weight was supported.

MOS was calculated as follows (adapted from Hof et al., 2005):

$$MOS = d + \left(\frac{v}{\sqrt{\frac{g}{l}}} \right) \tag{3.1}$$

Subjects were positioned facing the $+x$ direction. For anterior disturbances, $d = x_{toe} - x_{COM}$ and $v = v_{toe} - v_{COM}$. x_{COM} and v_{COM} are the anteroposterior COM position and velocity, respectively. The COM position was estimated from the marker positions on the arms, legs, pelvis, and trunk (Dempster, 1955 via Winter, 2005). x_{toe} and v_{toe} are the anteroposterior position and velocity of the toe marker, respectively. For posterior disturbances, $d = x_{COM} - x_{heel}$ and $v = v_{COM} - v_{heel}$. x_{heel} and v_{heel} are the anteroposterior position and velocity of the heel marker, respectively. The gravity term is $g = 9.81$ m/s^2, and l is the sagittal plane distance between the ankle center and the COM. *MOS* was calculated at disturbance onset (MOS_{DO}). Here, "dynamic stability" is a term that specifically refers to *MOS*. "Stability" is a more general term that applies to both *MOS* and *d*. Both stability measures were calculated at first step completion, that is, stepping-foot contact with the treadmill (d_{step} and MOS_{step}). For successful one-step responses, the minimum of stability values (d_{min} and MOS_{min}) after step completion were identified (Figures 7 and 8).

The primary dependent variable was $d_{threshold}$ expressed as a percentage of body height. If the subject recovered from the largest available displacement with one step, or if the subject was unable to recover from the smallest available displacement with one step, then no $d_{threshold}$ was recorded for that subject at a given v_{belt}. Stability and kinematic variables were calculated for the one-step trials that defined $d_{threshold}$. Step kinematics included step length (L_{step}) and the time after disturbance onset of step contact with the treadmill (t_{step}). t_{step} was also recorded relative to when the treadmill belt stopped moving (t_{step} - t_{belt}). Trunk flexion angle (*TFA*) and trunk flexion angular velocity (*TFV*) relative to vertical were calculated at first step completion. Peak shoulder flexion (posterior disturbances) or extension (anterior disturbances) velocity before first step completion was calculated for the stepping-limb side ($\omega_{SLshoulder}$) and non-stepping limb side ($\omega_{NSLshoulder}$).

For $d_{threshold}$, stability, and kinematic variables, stepwise multiple linear regressions were conducted with v_{belt} (as a percentage of body height), *age*, and an *age**v_{belt} interaction as independent variables. Separate regressions were conducted for anterior and posterior disturbances. The stepping method criterion for entry of a variable into the model was an F-test significance of $p < 0.05$. The stepping method criterion for removal of a variable from the model was an F-test significance of $p < 0.10$. If the final regression model contained an $age*v_{belt}$ interaction, separate one-way analyses of variance (ANOVAs) were conducted at each of the four treadmill belt velocities to determine the presence of differences between young, middle-aged, and older adults. If significant main effects were found, Tukey's post hoc tests were used to evaluate specific between-group differences. All statistics were evaluated using SPSS (SPSS Inc., Chicago, IL) with a significance level of $\alpha = 0.05$.

3.3 **Results**

3.3.1 **Anterior Disturbances**

Compared to middle-aged and older adults, young adults demonstrated a greater ability to recover from the largest displacements with one step (Figure 10) and demonstrated larger $d_{threshold}$ (Figure 11). The final linear regression model ($p < 0.001$, $r^2 = 0.436$) for $d_{threshold}$ (% body height) was the following:

$$d_{threshold} = 43.944 - 0.003*(age*v_{belt}) \qquad (3.2)$$

Based on separate one-way ANOVAs, significant main effects of age group were observed at treadmill belt velocities of 126 cm/s ($p = 0.015$) and 150 cm/s ($p < 0.001$). At a treadmill belt velocity of 126 cm/s, the $d_{threshold}$ of young subjects was significantly longer than that of older subjects ($p = 0.013$, Figure 11). At a treadmill belt velocity of 150 cm/s, the $d_{threshold}$ of young subjects was significantly longer than the $d_{threshold}$ of middle-aged subjects ($p < 0.001$) and older subjects ($p < 0.001$). No young subjects established a $d_{threshold}$ at a treadmill belt velocity of 176 cm/s.

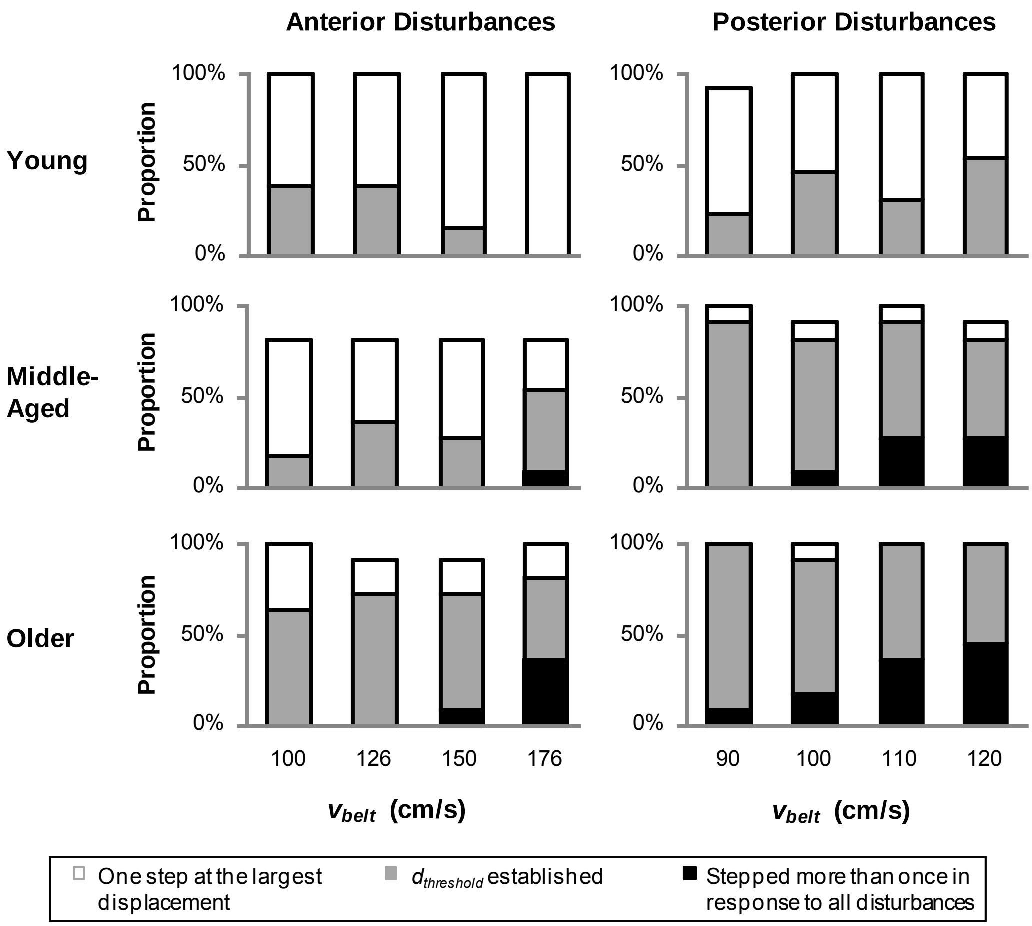

Figure 10. The proportion of subjects who established multiple stepping thresholds ($d_{threshold}$) at each disturbance velocity (v_{belt}), organized by group (young, middle-aged, or older) and disturbance direction (anterior or posterior).

[a] At each v_{belt}, subjects took one step at the largest displacement (white), established a $d_{threshold}$ (gray), or failed single-step responses at all displacements (black).

[b] Proportions do not sum to 100% if investigator error or subject drop-out resulted in no $d_{threshold}$ being established.

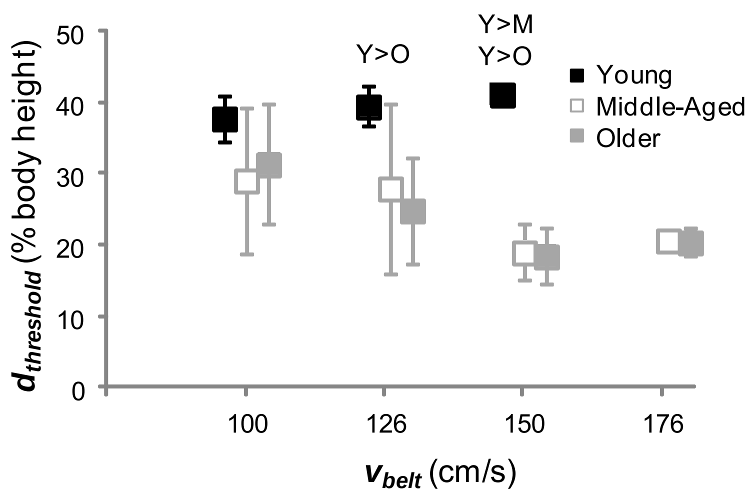

Figure 11. Anterior multiple-stepping thresholds ($d_{threshold}$) at each peak treadmill belt velocity (v_{belt}).

[a] Young (Y) subjects are denoted by black markers, middle-aged (M) subjects are denoted by white markers, and older (O) subjects are denoted by gray markers.

[b] Significant ($p < 0.05$) differences between groups are noted (e.g. young adults had significantly greater $d_{threshold}$ than older adults, Y>O).

The margin of stability associated with the first step became larger with increasing age, but no influence of age was observed for the distance between the COM and the heel marker. The final linear regression model ($p < 0.001$, $r^2 = 0.300$) for MOS_{step} (cm) was the following:

$$MOS_{step} = -17.868 + 0.540 * age \qquad (3.3)$$

The final linear regression model ($p < 0.001$, $r^2 = 0.232$) for MOS_{min} (cm) was the following:

$$MOS_{min} = -24.845 + 0.399*age \quad (3.4)$$

No variables were entered into the stepwise linear regression for MOS_{DO} (total mean ± s.d. = 11.6 ± 1.8 cm). The final linear regression model ($p = 0.003$, $r^2 = 0.149$) for d_{step} (cm) was the following:

$$d_{step} = 36.904 - 0.157*v_{belt} \quad (3.5)$$

No variables were entered into the stepwise linear regression for d_{min} (total mean ± s.d. = 14.1 ± 8.2 cm).

The response of young subjects was characterized by longer step lengths than the responses of middle-aged and older subjects. The final linear regression model ($p = 0.004$, $r^2 = 0.137$) for L_{step} (% body height) was the following:

$$L_{step} = 43.050 – 0.002*(age*v_{belt}) \quad (3.6)$$

Based on a one-way ANOVA, a significant main effect of age group was observed at a treadmill belt velocity of 150 cm/s ($p = 0.001$). The L_{step} of young subjects (47.3 ± 6.5% body height) was significantly longer than middle-aged subjects (28.9 ± 4.0% body height, $p = 0.003$) and older subjects (30.4 ± 4.7% body height, $p = 0.002$). No significant main effects of age group were observed at treadmill belt velocities of 100 cm/s, 126 cm/s, and 176 cm/s ($p \geq 0.051$).

After step completion by young subjects, the treadmill belt moved for a longer duration than after step completion by middle-aged and older subjects. The final linear regression model ($p < 0.001$, $r^2 = 0.568$) for $t_{step} – t_{belt}$ (ms) was the following:

$$t_{step} – t_{belt} = -305.430 + 0.067*(age*v_{belt}) \quad (3.7)$$

Based on separate one-way ANOVAs, a significant main effect of age group was observed at treadmill belt velocities of 100 cm/s ($p = 0.035$), 126 cm/s ($p = 0.01$), and 150 cm/s ($p < 0.001$). At a velocity of 100 cm/s, the $t_{step} – t_{belt}$ of young subjects (-246 ± 44 ms) was significantly more-negative than that of middle-aged subjects (-61 ± 149 ms, $p = 0.033$), but not older subjects (-100 ± 101 ms, $p = 0.098$). At a velocity of 126 cm/s, the $t_{step} – t_{belt}$ of young subjects (-148 ± 47 ms) was significantly more-negative than that of older subjects (71 ± 109 ms, $p = 0.008$), but not middle-aged subjects (12 ± 151 ms, $p = 0.102$). At a velocity of 150 cm/s, the $t_{step} – t_{belt}$ of young subjects (-88 ± 51 ms) was significantly more-negative than that of middle-aged subjects (38 ± 22 ms, $p = 0.001$) and older subjects (61 ± 23 ms, $p < 0.001$). No significant main effects of age group were observed at a treadmill belt velocity of 176 cm/s ($p = 0.428$).

No influence of age was observed for step time, trunk flexion angle, trunk flexion velocity, or shoulder extension velocities. The final linear regression model ($p = 0.006$, $r^2 = 0.127$) for *TFA* (deg) was the following:

$$TFA = 12.542 + 0.197*v_{belt} \quad (3.8)$$

No variables were entered into the stepwise linear regression for t_{step} (total mean ± s.d. = 497 ± 38 ms), *TFV* (-39 ± 42 deg/s), $\omega_{SLshoulder}$ (-260 ± 89 deg/s), or $\omega_{NSLshoulder}$ (-124 ± 65 deg/s).

3.3.2 **Posterior Disturbances**

Young adults demonstrated a superior ability to recover from the largest displacements with one step (Figure 9) and demonstrated the longest posterior $d_{threshold}$ (Figure 12). The posterior multiple-stepping thresholds of young subjects were longer than the thresholds of older subjects. The final linear regression model ($p < 0.001$, $r^2 = 0.373$) for $d_{threshold}$ (% body height) was the following:

$$d_{threshold} = 36.839 - 0.005*(age*v_{belt}) \qquad (3.8)$$

Based on separate one-way ANOVAs, significant main effects of age group were observed at treadmill belt velocities of 90 cm/s ($p = 0.042$), 100 cm/s ($p = 0.012$), and 110 cm/s ($p = 0.003$). The $d_{threshold}$ of young subjects was significantly larger than the $d_{threshold}$ of older subjects at treadmill belt velocities of 90 cm/s ($p = 0.033$), 100 cm/s ($p = 0.009$), and 110 cm/s ($p = 0.002$).

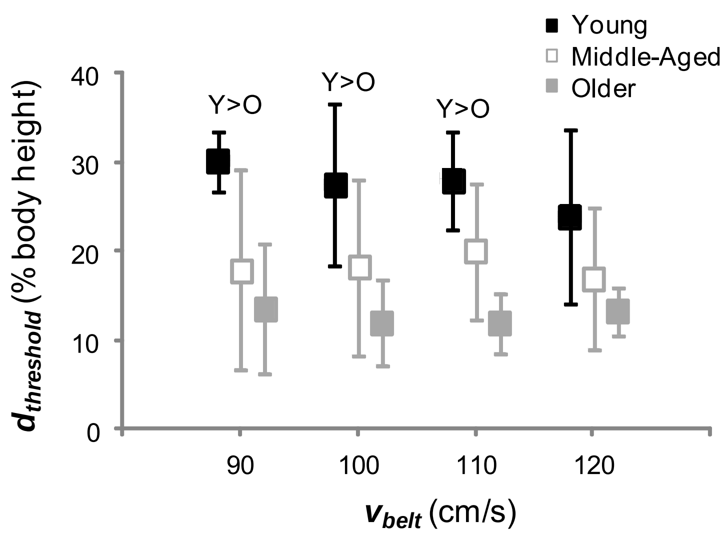

Figure 12. Posterior multiple-stepping thresholds ($d_{threshold}$) at each peak treadmill belt velocity (v_{belt}).

[a] Young (Y) subjects are denoted by black markers, middle-aged subjects are denoted by white markers, and older (O) subjects are denoted by gray markers.

[b] Significant ($p < 0.05$) differences between groups are noted (e.g. young adults had significantly greater values than older adults, Y>O).

The effects of increasing age on stability were inconsistent. With increasing age, the distance between the body COM and the heel marker at step completion was *smaller*. The final linear regression model ($p = 0.025$, $r^2 = 0.066$) for d_{step} (cm) was the following:

$$d_{step} = 28.951 - 0.079 * age \qquad (3.9)$$

Conversely, the minimum margin of stability established after the step was *larger* with increasing age. The final linear regression model ($p < 0.001$, $r^2 = 0.170$) for MOS_{min} (cm) was the following:

$$MOS_{min} = -9.198 + 0.327*age \qquad (3.10)$$

No variables were entered into the stepwise linear regression for MOS_{DO} (total mean ± s.d. = 16.3 ± 1.9 cm), MOS_{step} (52.2 ± 14.3 cm), or d_{min} (20.1 ± 6.2 cm).

Step lengths were shorter and step times were longer with increasing age. The final linear regression model ($p < 0.001$, $r^2 = 0.182$) for L_{step} (% body height) was the following:

$$L_{step} = 33.447 - 0.192*age \qquad (3.11)$$

The final linear regression model ($p < 0.001$, $r^2 = 0.386$) for t_{step} (ms) was the following:

$$t_{step} = 322.179 + 2.305*age \qquad (3.12)$$

After step completion, the treadmill belt moved for a longer duration for young subjects than for older subjects. A negative $t_{step} - t_{belt}$ represents a step that occurred before the treadmill belts stopping moving, and a positive $t_{step} - t_{belt}$ reflects a step that was completed after the belts stopped moving. The final linear regression model ($p < 0.001$, $r^2 = 0.503$) for $t_{step} - t_{belt}$ (ms) was the following:

$$t_{step} - t_{belt} = -374.000 + 0.126*(age*v_{belt}) \qquad (3.13)$$

Based on separate one-way ANOVAs, significant main effects of age group on $t_{step} - t_{belt}$ were observed at all treadmill belt velocities ($p \leq 0.032$). At each velocity, the $t_{step} - t_{belt}$

of young subjects (mean ± s.d. over all velocities = -115 ± 141) were significantly more negative than those of older subjects (189 ± 104, $p \leq 0.025$), but not middle-aged subjects (50 ± 181, $p \geq 0.085$).

With increasing age, trunk extension angle and trunk extension velocity at step completion were smaller. The final linear regression model ($p < 0.001$, $r^2 = 0.254$) for *TFA* (deg, negative values denote extended trunk angles) was the following:

$$TFA = -21.606 + 0.004*(age*v_{belt}) \tag{3.14}$$

Based on separate one-way ANOVAs, significant main effects of age group on *TFA* were observed at a treadmill belt velocity of 100 cm/s ($p = 0.012$). At this velocity, the *TFA* of young subjects (-15 ± 8 deg) were significantly more negative than those of older subjects (-3 ± 6 deg, $p = 0.011$), but not middle-aged subjects (-11 ± 7 deg, $p = 0.494$). No significant main effects of age group on *TFA* were observed at other treadmill belt velocities ($p \geq 0.263$). The final linear regression model ($p = 0.026$, $r^2 = 0.065$) for *TFV* (deg/s, negative values denote trunk extension) was the following:

$$TFV = -46.226 + 0.580*age \tag{3.15}$$

With increasing age, smaller peak shoulder flexion velocity was observed on the stepping-limb side. The final linear regression model ($p < 0.001$, $r^2 = 0.200$) for $\omega_{SLshoulder}$ (deg/s, positive values denote shoulder flexion) was the following:

$$\omega_{SLshoulder} = 512.627 - 3.855*age \tag{3.16}$$

No variables were entered into the stepwise linear regression for $\omega_{NSLshoulder}$ (total mean ± s.d. = 84 ± 127 deg/s).

3.4 Discussion

The purpose of this study was to investigate the influence of age on anterior and posterior thresholds of multiple compensatory steps. It was hypothesized that anterior and posterior stepping thresholds would be reduced with increasing age. This hypothesis was supported.

3.4.1 Anterior Disturbances

Young adults demonstrated a greater ability to recover from anterior disturbances with one step. These results are in accordance with previous findings that older adults take more steps in response to anterior disturbances from surface translations (Brauer et al., 2002; McIlroy and Maki, 1996) and waist pulls (Schulz et al., 2005). Unlike previous research, this study only considered second steps that extended the anterior edge of the base of support. The second step exhibited by older adults in previous studies often did not extend past the first step (McIlroy and Maki, 1996; Schulz et al., 2005). Second steps that do not extend past the first step may reflect age-related deficits in the control of COM frontal plane motion during the initial step (Rogers et al., 2001). By only considering second steps that extended the anterior base of support, however, this study has identified age-related deficits in the control of sagittal plane motion as well. In order to do so, larger disturbance displacements were used in this study (up to 75 cm) than in previous investigations (up to 15 – 30 cm; Brauer et al., 2002, McIlroy and Maki, 1996; Rogers et al., 2001; Schulz et al., 2005).

Although the margin of stability established with a compensatory stepping response increased with age (Equations 3.3 and 3.4), this result does not necessarily imply that an increase in age corresponds with an improved ability to maintain dynamic stability with a compensatory step. Most likely, the *MOS* observed in this study were affected by the continued belt movements observed after step completion by young subjects (negative $t_{step} - t_{belt}$), but not middle-aged and older subjects. For the disturbances included in the present analysis, the displacements and, therefore, durations of the disturbances were longer for young subjects than for middle-aged and older subjects. Compared to steps completed on stationary belts, continued belt movements would decrease the *MOS* established with the step by making the v_{toe} value negative. Subsequently, the v term incorporated into the calculation of *MOS* would become more negative (Equation 3.1).

The only kinematic variable that differed between age groups was step length. Young subjects demonstrated longer L_{step} than middle-aged and older subjects at a v_{belt} of 150 cm/s. At this v_{belt}, young subjects demonstrated larger $d_{thresholds}$ than middle-aged and older subjects (Figure 11). Therefore, a longer step may have been an important aspect in allowing young subjects to achieve longer multiple-stepping thresholds. This result highlights the importance of the first compensatory step length, which has been identified as an important aspect in successfully recovering from large surface translations (Owings et al., 2001) and over-ground trips (Pavol et al., 2001).

Even though older and middle-aged subjects responded to smaller disturbance displacements than young subjects, the resulting trunk flexion angle was not influenced by age. Most likely, an increase in the disturbance displacement is accompanied by an increase in the external impulse applied to the subject. If this is the case, a smaller external impulse had a similar effect on trunk rotation of middle-aged and older subjects as a larger impulse had on young subjects. This result suggests that increasing age diminishes the ability to resist and reverse trunk flexion. Previous research has suggested that age-related declines in the ability to absorb kinetic energy (Hall and Jensen, 2002) and reduce linear momentum of the proximal segments (Jensen et al., 2001) result in a lower threshold for single compensatory steps. Such age-related inabilities to absorb kinetic energy and reduce linear momentum may have resulted in the observed, age-related reduction in anterior thresholds for multiple compensatory steps.

3.4.2 **Posterior Disturbances**

Young adults demonstrated a greater ability to recover from posterior disturbances with one step. These results are in line with previous findings that older adults take more steps in response to posterior disturbances from surface translations (McIlroy and Maki, 1996) and waist pulls (Luchies et al., 1994; Schulz et al., 2005). Larger disturbance displacements were used in this study (up to 65 cm) than in previous investigations (up to 6 – 30 cm; Luchies et al., 1994, McIlroy and Maki, 1996; Schulz et al., 2005). However, $d_{threshold}$ were often established for middle-aged and older adults with displacements less than 30 cm (≈17% body height).

With increasing age, the distance between the COM and the edge of the base of support established with the step decreased. This result is in accordance with previous observations that, compared to young adults, older adults established smaller distances with a step in response to posterior waist pulls (Schulz et al., 2005). Conversely, MOS_{min} became larger with increasing age. As observed for anterior steps, this observation may be attributed to young adults completing the step before the treadmill belt had stopped moving (negative $t_{step} – t_{belt}$). This was not the case for older and middle-aged adults whose steps were completed on a stationary treadmill belt.

The age-related declines in posterior $d_{threshold}$ were most likely due to a shortening of L_{step} and a prolonging of t_{step} that occurred with increasing age. Young adults have previously demonstrated 56 to 79% longer posterior step lengths (29 cm and 14 cm) in response to waist pulls than older adults (16.2 cm and 9 cm, Luchies et al.; Schulz et al., 2005). Posterior L_{step} values of young (46.5 ± 6.9 cm), middle-aged (38.1 ± 15.6 cm), and older adults (33.0 ± 11.4 cm) were longer than previous observations, but were all well within the maximum voluntary posterior step length demonstrated by older adults (≈ 61 cm, Medell and Alexander, 2000). The longer, faster L_{step} of young adults compared to older adults allowed for successful, one-step recoveries and longer d_{step} despite having greater trunk extension angles and velocities.

With increasing age, $\omega_{SLshoulder}$ became smaller. Shoulder flexion moves the COM anteriorly and may reduce trunk extension. In response to a slip, young adults have demonstrated a greater contribution from shoulder flexion in reducing trunk extension than older adults (Troy et al., 2009). The asymmetric peak $\omega_{SLshoulder}$ and $\omega_{NSLshoulder}$ may represent a strategy to control angular momentum about the body's vertical axis in such a way as to increase L_{step}, a strategy observed for forward steps during trip recovery (Pijnappels et al, 2010).

3.4.3 **Limitations**

The $d_{threshold}$ identified in this study were influenced by limitations on disturbance displacements. The $d_{threshold}$ of young adults may be larger than that reported here, as suggested by the number of young subjects who were able to recover from the largest displacements with one step (Figure 10). The age-related differences in $d_{threshold}$ observed in this study may have resulted from different starting displacements for young adults compared to middle-aged and older adults. The starting displacements were reduced for middle-aged and older adult in order to reduce the incidence of trials that led to falls into the safety harness. Experiencing a fall may decrease the subject's motivation for attempting further one step recoveries, instead acting only to avoid another fall. Despite this precaution, falls still did occur. Two young subjects fell once in response to a posterior disturbance, two middle-aged subjects fell repeatedly in response to anterior and posterior disturbances, and two older subjects fell repeatedly in response to anterior and posterior disturbances. Older and middle-aged subjects also noted discomfort of back muscles. One middle-aged subject chose to end her participation during the protocol due to back muscle discomfort. The risk of muscle discomfort was explicit in the informed consent process and no subjects who participated in the study reported regularly occurring, back-related issues. Consequently, the occurrence of falls and muscle discomfort suggests that this protocol may have limited clinical utility. However, the results of this study can inform targets for future intervention.

Although the effects of increasing age on multiple-stepping thresholds were determined, further study is warranted to better understand what factors influence stepping thresholds. For anterior disturbances, 56.4% of the variance in $d_{threshold}$ was not explained by age and v_{belt}. For posterior disturbances, 62.7% of the variance in $d_{threshold}$ was not explained by v_{belt}. Perhaps greater variance in $d_{threshold}$ could have been explained by including tests of physical function (e.g. strength, power production, reaction time, range of motion). Furthermore, consideration of the within-subject repeatability of multiple-stepping thresholds may be of value if efforts toward clinical applications of these or similar methods are to be pursued.

3.4.4 **Conclusions**

To the best of our knowledge, this is the first study to identify the anterior and posterior displacement thresholds for multiple compensatory steps by young, middle-aged, and older adults. This study demonstrated few differences in thresholds or kinematic variables between adults aged 55-64 and adults aged 65 and older. This suggests, therefore, that fall-risk assessments and interventions may be appropriately considered for adults who are younger than 65 years—an age above which the literature has implicitly established as being of greatest concern. From the results of this study, direction-specific targets can be identified for intervention. For anterior disturbances, focus should be made on reducing trunk flexion and increasing step length. For posterior disturbances, interventions should focus on increasing step length, maintaining or decreasing step time, and incorporating effective use of the upper extremities. Potential avenues for intervention include strength training of the lower extremities (Pijnappels et al., 2008) and compensatory step training (Bieryla et al., 2007; Mansfield et al., 2010). It is not known, however, if improving the ability to recover from a disturbance in one step translates to a reduced fall-risk in daily life.

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4. THE INFLUENCE OF INCREASING AGE ON THE THRESHOLDS OF MULTIPLE-COMPENSATORY STEPS: DISTURBANCES WHILE WALKING

4.1 Introduction

Trips and slips cause the majority (59-77%) of falls by independent, community-dwelling older adults (Berg et al., 1997; Hill et al., 1999). Successful recovery from a trip or slip requires a compensatory stepping response (Pavol et al., 2001; Troy et al., 2008). Age-related deficiencies in the ability to recover from an anterior or posterior disturbance with one compensatory step are well documented (Brauer et al., 2002; Luchies et al., 1994; McIlroy and Maki, 1996; Schulz et al., 2005; Chapter 3). These previous studies delivered external disturbances to subjects in bilateral stance and static posture. In contrast, trips and slips occur as people walk and, consequently, involve different sensory and motor processes prior to and during the response. The muscle response to disturbances of standing posture must compensate for the displacement of the body relative to the base of support. In addition to this role, the muscle response to disturbances during walking must also reposition the displaced swing foot (Berger et al., 1984). To the best of the author's knowledge, it is unknown if the previously observed, age-related declines in compensatory stepping are altered by an initial walking condition. Compared to static posture, an initial walking condition allows for an investigation of compensatory stepping that requires a muscle response that has greater similarity with the response to common fall causes—namely trips and slips.

It was previously determined that, with increasing age, the maximum disturbance displacement from which an individuals can recover with one compensatory step was reduced. In other words, the *multiple compensatory stepping threshold* was reduced with increasing age. The purpose of this study was to extend this previous work by investigating the influence of age on anterior and posterior thresholds of multiple compensatory steps following disturbances applied as subjects walk. To do this, young adults (18-35 years of age), middle-aged adults (55-64 years of age), and older adults (65 years and older) were included in this study. It was hypothesized that anterior and posterior stepping thresholds would be reduced with increasing age. Anteroposterior margin of stability (*MOS*; Curtze et al., 2010; Hof et al., 2005; Hof et al., 2007; Hof et al., 2010) and the distance (*d*) between the body's center of mass (COM) and edge of the base of support were evaluated as indicators of stability during the compensatory stepping response. Step, trunk, and shoulder kinematics in the sagittal plane were evaluated to provide insight on modifiable aspects of the stepping response that may influence stepping thresholds and stability.

4.2 **Methods**

This protocol was approved by the University of Illinois at Chicago Institutional Review Board and subjects provided written, informed consent before participation. Thirteen young adults (male:female = 6:7; age = 31.1 ± 0.8 years; height = 172.5 ± 8.9 cm; mass = 74.4 ± 13.5 kg), 10 middle-aged adults (male:female = 5:5; age = 57.9 ± 2.5 years; height = 169.1 ± 9.8 cm; mass = 77.2 ± 11.7 kg), and 10 older adults (male:female = 5:5; age = 74.7 ± 7.4 years; height = 169.2 ± 11.0 cm; mass = 73.3 ± 14.6 kg) participated in this study. No significant main effects of age group (one-way ANOVA) existed for height ($p = 0.632$) or mass ($p = 0.793$). Subjects were not allowed to participate if they reported neurological or musculoskeletal injuries or disorders that would limit their safe participation. In addition, older and middle-aged adults were screened using DEXA (QDR® 4500 Elite, Hologic®, Inc, Bedford, MA) for a minimum bone mineral density of 0.61 g/cm^2 at the left femoral neck.

Subjects wore their own comfortable walking shoes, and were outfitted with a safety harness. The harness was instrumented with a load cell (Omega Engineering, Inc., Stamford, CT) to measure the force supported by the harness. Twenty-two retroreflective markers were placed on the arms, legs, and torso of each subject (Kadaba et al., 1990). As a warm-up and as a means to familiarize the subjects with the data collection environment, subjects walked on a microprocessor-controlled, stepper motor-driven, dual-belt treadmill (ActiveStep™, Simbex, Lebanon, NH; Figure 1) for five minutes at a velocity of 1.5 m/s. Two older subjects were unable to maintain this velocity. Instead, they walked for five minutes at a velocity of 1.0 m/s.

Disturbances were delivered during walking so that compensatory steps were taken with the preferred stepping limb. For each subject, a preferred stepping limb was determined from the subject's previous response to surface translations delivered while the subject was standing. The preferred stepping limb was the limb with which the subject most frequently took initial, anterior steps in a previous protocol (Chapter 3). If the subject had not participated in the previous protocol, three identical anterior disturbances were delivered to the subject (peak velocity of 100 cm/s, displacement of

65 cm for young subjects and 40 cm for middle-aged and older subjects). The limb with which the subject stepped most frequently was determined to be the preferred stepping limb.

At the beginning of each disturbance trial, the treadmill belts gradually accelerated for 3 seconds (young subjects) or 5 seconds (middle-aged and older subjects) to a velocity of 1.5 m/s (1.0 m/s for two older subjects). While the subjects walked at the constant velocity, the treadmill operating system detected foot strikes of the preferred (anterior disturbances) or non-preferred (posterior disturbances) stepping limb. Foot strikes were detected from the current drawn by the separate motors of the left and right treadmill belts. The disturbance was triggered between the fifth to tenth foot strike after the constant velocity was achieved. Foot-strike selection was pseudorandom and evenly distributed amongst disturbances.

After the pre-selected foot strike had been identified, the velocity of the treadmill belts was maintained for an additional 100 ms. The purpose of this delay was to allow the subject to transition to single-stance before the disturbance was delivered. Despite this delay, the disturbance was consistently triggered in double support for two young subjects and three middle-aged subjects. These subjects were consequently instructed to “take shorter steps” while walking so that they would achieve single-limb stance before responding to the disturbance.

After the 100 ms delay, the treadmill belts decelerated to 0 m/s in 120 ms after which the treadmill belts then followed an anterior or posterior disturbance “velocity profile” that was trapezoidal in shape (Figures 13 and 14). Each velocity profile had

Anterior Disturbance

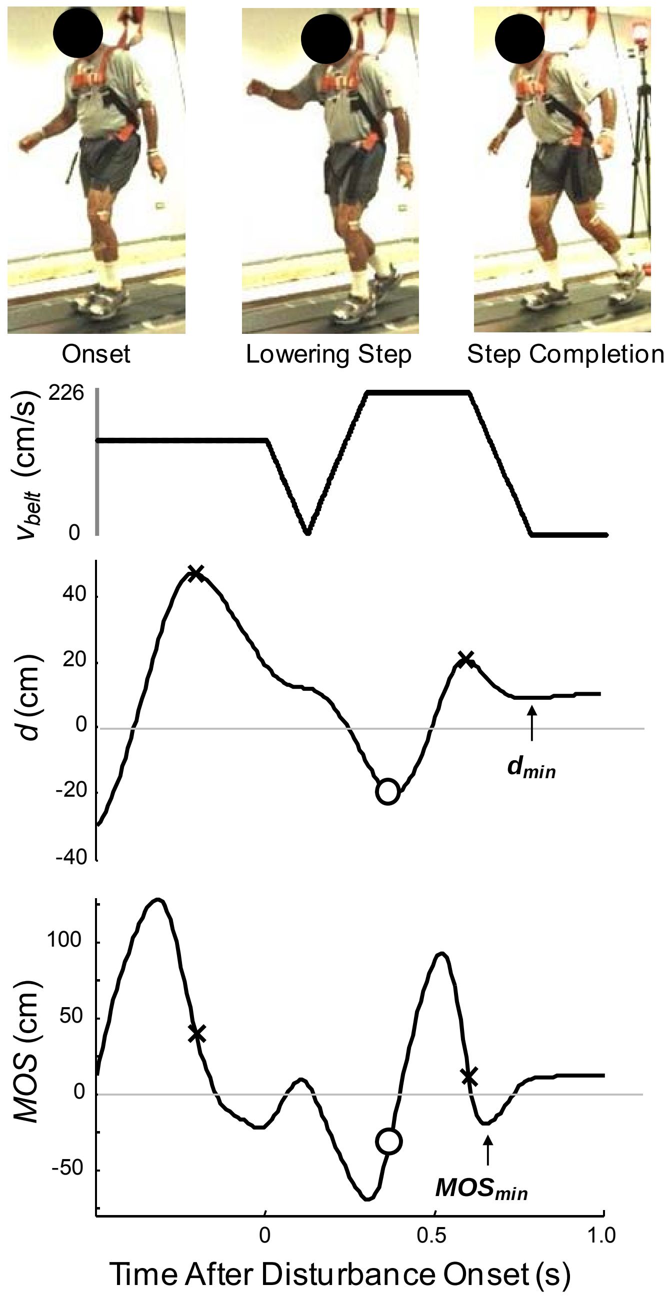

Figure 13. Time series for v_{belt}, d, and MOS of an older subject recovering from an anterior disturbance (v_{belt} = 226 cm/s, displacement = 109 cm) with one compensatory step.

[a] MOS and d (Equation 4.1) are displayed for the right limb. The limb was in stance at disturbance onset (0 s). After a lowering step with the left limb, the right limb was used to perform the compensatory step.

[b] Toe-off is marked by O. Step completion is marked by X. MOS_{min} and d_{min} are identified by an arrow.

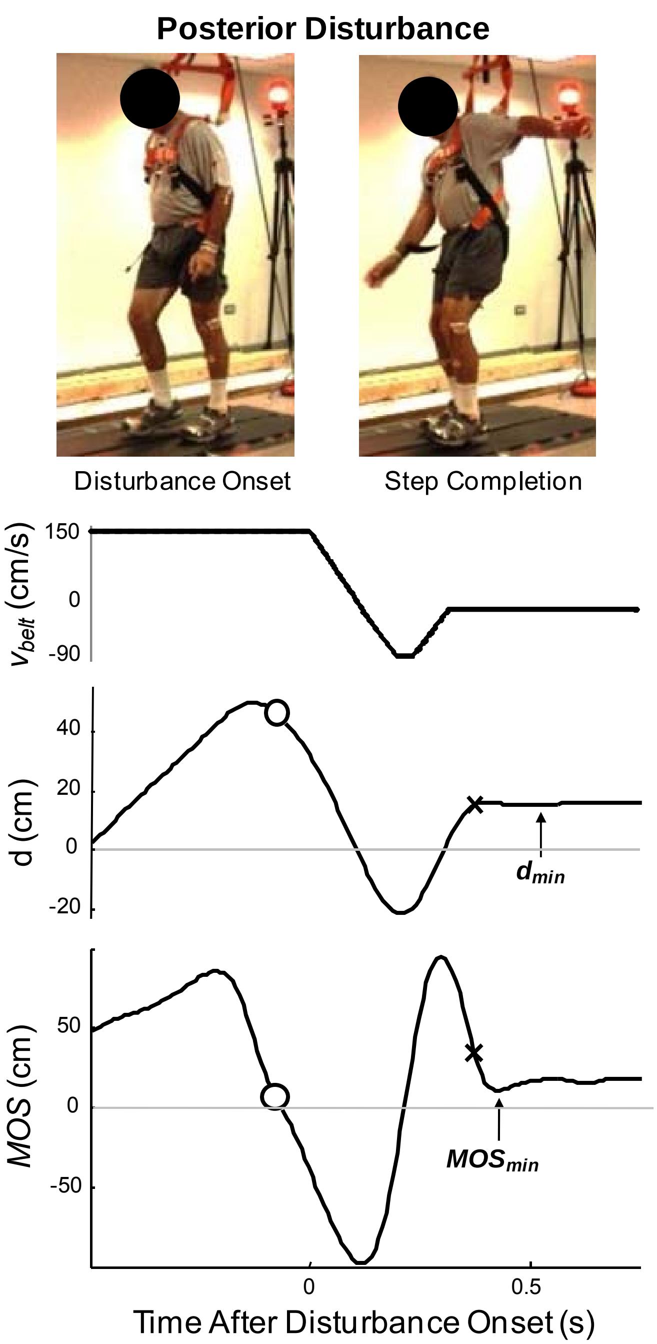
Posterior Disturbance
Disturbance Onset
Step Completion
v_{belt} (cm/s)
150
0
-90
d (cm)
40
20
0
-20
d_{min}
MOS (cm)
50
0
-50
MOS_{min}
0
0.5
Time After Disturbance Onset (s)

Figure 14. Time series for v_{belt}, d, and MOS of an older subject recovering from a posterior disturbance (v_{belt} = 90 cm/s, displacement = 20 cm) with one compensatory step.

[a] MOS and d (Equation 4.1) are displayed for the right limb. The limb was in swing at disturbance onset (0 s).

[b] Toe-off is marked by O. Step completion is marked by X. MOS_{min} and d_{min} are identified by an arrow.

three phases, including an acceleration phase, a constant velocity phase, and a deceleration phase. Disturbance direction was described relative to the direction of the step necessary to avoid a fall. For anterior disturbances, that is, posteriorly-directed belt motion that required forward steps, the absolute peak disturbance velocities (v_{belt}) were 176, 200, 226, or 250 cm/s. For posterior disturbances, that is, anteriorly-directed reversals of belt motion that required backward steps, v_{belt} was 70, 80, 90, or 100 cm/s. For anterior disturbances, the absolute displacement ranged from approximately 50 to 150 cm. For posterior disturbances, the absolute displacements ranged from approximately 10 to 65 cm. Disturbance displacements were separated by approximately 5 cm increments (Figure 15, all circles). The acceleration and deceleration phases were 60 to 200 ms in duration, and ranged from 1125 cm/s^2 to 1333 cm/s^2.

Subjects were instructed to “try to take only one step” in response to disturbances. For posterior disturbances, a successful response entailed one posterior compensatory step. The most common response to an anterior disturbance involved a lowering step, in which the swing limb was lowered to the treadmill, and then an attempted single compensatory step with the contralateral limb. Therefore, a successful response to an anterior disturbance was defined as a response consisting of a lowering step followed by a single compensatory step. Subjects were informed that a second compensatory step was acceptable if it did not extend beyond the initial compensatory step, that is, more anterior to a forward compensatory step or posterior to a backward compensatory step. Subjects were informed by the investigator if the second compensatory step extended past the first compensatory step.

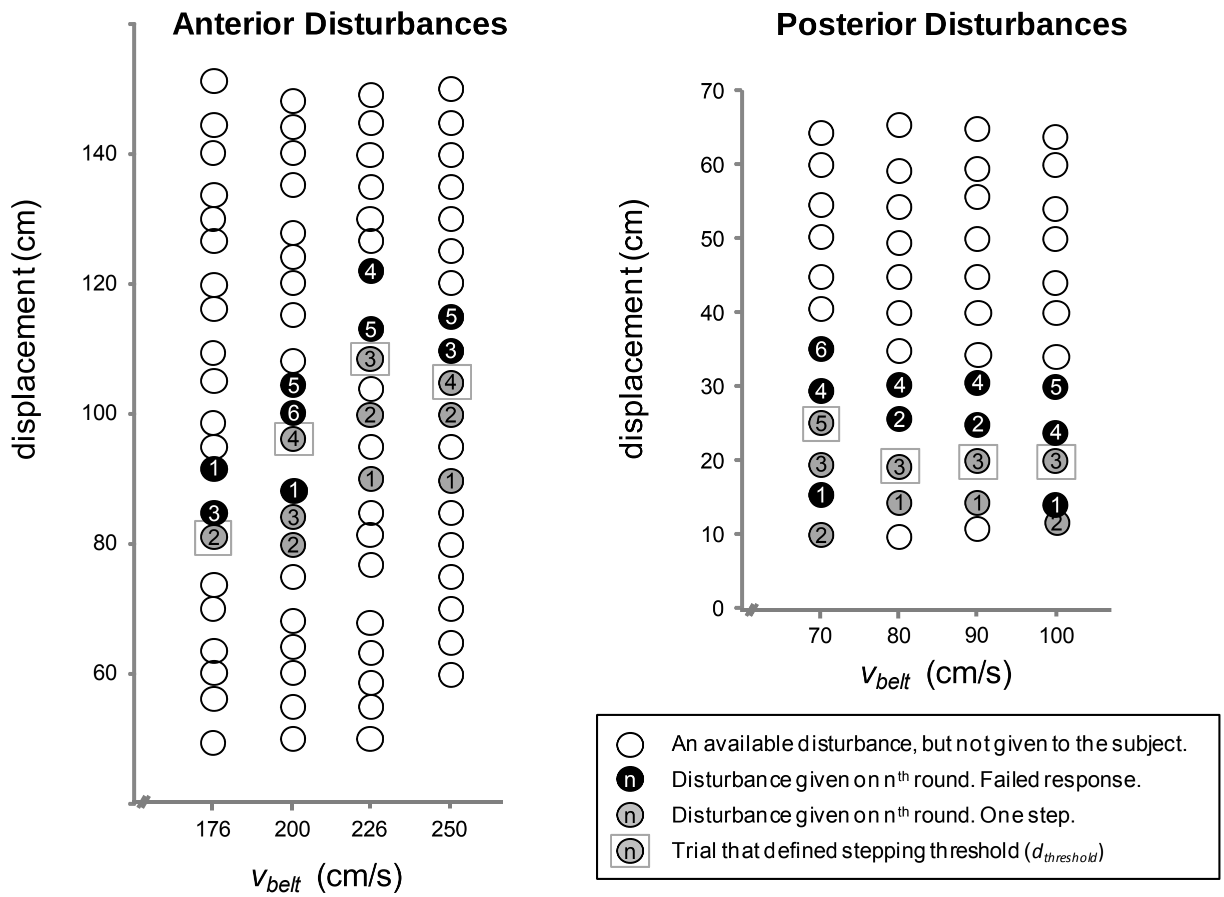

Figure 15. An example of an older subject's progression of disturbances.

[a] The available disturbances are represented by circles positioned at their respective absolute displacement and absolute peak velocity (v_{belt}).

[b] White circles represent disturbances that were not delivered to the subject. Black circles represent disturbances that resulted in failed responses. Gray circles represent disturbances that resulted in only one step.

[c] A gray box around the circle represents a disturbance that was used to define the multiple-stepping threshold displacement ($d_{threshold}$) at a given v_{belt}.

[d] The number within the circle represents the block of disturbances in which the disturbance was delivered.

The following progression of disturbances was designed to identify thresholds at each v_{belt} while limiting the total number of disturbances. Initially, subjects were given eight practice trials, one at each v_{belt}. The first four practice trials were anterior disturbances for which v_{belt} sequentially increased. The next four practice trials were posterior disturbances for which v_{belt} sequentially increased. Initial anterior disturbances had displacements of approximately 90 cm. Initial posterior disturbances had displacements of approximately 25 cm for young subjects and 15 cm for middle-aged and older subjects. The following block of eight trials represented the beginning of the test protocol, and consisted of the same eight disturbances as in the practice block but presented in a randomized order (Figure 15, circles with a "1"). The remainder of the test protocol was comprised of blocks of up to eight disturbances, with each block consisting of randomized disturbances of unique velocities. Subject responses were either successful responses consisting of one compensatory step, or were failed responses consisting of multiple steps or use of the harness. Based on the outcome, the disturbance absolute displacement at a given v_{belt} was either increased (one step) or decreased (failed response) by two increments (10 cm) for the subsequent block of disturbances. If this displacement was not available or had already been given to the subject, then a one-increment (5 cm) change in the displacement was made for the subsequent block. This progression continued until multiple-stepping thresholds were established (Figure 15).

Thresholds were defined as the largest absolute displacement ($d_{threshold}$) at a given v_{belt} from which a subject could recover with only one compensatory step (Figure 15, gray circles within gray boxes). Second steps were disregarded if they did not extend the edge of the base of support in the direction of the disturbance. The values of the established stepping thresholds were supported by failed responses observed at displacements that were 5 cm and 10 cm (if available) greater than $d_{threshold}$. Subjects were exposed, on average, to 63.9 ± 11.0 total disturbances (minimum = 44, maximum = 79), and subjects were allowed to rest as requested.

Marker positions were recorded at 120 Hz by eight cameras (Cortex, Motion Analysis, Santa Rosa, CA). The three-dimensional coordinates of each marker were exported to text files from which custom programs (MATLAB, The Mathworks, Inc., Natick, MA) were used to calculate dependent variables. The time of disturbance onset, that is, time at which the treadmill belt began to decelerate to 0 m/s, was determined from the anteroposterior velocity of the stance-limb toe marker. Second steps were verified from the positions of the heel or toe markers. The harness was considered to be engaged if greater than 50% body weight was supported.

MOS was calculated as follows (adapted from Hof et al., 2005):

$$MOS = d + \left(\frac{v}{\sqrt{\frac{g}{l}}} \right) \qquad (4.1)$$

Subjects were positioned facing the $+x$ direction. For anterior disturbances, $d = x_{toe} - x_{COM}$ and $v = v_{toe} - v_{COM}$. x_{COM} and v_{COM} are the anteroposterior COM position (Winter, 2005) and velocity, respectively. The COM position was estimated from the marker positions on the arms, legs, and trunk (Dempster, 1955 via Winter, 2005). x_{toe} and v_{toe} are the anteroposterior position and velocity of the toe marker, respectively. For posterior disturbances, $d = x_{COM} - x_{heel}$ and $v = v_{COM} - v_{heel}$. x_{heel} and v_{heel} are the anteroposterior position and velocity of the heel marker, respectively. The gravity term is g = 9.81 m/s^2, and l is the sagittal plane distance between the ankle center and the COM. Here, "dynamic stability" is a term that specifically refers to *MOS*. "Stability" is a more general term that applies to both *MOS* and *d*. Both stability measures were calculated at first compensatory step completion, that is, stepping-foot contact with the treadmill (d_{step} and MOS_{step}). For successful one-step responses, the minimum of stability values (d_{min} and MOS_{min}) after compensatory step completion were identified.

The primary dependent variable was $d_{threshold}$ expressed as a percentage of body height. If the subject recovered from the largest available displacement with one compensatory step, or if the subject was unable to recover from the smallest available displacement with one compensatory step, then no $d_{threshold}$ was recorded for that subject at a given v_{belt}. Stability and kinematic variables were calculated for the one-step trials that defined $d_{threshold}$. Step kinematics included compensatory step length (L_{step}) and the time after disturbance onset of step completion (t_{step}). Also, step time was measured relative to when the treadmill belt stopped moving (t_{step} - t_{belt}). For anterior disturbances, lowering step length (L_{low}) and time after disturbance onset (t_{low}) were measured. Trunk flexion angle (*TFA*) and trunk flexion angular velocity (*TFV*) relative to vertical were calculated at first compensatory step completion. Peak shoulder flexion velocity (posterior disturbances) or extension velocity (anterior disturbances) before completion of the first compensatory step was calculated for the compensatory stepping-limb side ($\omega_{SLshoulder}$) and non-stepping limb side ($\omega_{NSLshoulder}$).

For $d_{threshold}$, stability, and kinematic variables, stepwise multiple linear regressions were conducted with v_{belt} (as a percentage of body height), *age*, and an *age**v_{belt} interaction as independent variables. Separate regressions were conducted for anterior and posterior disturbances. The stepping method criterion for entry of a variable into the model was an F-test significance of $p < 0.05$. The stepping method criterion for removal of a variable from the model was an F-test significance of $p < 0.10$. If the final regression model contained an *age**v_{belt} interaction, separate one-way analyses of variance (ANOVAs) were conducted at each of the four treadmill belt velocities to determine the presence of differences between young, middle-aged, and older adults. In order to assess if proactive adjustments were made when subjects anticipated a disturbance, average step lengths were compared between the five-minute walk and the gait preceding a disturbance. This assessment was done with a mixed-model ANOVA with within-subject factors (condition) and between-subject factors (age group). Subjects who were instructed to take shorter steps, as well as the older subjects who walked at a slower velocity, were removed from this part of the analysis. For all ANOVAs, Tukey's *post hoc* analyses were used to evaluate group differences subsequent to finding significant main effects. All statistics were evaluated using SPSS (SPSS Inc., Chicago, IL) with a significance level of $\alpha = 0.05$.

4.3 **Results**

4.3.1 **Step Kinematics Before the Disturbance**

Subjects walked with shorter step length during the trials in which a disturbance was anticipated. The step lengths were, on average, 1.2 percent body height shorter than when no disturbance was anticipated ($p < 0.001$). During the five minute walk, steps were 38.2 ± 0.4% body height. During the trials in which a disturbance was anticipated, the step length was 37.0 ± 0.4% body height. The main effect of age group ($p = 0.843$) and the interaction of age group and condition ($p = 0.487$) on average step length were not significant. Disturbance onset occurred 206 ± 57 ms (mean ± standard deviation) after heel strike and 17 ± 100 ms after toe off. The majority of subjects performed the compensatory steps with their right limb. Twenty-six subjects received disturbances requiring right compensatory steps, and seven subjects received disturbances requiring left compensatory steps.

4.3.2 **Anterior Disturbances**

Increasing age was associated with a small decline in the ability to respond to anterior displacements with one compensatory step. Young subjects demonstrated a greater frequency of responding to the largest displacements with one compensatory step than middle-aged and older subjects (Figure 16). Increasing age was associated with a decrease in the anterior multiple-stepping threshold (Figure 17). The final linear regression model ($p = 0.025$, $r^2 = 0.069$) for $d_{threshold}$ (% body height) was the following:

$$d_{threshold} = 76.427 - 0.175 * age \qquad (4.2)$$

The majority of subjects consistently performed a lowering step before the compensatory step. However, one young subject consistently attempted a single step with the limb that was in swing at disturbance onset. With such a response, no threshold could be established due to constraints of the treadmill length (i.e. the subject's non-stepping foot would come off the back of the treadmill).

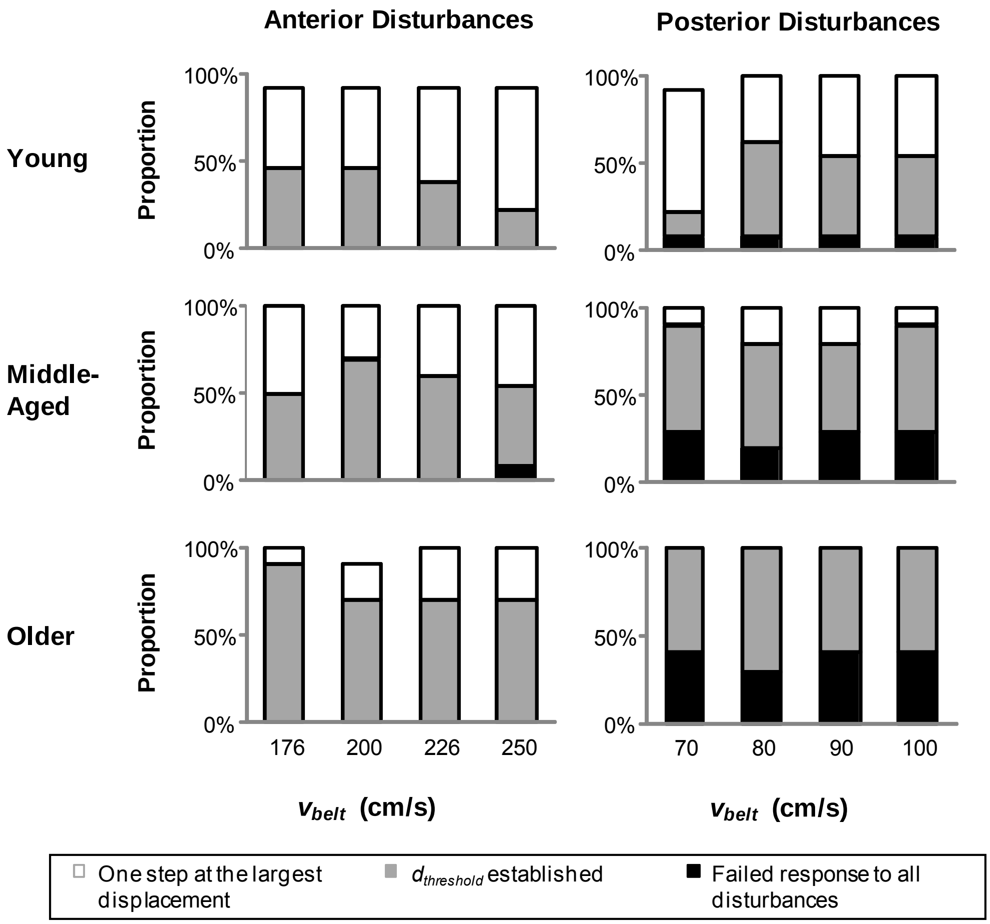

Figure 16. The proportion of subjects who established multiple stepping thresholds ($d_{threshold}$) at each disturbance velocity (v_{belt}), organized by age group (young, middle-aged, or older) and disturbance direction (anterior or posterior).

[a] At each v_{belt}, subjects took one compensatory step at the largest displacement (white), established a $d_{threshold}$ (gray), or failed single-step responses at all displacements (black).

[b] Proportions do not sum to 100% if investigator error resulted in no $d_{threshold}$ being established or if a subject did not use a lowering step before an anterior compensatory step.

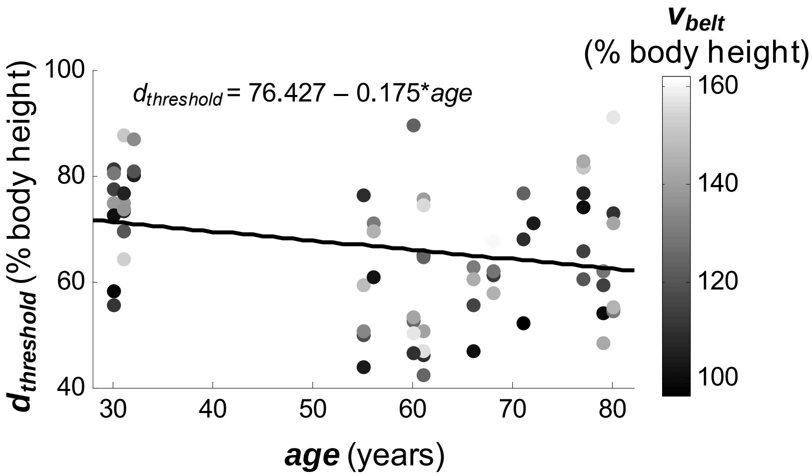

Figure 17. Anterior multiple-stepping thresholds ($d_{threshold}$) as a function of *age*.

[a] Increasing v_{belt} is represented by a gradient transition from black to grey. v_{belt} and the $age*v_{belt}$ interaction were not chosen by the stepwise regression.

Increasing age had no significant influence on stability measures. The final linear regression model ($p = 0.034$, $r^2 = 0.062$) for MOS_{step} (cm) was the following:

$$MOS_{step} = 50.890 - 0.230*v_{belt} \quad (4.3)$$

The overall mean and standard deviation for MOS_{step} was 21.9 ± 16.7 cm. The final linear regression model ($p < 0.001$, $r^2 = 0.165$) for d_{step} (cm) was the following:

$$d_{step} = 39.053 - 0.001*(age*v_{belt}) \quad (4.4)$$

Separate one-way ANOVAs revealed a significant main effect of age group at a treadmill belt velocity of 176 cm/s ($p = 0.043$). However, at this velocity, *post hoc* analyses revealed no significant between-group differences ($p \geq 0.063$). No significant main effects of age group ($p \geq 0.161$) were observed at the remaining treadmill belt velocities. The overall mean and standard deviation for d_{step} was 28.6 ± 9.4 cm. No variables were entered into the stepwise linear regressions for MOS_{min} (total mean ± s.d. = -14.6 ± 17.9 cm) or d_{min} (12.9 ± 9.3 cm).

The only step-related kinematic variable to be significantly influenced by age was lowering step length, which became shorter with increasing age. The final linear regression model ($p = 0.006$, $r^2 = 0.101$) for L_{low} (% body height) was the following:

$$L_{low} = 28.410 - 0.086 * age \quad (4.5)$$

No variables were entered into the stepwise linear regressions for L_{step} (total mean ± s.d. = 32.3 ± 9.2% BH), t_{low} (275 ± 82 ms), or t_{step} (645 ± 102 ms). The final linear regression model ($p = 0.022$, $r^2 = 0.072$) for $t_{step} - t_{belt}$ (ms) was the following:

$$t_{step} - t_{belt} = -137.770 + 0.011 * (age * v_{belt}) \quad (4.6)$$

Separate one-way ANOVAs revealed no significant main effect of age group on $t_{step} - t_{belt}$ at any treadmill belt velocities ($p \geq 0.433$; total mean ± s.d. = -58 ± 13 ms).

Trunk and shoulder kinematic variables were not influenced by age. The final linear regression model ($p < 0.001$, $r^2 = 0.274$) for TFA (deg) was the following:

$$TFA = 2.814 + 0.001 * (age * v_{belt}) \quad (4.7)$$

Separate one-way ANOVAs revealed no significant main effect of age group on TFA at any treadmill belt velocities ($p \geq 0.122$; total mean ± s.d. = 13.6 ± 7.5 deg). No variables were entered into the stepwise linear regressions for *TFV* (total mean ± s.d. = 1 ± 23 deg/s), $\omega_{SLshoulder}$ (-147 ± 89 deg/s), or $\omega_{NSLshoulder}$ (-210 ± 129 deg/s).

4.3.3 **Posterior Disturbances**

Increasing age was associated with a decline in the ability to respond to posterior displacements with one step. Compared to older and middle-aged subjects, a larger proportion of young subjects responded to the largest displacements with one compensatory step (Figure 16). Conversely, a larger proportion of middle-aged and older subjects were unable to respond to any posterior disturbances with one step. In response to the lowest disturbance velocity, the posterior multiple-stepping thresholds of older subjects were shorter than the thresholds of young and middle-aged subjects. The final linear regression model ($p = 0.002$, $r^2 = 0.158$) for $d_{threshold}$ (% body height) was the following:

$$d_{threshold} = 33.752 - 0.004*(age*v_{belt}) \qquad (4.8)$$

Based on separate one-way ANOVAs, significant main effect of age group was observed at a treadmill belt velocity of 70 cm/s ($p = 0.042$). At this treadmill belt velocity, the $d_{threshold}$ of older subjects was significantly shorter than the $d_{threshold}$ of young subjects ($p = 0.018$) and middle-aged subjects ($p = 0.024$, Figure 18).

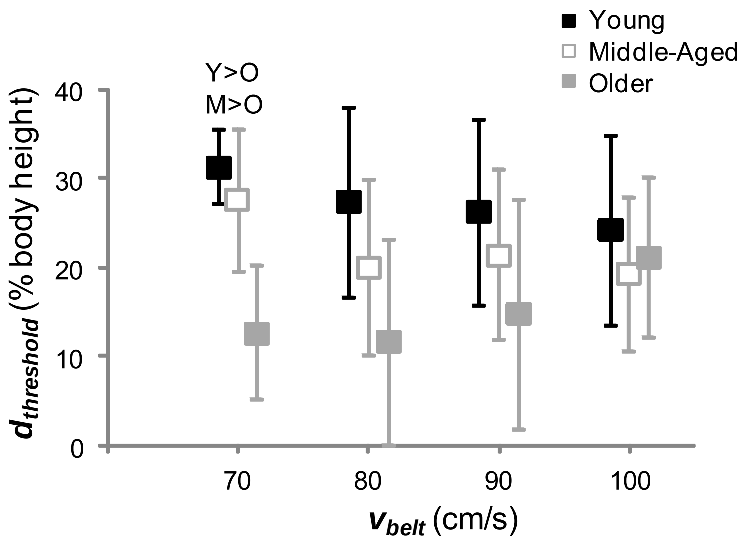

Figure 18. Posterior multiple-stepping thresholds ($d_{threshold}$) at each peak treadmill belt velocity (v_{belt}).

[a] Young (Y) subjects are denoted by black markers, middle-aged (M) subjects are denoted by white markers, and older (O) subjects are denoted by gray markers.

[b] Significant ($p < 0.05$) differences between groups are noted (e.g. young adults had significantly greater $d_{threshold}$ than older adults, Y>O).

With increasing age, the distance between the body COM and the heel marker was shorter. The final linear regression model ($p < 0.001$, $r^2 = 0.354$) for d_{step} (cm) was the following:

$$d_{step} = 46.300 - 0.303 * age \qquad (4.9)$$

The final linear regression model ($p < 0.001$, $r^2 = 0.194$) for d_{min} (cm) was the following:

$$d_{min} = 30.222 - 0.203 * age \qquad (4.10)$$

No variables were entered into the stepwise linear regressions for MOS_{step} (total ± mean = 56.3 ± 22.6 cm) or MOS_{min} (4.6 ± 12.1 cm)

Step lengths were shorter with increasing age. The final linear regression model ($p < 0.001$, $r^2 = 0.300$) for L_{step} (% body height) was the following:

$$L_{step} = 40.789 - 0.304 * age \tag{4.11}$$

No variables were entered into the stepwise linear regressions for t_{step} (total mean ± s.d. = 379 ± 84 ms). Compared to the timing of steps by older subjects, the treadmill belts moved for a longer duration after step completion by the young and middle-aged subjects. The final linear regression model ($p < 0.001$, $r^2 = 0.288$) for $t_{step} - t_{belt}$ (ms) was the following:

$$t_{step} - t_{belt} = -450.014 + 0.116 * (age * v_{belt}) \tag{4.12}$$

Separate one-way ANOVAs revealed a significant main effect of age group at treadmill belt velocities of 70 cm/s ($p = 0.012$) and 90 cm/s ($p = 0.046$). At the treadmill belt velocity of 70 cm/s, older subjects (1 ± 149 ms) had significantly less-negative $t_{step} - t_{belt}$ than young subjects (-398 ± 86 ms, $p = 0.020$) and middle-aged subjects (-333 ± 184 ms, $p = 0.021$). At the treadmill belt velocity of 90 cm/s, older subjects (15 ± 207 ms) had significantly less-negative $t_{step} - t_{belt}$ than young subjects (-264 ± 91 ms, $p = 0.046$), but not middle-aged subjects (-264 ± 91 ms, $p = 0.672$). No significant main effects of age group on $t_{step} - t_{belt}$ were observed at the remaining treadmill belt velocities ($p \geq 0.087$).

Less trunk extension was observed with increasing age. The final linear regression model ($p = 0.006$, $r^2 = 0.123$) for *TFA* (deg, negative values denote an extended trunk angle) was the following:

$$TFA = -14.285 + 0.148 * age \qquad (4.13)$$

No variables were entered into the stepwise linear regressions for *TFV* (total mean ± s.d. = -7 ± 50 deg/s).

With increasing age, less shoulder flexion velocity on the stepping-limb side was observed. The final linear regression model ($p = 0.046$, $r^2 = 0.069$) for $\omega_{SLshoulder}$ (deg/s) was the following:

$$\omega_{SLshoulder} = 288.529 - 1.909 * age \qquad (4.14)$$

No variables were entered into the stepwise linear regressions for $\omega_{NSLshoulder}$ (total mean ± s.d. = 224 ± 160 deg/s).

4.4 **Discussion**

The purpose of this study was to investigate the influence of age on anterior and posterior thresholds of multiple compensatory steps using disturbances applied as subjects walked. It was hypothesized that anterior and posterior stepping thresholds would be reduced with increasing age. Although reductions in stepping thresholds were observed with increasing age, the results did not strongly or consistently demonstrate evidence of age-related declines in multiple-compensatory stepping thresholds.

4.4.1 **Anterior Disturbances**

Although increasing age was associated with a decline in $d_{threshold}$ (Equation 4.2, Figure 17), only 6.9% of the variance in $d_{threshold}$ was explained by age. Most likely, the observed declines in thresholds associated with increasing age do not have a strong-enough relationship to be considered meaningful. More conclusive evidence of age-related declines in anterior, multiple-stepping thresholds was observed when disturbances were delivered to subjects in an initial static, standing posture (Chapter 3). The ability to detect age-related influences in the present study may have been limited by the constraints of the treadmill. Less that 50% of young subjects established an anterior $d_{threshold}$ at each treadmill belt velocity, instead recovering from the longest available disturbance with one compensatory step. Longer disturbances were unavailable due to limitations of the treadmill belt length. Presumably, faster treadmill belt disturbances may have evoked more thresholds by young subjects. However, faster treadmill belt velocities were not chosen in order to reduce the occurrence of falls by older adults. Also, because of constraints on treadmill belt acceleration, the minimum available displacement at larger belt velocities may have been too large to capture the thresholds of older and middle-aged subjects.

Given that the relationship between age and stepping thresholds was weak, it is not surprising that stability measures were not strongly influenced by age, either. The ability to recover from a negative MOS_{min} without taking an additional step was most likely associated with steps completed, on average, before the belts stopped moving (negative $t_{step} - t_{belt}$). The deceleration of the treadmill belts likely provided a reaction force that helped subjects to restore dynamic stability (Bothner et al., 2001).

Recovering from negative *MOS* without single (Chapter 2) or additional (Chapter 3) steps has been observed when *MOS* was measured as support surfaces were in motion.

The only significant, albeit weak ($r^2 = 0.101$), relationship between a kinematic variable and age was for lowering step length. Lowering steps are a common response to trips in late swing of young (Eng et al., 1994) and older adults (Pavol et al., 2001). Unlike the responses observed in this study, the lowering step length of a trip recovery may be limited by the presence of the tripping obstacle. In contrast, this is not the case when one's foot prematurely contacts the ground during swing. Therefore, L_{low} is a modifiable aspect of the recovery response that is pertinent to some, but not all, scenarios of trip recovery. The L_{step} observed in this study (32.3 ± 9.2% body height) were of similar length as those observed in response to disturbances from an initial standing position (35.9 ± 7.7% body height, Chapter 3), but were 17% shorter than the recovery step length demonstrated by older adults successfully recovering from a trip (49.4 ± 5.7% body height, Pavol et al., 2001). Although the initial condition of standing or walking did not appear to influence L_{step}, neither task replicated the step length required to recover from an overground trip. The *TFA* observed in this study (13.6 ± 7.5 deg) were about 49% of the *TFA* observed in response to disturbances from an initial standing position (27.9 ± 9.2 deg, Chapter 3) and about 38% of that observed for older adults recovering for a trip (36.0 ± 12.6 deg, Pavol et al., 2001). Despite having larger v_{belt} than that of stationary disturbances, the walking disturbances of this study do not appear to have challenged the ability to control of trunk orientation to the same extent as stationary disturbances or trips. Perhaps other methods of treadmill disturbances, such as obstructing the swing limb with a rope (Cordero et al., 2003) or obstacle (Schillings et al., 1996), moving a platform underneath the treadmill (Shapiro and Melzer, 2010), or anterior pulls of the waist (Misiaszek and Krauss, 2005), may induce greater *TFA*. Further protocol development is needed to design treadmill disturbances that challenge trunk orientation to the extent of an over-ground trip, but also allow for controlled disturbance magnitudes.

4.4.2 **Posterior Disturbances**

Significant reductions in stepping thresholds were only observed at the lowest treadmill belt velocity. As with posterior disturbances, the ability to detect the influence of age on multiple-stepping thresholds may have been limited by the constraints of the treadmill. At each velocity, the majority of young subjects did not establish a $d_{threshold}$. Instead, many young subjects recovered from the largest displacements with one step (Figure 16). Larger displacements were not available due to the treadmill belt length. Larger treadmill belt velocities were not used in order to reduce the incidence of falling and to allow for a minimum displacement below the thresholds of older subjects. Despite this intention to capture the thresholds of older subjects, the ability to respond to *any* disturbance with one step appeared to be reduced with increasing age (Figure 16). Because the disturbance magnitudes were too small to capture the thresholds of many young subjects, and were too large to capture the thresholds of many older subjects, any existing differences in the compensatory stepping ability of young and older subjects were not consistently detected by analyzing $d_{threshold}$.

With increasing age, the distance between the COM and the edge of the base of support was reduced. This trend was observed for the d_{step} (Equation 4.9) and d_{min} (Equation 4.10). A similar influence of increasing age on d_{step} was observed when disturbances were delivered from an initial standing position (Chapter 3). The age-related decline in the distance between the COM and the edge of the base of support was most likely due to age-related reductions in step length (Equation 4.11). L_{step} demonstrated similar effects of age for responses to posterior disturbances while standing (Chapter 3). Control of the distance from the recovery foot to the COM is a critical factor to recovering from a slip on artificial ice (Troy et al., 2008). However, recovery from a slip also requires control of frontal plane COM motion and foot placement, an aspect that is not replicated on anteroposterior platform translations (Troy and Grabiner, 2006) and was not considered in this study.

With increasing age, less shoulder flexion on the stepping-limb side was observed (Equation 4.14). However, age only explained 4.6% of the variance in $\omega_{SLshoulder}$. A similar trend with increasing age was observed for shoulder flexion in response to posterior disturbances from a standing position that did not require a step (Chapter 2) and posterior disturbances from a standing position that required a step (Chapter 3). The upper extremities can play an active role in responding to posterior disturbances in that shoulder flexion contributes to the reduction of trunk extension during slips (Troy et al., 2009). The contribution of the arms is further supported by the observation that restricting arm use during a posterior waist pull during gait results in larger muscle activation of the lower extremities (Misiaszek and Krauss, 2005). Considering the positive contribution that upper extremity rotations can make to the compensatory stepping response (Troy et al., 2009), these age-related trends in use of the upper extremities may warrant further investigation.

4.4.3 **Proactive Changes to Gait**

All age groups significantly reduced their step lengths during trials in which a g a disturbance was anticipated. The reductions in step length were small (≈ 2 cm), and may not be of biomechanical importance. Previous studies have not observed step length modifications on a treadmill when anticipating a stumble (Cordero et al., 2003). However, proactive reductions in step length before an anticipated slip have been observed that brought the COM of the body closer to the stepping foot (Bhatt et al., 2006). Based on these results, the observed reductions in step length may have been proactive adaptations to posterior disturbances.

4.4.4 **Limitations**

The $d_{threshold}$ identified in this study were influenced by constraints on disturbance displacements. Minimum displacements were limited by the acceleration capabilities of the treadmill (i.e. with the maximum treadmill acceleration, v_{belt} could not be achieved without surpassing a minimum displacement). Maximum displacements were limited by treadmill length. Age-specific $d_{threshold}$ may be larger or smaller than reported, as suggested by the number of subjects who did not establish a threshold (Figure 16). The observed, age-related differences in the ability to recover from any posterior disturbances with one step may have resulted from different starting displacements for young adults compared to middle-aged and older adults. As a result, young adults were given more attempts to respond to posterior disturbances with one step as their progression approached the disturbances with minimum displacements. The starting displacements were reduced for middle-aged and older adults in order to reduce the incidence of falls. Experiencing a fall may decrease the subject's motivation for attempting one step, instead acting only to avoid another fall. Despite this precaution, falls still occurred after posterior disturbances. Two young subjects, two middle-aged subjects, and three older subjects experienced falls. Older and middle-aged subjects also noted muscle discomfort, with one older adult choosing to end his participation after the first anterior disturbance due to discomfort in his foot. The occurrence of falls and risk of discomfort suggests that the magnitude of disturbances used in this study may have limited clinical applicability.

For anterior and posterior disturbances, respectively, 93.8% and 84.2% of the variance in $d_{threshold}$ was not explained by linear regression models that considered age and v_{belt}. Perhaps greater variance in $d_{threshold}$ could have been explained by including tests of physical function (e.g. strength, power production, reaction time, range of motion). Furthermore, consideration of the within-subject repeatability of multiple-stepping thresholds may be of value if efforts toward clinical applications of these or similar methods are to be pursued.

4.4.5 **Conclusions**

To the best of the author's knowledge, this is the first study to evaluate age-related declines in multiple compensatory stepping thresholds as subjects walked. Although age-related reductions in thresholds were observed in this study, the observations were inconsistent across velocities. A major limitation of this study was the ability to identify thresholds with the available disturbances. The disturbance displacements were often too small to evoke multiple-stepping responses in young subjects, and were often too large to allow for single-stepping responses by older adults. Further investigations should consider a larger range of disturbance magnitudes. The disturbance velocity profiles developed for this study, however, can serve as a point of departure for developing subsequent studies.

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5. CONCLUSIONS

The purpose of the preceding studies was to investigate the effects of increasing age on anterior and posterior compensatory stepping thresholds ($d_{threshold}$). It was hypothesized that $d_{threshold}$ would be reduced with increasing age. As an exception, it was hypothesized that posterior, single-stepping $d_{threshold}$ would not be reduced with increasing age groups. These hypotheses were partially supported.

- When given anterior disturbances from an initial standing position and instructed to "try not to step," young adults demonstrated larger single-stepping $d_{threshold}$ than that of middle-aged and older adults (Chapter 2). This result supported our hypotheses.
- When subjects were given posterior disturbances from an initial standing position and instructed to "try not to step," increasing age did not significantly influence posterior, single-stepping $d_{threshold}$ (Chapter 2). This result supported our hypotheses.
- When given anterior disturbances from an initial standing position and instructed to "try to take only one step," young adults demonstrated larger anterior, multiple-stepping $d_{threshold}$ than middle-aged and older adults (Chapter 3). These results supported our hypotheses.
- When given posterior disturbances from an initial standing position and instructed to "try to take only one step," young adults demonstrated larger posterior, multiple-stepping $d_{threshold}$ than older adults, but not middle-aged adults (Chapter 3). These results partially supported our hypotheses.

- When anterior disturbances were given during walking and subjects were instructed to "try to take only one step," a significant ($p = 0.025$), yet weakly correlated ($r^2 = 0.069$) decline in anterior, multiple-stepping $d_{threshold}$ was observed with increasing age. Although these results supported our hypotheses, they may not have substantial biomechanical or clinical implications.
- When posterior disturbances were given during walking and subjects were instructed to "try to take only one step," young adults demonstrated significantly larger posterior, multiple-stepping $d_{threshold}$ than middle-aged and older adults. However, significant differences were only observed at the lowest treadmill belt velocity. Therefore, age-related differences in stepping thresholds were not consistently observed across all disturbance velocities. These results partially supported our hypotheses.

The results of the present studies can assist in the identification of viable targets for fall-prevention interventions. With increasing age, the ability to maintain dynamic stability without stepping decreased. When a compensatory step was taken, middle-aged and older adults demonstrated a diminished ability to reduce or reverse trunk rotation and consistently took shorter steps. Reducing trunk rotation during the step and taking a long compensatory step are factors that are relevant to recovering from a trip (Pavol et al., 2001) or a slip (Troy et al., 2008). Therefore, the effectiveness of interventions may be enhanced by an increased focus on step length and the effects of the disturbance on the COM movement and trunk rotation.

A novel finding of the present studies is that compensatory stepping appears to begin degrading at an age as young as 55 years. Consequently, although falls appear to be considered an issue for adults 65 years of age and older, it is not unreasonable to think that the problem begins to manifest itself much earlier. The injury rate associated with falls does not noticeably increase until after the age of 65 (Figure 19; CDC, 2011). However, an innovative approach to fall prevention may be to intervene at a younger age, when the compensatory stepping response has been shown to degrade, but before the resulting rate of fall injuries has been shown to increase. Such an intervention may have prospective benefits on the trained individual as they enter old age.

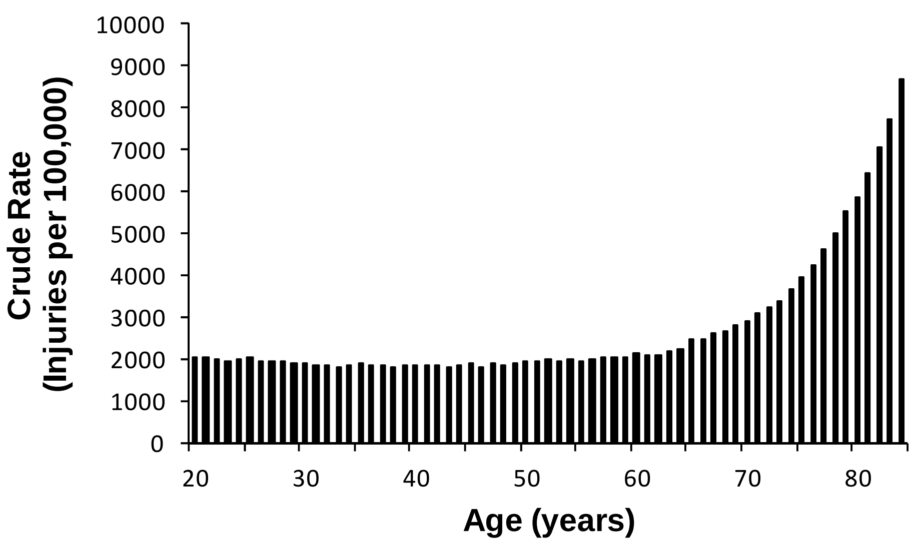

Figure 19. Crude rate of nonfatal injuries due to unintentional falls in the United States (2001-2009; CDC, 2011).

Previous research suggests that the age-related decline in the ability to limit dynamic instability and reduce trunk rotation is due to an insufficient muscular response of the lower extremities. The ability to rapidly develop plantarflexor moments during isometric and isokinetic contractions decreases with age (Thelen et al., 1996). Furthermore, older adults have demonstrated longer ankle muscle onset latencies in response to surface translations (Allum et al., 2002). These age-related deficits suggest that interventions may benefit from a targeted attempt to improve the response of the muscles about the ankle. However, previous research suggests that interventions should also target the muscles about the knee and hip. In response to 10 cm surface translations, no age-related differences between young and older subjects were reported in the maximum muscle moment or the rate of moment production at the ankle (Hall et al., 1999). Compared to young adults, older adults have demonstrated smaller negative muscle power at the knees and hips, but not the ankles, during a non-stepping response to anterior disturbances (Hall and Jensen, 2002). During stepping responses, the moments produced about the ankle, knee, and hip of the stance limb are all likely of importance in reducing angular momentum. Compared to young adults and older non-fallers, older adults who fell in response to a trip produced joint moments less rapidly at the ankle, knee, and hip and produced smaller peak moments about the ankle of the stance limb (Pijnappels et al., 2005). By improving the response of the ankle, knee and hip musculature, the ability of an individual to avoid falling after a disturbance may be improved.

Although targets of intervention have been suggested, and improving the lower-extremity muscle response appears to be an important factor in reaching these targets, the most effective avenues for attaining positive changes remain a focus of current research. Exercise interventions have been shown to reduce fall risk (rate ratio: RR = 0.83-0.90) and fall rate (RR = 0.78-0.86; Chang et al., 2004; Gillespie et al., 2009; Province et al.,1995, Sherrington et al., 2008). The most effective exercise approaches have been suggested to be multiple component group exercise, Tai Chi as a group exercise, and individually prescribed multiple component exercise carried out at home (Gillespie et al., 2009). On average, these three exercise interventions demonstrably decreased fall rate (RR = 0.69) and fall risk (RR = 0.75). The results of the present studies suggest that these results could be improved with exercise interventions focused on the musculature of the lower extremities. Strength training, however, may not be the most effective avenue for realizing positive changes. Previous research suggests that healthy older adults who fall in response to a disturbance have the capability to produce the muscular response necessary for recovery. When given a surface translation requiring multiple forward compensatory steps, older adults who initially fell successfully avoided falling in response to a second, identical disturbance (Owings et al., 2001). Similarly, older adults who initially fell after being tripped successfully recovered from a successive trip (Pijnappels et al., 2005). Therefore, the fall incidence of older adults may be linked to how an individual responds to a disturbance, in contrast to the maximum capabilities of the muscles involved in the response.

By considering the basic principles of physical training and exercise prescription (e.g. *specificity*), opportunities to improve the effects of exercise on fall prevention become evident (Oddsson et al., 2007). The principle of training specificity “requires that a person experiences training conditions (e.g. perturbations exercises) that match real-life conditions (balance recovery situations) as closely as possible" (p. 387, Granacher et al., 2011). Exercise interventions such as Tai Chi or strength training do not provide external disturbances during gait, nor do they generally require compensatory stepping responses. Therefore, exercise interventions that are *specific* to common fall causes, such as trips and slips, may be more effective at decreasing the fall incidence of older adults. Previously executed or suggested interventions that entail a degree of specificity include providing surface translations or waist pulls as people stand, walk in place, or walk on a treadmill (Bieryla et al., 2007; Mansfield et al, 2010; Shimada et al., 2004; Shapiro et al., 2010). Such training has improved the compensatory stepping response to disturbances (Mansfield et al, 2010), and trips (Bieryla et al., 2007), and has prospectively reduced fall incidence (Shimada et al., 2004). The disturbance methods developed for the present studies may serve as an initial point of departure for developing future, specific training regiments. If exercise interventions integrate task-specific training, the potential for exercise to reduce the high fall incidence and injury rate of older adults may be realized.

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APPENDIX

UNIVERSITY OF ILLINOIS
AT CHICAGO

Office for the Protection of Research Subjects (OPRS)
Office of the Vice Chancellor for Research (MC 672)
203 Administrative Office Building
1737 West Polk Street
Chicago, Illinois 60612-7227

Approval Notice
Initial Review (Response To Modifications)

December 23, 2009

Jeremy Crenshaw, MS
Department of Kinesiology and Nutrition
808 S. Wood Street, Rm. 690 CME
Chicago, IL 60612
Phone: (312) 413-9432 / Fax: (312) 413-3699

RE: **Protocol # 2009-1058**
"A Comparison of Compensatory Stepping During Standing and Walking"

Dear Dr. Crenshaw:

Your Initial Review (Response To Modifications) was reviewed and approved by the Expedited review process on December 16, 2009. You may now begin your research

Please note the following information about your approved research protocol:

Protocol Approval Period: December 16, 2009 - December 15, 2010
Approved Subject Enrollment #: 30
Additional Determinations for Research Involving Minors: These determinations have not been made for this study since it has not been approved for enrollment of minors.
Performance Sites: UIC
Sponsor: None
Research Protocol(s):

a) "A comparison of compensatory stepping during standing and walking," Version #1.0, 11/17/2009

Recruitment Material(s):

a) "Research Subjects Needed!," Version #1.0, 11/17/2009
b) Compensatory Stepping, Version #1.0, 11/17/2009
c) Compensatory Stepping, Telephone Script, Version #1.1, 12/08/2009

Informed Consent(s):

a) Compensatory Stepping, Version #1.0, 11/17/2009

Your research meets the criteria for expedited review as defined in 45 CFR 46.110(b)(1) under the following specific categories:

(4) Collection of data through noninvasive procedures (not involving general anesthesia or sedation) routinely employed in clinical practice, excluding procedures involving X-rays or microwaves. Where medical devices are employed, they must be cleared/approved for marketing. (Studies intended to evaluate the safety and effectiveness of the medical device are not generally eligible for expedited review, including studies of cleared medical devices for new indications.)., **(6)** Collection of data from voice, video, digital, or image recordings made for research purposes.

Please note the Review History of this submission:

Receipt Date	Submission Type	Review Process	Review Date	Review Action
11/17/2009	Initial Review	Expedited	11/20/2009	Modifications Required
12/08/2009	Response To Modifications	Expedited	12/16/2009	Approved

Please remember to:

→ Use your **research protocol number** (2009-1058) on any documents or correspondence with the IRB concerning your research protocol.

→ Review and comply with all requirements on the enclosure, "**UIC Investigator Responsibilities, Protection of Human Research Subjects**"

Please note that the UIC IRB has the prerogative and authority to ask further questions, seek additional information, require further modifications, or monitor the conduct of your research and the consent process.

Please be aware that if the scope of work in the grant/project changes, the protocol must be amended and approved by the UIC IRB before the initiation of the change.

We wish you the best as you conduct your research. If you have any questions or need further help, please contact OPRS at (312) 996-1711 or me at (312) 413-7323. Please send any correspondence about this protocol to OPRS at 203 AOB, M/C 672.

Sincerely,

Jennifer Joaquin, MPH
IRB Coordinator, IRB # 1
Office for the Protection of Research Subjects

Enclosure(s):

1. **UIC Investigator Responsibilities, Protection of Human Research Subjects**
2. **Informed Consent Document(s):**
    a) Compensatory Stepping, Version #1.0, 11/17/2009
3. **Recruiting Material(s):**
    a) "Research Subjects Needed!," Version #1.0, 11/17/2009
    b) Compensatory Stepping, Version #1.0, 11/17/2009
    c) Compensatory Stepping, Telephone Script, Version #1.1, 12/08/2009

cc: Charles B. Walter, Department of Kinesiology and Nutrition, M/C 517
Mark D. Grabiner, Faculty Sponsor, Department of Kinesiology and Nutrition, M/C 994

UNIVERSITY OF ILLINOIS
AT CHICAGO

Office for the Protection of Research Subjects (OPRS)
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Approval Notice
Continuing Review

August 24, 2011

Jeremy Crenshaw, MS
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808 S. Wood Street, Rm. 690 CME
Chicago, IL 60612
Phone: (312) 413-9432 / Fax: (312) 413-3699

RE: **Protocol # 2009-1058**
"A Comparison of Compensatory Stepping During Standing and Walking"

Dear Mr. Crenshaw:

Your Continuing Review was reviewed and approved by the Expedited review process on August 19, 2011. You may now continue your research.

Please note the following information about your approved research protocol:

Please note that Appendix M is required when submitting the Continuing Review. Please submit Appendix M via an Amendment per the OPRS policy.

Protocol Approval Period: August 19, 2011 - August 17, 2012
Approved Subject Enrollment #: 50 (35 subjects enrolled) closed to accrual
Additional Determinations for Research Involving Minors: These determinations have not been made for this study since it has not been approved for enrollment of minors.
Performance Sites: UIC
Sponsor: None

Research Protocol(s):

b) "A comparison of compensatory stepping during standing and walking," Version 2.0, 06/29/2010

Your research meets the criteria for expedited review as defined in 45 CFR 46.110(b)(1) under the following specific categories:

(4) Collection of data through noninvasive procedures (not involving general anesthesia or sedation) routinely employed in clinical practice, excluding procedures involving X-rays or microwaves. Where medical devices are employed, they must be cleared/approved for marketing. (Studies intended to evaluate the safety and effectiveness of the medical device are not generally eligible for expedited review, including studies of cleared medical devices for new indications.)
Examples: (a) physical sensors that are applied either to the surface of the body or at a distance and do not involve input of significant amounts of energy into the subject or an invasion of the subject's privacy; (b) weighing or testing sensory acuity; (c) magnetic resonance imaging; (d) electrocardiography, electroencephalography, thermography, detection of naturally occurring radioactivity, electroretinography, ultrasound, diagnostic infrared imaging, doppler blood flow, and echocardiography; (e) moderate exercise, muscular strength testing, body composition assessment, and flexibility testing where appropriate given the age, weight, and health of the individual.
(6) Collection of data from voice, video, digital, or image recordings made for research purposes.

Please note the Review History of this submission:

Receipt Date	Submission Type	Review Process	Review Date	Review Action
08/16/2011	Continuing Review	Expedited	08/19/2011	Approved

Please remember to:

→ Use your **research protocol number** (2009-1058) on any documents or correspondence with the IRB concerning your research protocol.

→ Review and comply with all requirements on the enclosure,
"**UIC Investigator Responsibilities, Protection of Human Research Subjects**"

Please note that the UIC IRB has the prerogative and authority to ask further questions, seek additional information, require further modifications, or monitor the conduct of your research and the consent process.

Please be aware that if the scope of work in the grant/project changes, the protocol must be amended and approved by the UIC IRB before the initiation of the change.

We wish you the best as you conduct your research. If you have any questions or need further help, please contact OPRS at (312) 996-1711 or me at (312) 996-0548. Please send any correspondence about this protocol to OPRS at 203 AOB, M/C 672.

Sincerely,

Brandi L. Drumgole, B.S.
IRB Coordinator, IRB # 1
Office for the Protection of Research Subjects

Enclosure(s):

4. **UIC Investigator Responsibilities, Protection of Human Research Subjects**
5. **Data Security Enclosure**

cc: Charles B. Walter, Department of Kinesiology and Nutrition, M/C 517
Mark D. Grabiner, Faculty Sponsor, Department of Kinesiology and Nutrition, M/C 994

UNIVERSITY OF ILLINOIS
AT CHICAGO

Office for the Protection of Research Subjects (OPRS)
Office of the Vice Chancellor for Research (MC 672)
203 Administrative Office Building
1737 West Polk Street
Chicago, Illinois 60612-7227

Approval Notice
Initial Review (Response to Modifications)

March 11, 2011

Jeremy Crenshaw, MS
Department of Kinesiology and Nutrition
1919 W. Taylor
Room 651, M/C 994
Chicago, IL 60612
Phone: (312) 413-9432 / Fax: (312) 413-3699

RE: Protocol # 2011-0005
"The Influence of the Assessment Task on Dynamic Stability Maintenance in Older Adults"

Dear Mr. Crenshaw:

Your Initial Review (Response to Modifications) was reviewed and approved by the Expedited review process on February 23, 2011. You may now begin your research

Please note the following information about your approved research protocol:

Protocol Approval Period:	February 23, 2011 - January 18, 2012
Approved Subject Enrollment #:	30 Total
Performance Sites:	UIC
Sponsor:	None

Research Protocol(s):

c) Research Protocol: "The influence of the assessment task on dynamic stability maintenance in older adults," Version 2.0, dated 2/10/2011

Recruitment Material(s):

d) "Research Subjects Needed 55 Years or Older," UIC Research Protocol: 2011-0005, Version 2.0, 2/10/2011
e) Internet Advertisement: Older Adults Balance Study, Version 2.0, 2/10/2011
f) Telephone Script: Older Adults Balance Study, Version 2.0, 2/10/2011

Informed Consent(s):

b) Older Adults Balance Study, Version 2.0, 2/10/2011
c) Waiver of Documentation of Consent for Telephone Screening Only, granted under 45 CFR 46.117(c)

Additional Determinations for Research Involving Minors: These determinations have not been made for this study since it has not been approved for enrollment of minors.

Please note the Review History of this submission:

Receipt Date	Submission Type	Review Process	Review Date	Review Action
01/04/2011	Initial Review	Convened	01/19/2011	Modifications Required
02/11/2011	Response To Modifications	Expedited	02/23/2011	Approved

Please remember to:

→ Use your **research protocol number** (2011-0005) on any documents or correspondence with the IRB concerning your research protocol.

→ Review and comply with all requirements on the enclosure,
"**UIC Investigator Responsibilities, Protection of Human Research Subjects**"

Please note that the UIC IRB has the prerogative and authority to ask further questions, seek additional information, require further modifications, or monitor the conduct of your research and the consent process.

Please be aware that if the scope of work in the grant/project changes, the protocol must be amended and approved by the UIC IRB before the initiation of the change.

We wish you the best as you conduct your research. If you have any questions or need further help, please contact OPRS at (312) 996-1711 or me at (312) 355-1404. Please send any correspondence about this protocol to OPRS at 203 AOB, M/C 672.

Sincerely,

Sheilah R. Graham, BS
IRB Coordinator, IRB # 1
Office for the Protection of Research Subjects

Enclosure(s):

6. **UIC Investigator Responsibilities, Protection of Human Research Subjects**
7. **Informed Consent Document(s):**
   b) Older Adults Balance Study, Version 2.0, 2/10/2011
8. **Recruiting Material(s):**
   d) "Research Subjects Needed 55 Years or Older," UIC Research Protocol: 2011-0005, Version 2.0, 2/10/2011
   e) Internet Advertisement: Older Adults Balance Study, Version 2.0, 2/10/2011
   f) Telephone Script: Older Adults Balance Study, Version 2.0, 2/10/2011
9. **Data Security Enclosure**

cc: Charles B. Walter, Department of Kinesiology and Nutrition, M/C 517
Mark D. Grabiner, Faculty Sponsor, Department of Kinesiology and Nutrition M/C 994
Allan Jackimek, Director, Environmental Health and Safety Office, M/C 932

VITA

NAME	Jeremy Richard Crenshaw
EDUCATION	Doctor of Philosophy Movement Sciences University of Illinois at Chicago Chicago, IL 2011
	Master of Science Exercise Science University of Delaware Newark, DE 2007
	Bachelor of Science Exercise Science Minor: Biology Truman State University Kirksville, MO 2003

EMPLOYMENT

2007-2011	RESEARCH ASSISTANT University of Illinois at Chicago Responsibilities include protocol development, obtaining and maintaining internal review board approval, data collection, integration and maintenance of laboratory technology, data processing, software programming, drafting manuscripts, and presenting abstracts. Data collection involves motion analysis, including the measurement of kinetics, kinematics, and electromyography. Motion analysis is used to record overground and treadmill gait, as well as responses to trips, slips, or translations of a microprocessor-controlled platform. Populations of interest include young adults, older adults, individuals with above-knee or below-knee amputations, and individuals with peripheral vascular disease.

2006-2009 TEACHING ASSISTANT
University of Illinois at Chicago
Served as laboratory instructor for Human Physiological Anatomy I and Human Physiological Anatomy II. Served as primary instructor for Weight Training I, Aerobic Conditioning I, and Human Anatomy Workshop for Occupational Therapy.

2004-2006 RESEARCH ASSISTANT
University of Delaware
Responsibilities included data collection, data processing, software programming, and drafting manuscripts and abstracts. Data collection involved motion analysis, including the measurement of kinetic and kinematic variables. Projects involved the analysis of overground gait, as well as the implementation of functional walking and stair-climbing tests. The population of interest was adults with knee osteoarthritis.

2004-2005 TEACHING ASSISTANT
University of Delaware
Primary instructor for Exercise and Conditioning, Strength Training and Conditioning, and Walking for Fitness.

2003 RESEARCH ASSISTANT
University of Kansas
Responsibilities included protocol development, data collection, data processing, and presenting data. Data collection involved motion analysis. The project involved the analysis of upper extremity movements and perceived effort.

TEACHING:

2008-2009 Human Anatomy Workshop for Occupational Therapy
University of Illinois at Chicago
Primary Instructor
Graduate Level
Cadaver-based laboratory course that focused on the skeleton, joints and ligaments, skeletal muscles, brain, spinal cord, brachial plexus, and lumbosacral plexus.

2006-2007	Human Physiological Anatomy I University of Illinois at Chicago Laboratory Instructor Undergraduate Level Cadaver-based laboratory course that focused on the skeleton, joints and ligaments, skeletal muscles, spinal cord, brachial plexus, and lumbosacral plexus.
2006-2007	Human Physiological Anatomy II University of Illinois at Chicago Laboratory Instructor Undergraduate Level Cadaver-based laboratory course that focused on the brain, cranial nerves, sensory organs, vascular system, lymphatic system, respiratory system, digestive system, urinary system and reproductive system.
2006-2007	Weight Training I, Aerobic Conditioning I University of Illinois at Chicago Primary Instructor Undergraduate Level Introduction to exercise, including basic cardiovascular and muscle physiology, exercise principles, and training fundamentals.
2004-2005	Exercise and Conditioning Strength Training and Conditioning Walking for Fitness University of Delaware Primary Instructor Undergraduate Level Introduction to exercise, including basic cardiovascular and muscle physiology, exercise principles, and training fundamentals.

MENTORING

2008	Julie Cain (University of Illinois at Chicago) “Predicting an imminent fall using 3D trunk acceleration”
2008	Gina Maro (University of Illinois at Chicago) “Trip recovery strategies of a transfemoral amputee”

2008 Edward Uram III (University of Illinois at Chicago) "The compensatory stepping response of a transfemoral amputee"

2007 Kris McKinney (University of Illinois at Chicago) "The effect of backward walking practice on step-width variability"

PROFESSIONAL MEMBERSHIPS

2005-2011 American Society of Biomechanics

HONORS AND AWARDS

2006 Outstanding Graduate Student in Exercise Science (University of Delaware)

2003 Departmental Honors (Truman State University)

RESEARCH

Publications

Crenshaw J.R., Rosenblatt, N.J., Hurt, C.P., Grabiner, M.D.: The discriminant capabilities of stability measures, trunk kinematics, and step kinematics in classifying successful and failed compensatory stepping responses by young adults. J Biomech Accepted: 2011.

Hurt, C.P., Rosenblatt, N., **Crenshaw, J.R.**, Grabiner, M.D.: Variation in trunk kinematics influences variation in step width during treadmill walking by older and younger adults. Gait Posture 31;461-464:2010.

Barrios, J.A., **Crenshaw, J.R.**, Royer, T.D., Davis, I.S.: Walking shoes and laterally wedged orthoses in the clinical management of medial tibiofemoral osteoarthritis: a one-year prospective controlled trial. Knee 16;136-142:2009.

In Submission

Crenshaw, J.R., Kaufman, K.R., Grabiner, M.D.: Kinematics of unilateral, above-knee amputees following a trip. Arch Phys Med Rehabil.

Conference Proceedings

Crenshaw, J.R., Cain, J.B., Grabiner, M.D.: Dynamic stability during successful and failed compensatory stepping responses. Proceedings of the American Society of Biomechanics, Providence, RI, 2010.

Crenshaw, J.R., Kaufman, K.R., Grabiner, M.D.: Failed trip recoveries of above-knee amputees suggest possible fall-prevention interventions. Proceedings of the American Society of Biomechanics, Providence, RI, 2010.

Crenshaw, J.R., Kaufman, K.R., Grabiner, M.D.: Compensatory step training of unilateral, above-knee amputees: a potential intervention for reducing trip-related falls. Proceedings of the American Society of Biomechanics, Providence, RI. 2010

Cain, J.B., **Crenshaw, J.R.**, Kaufman, K.R., Grabiner, M.D.: Trunk kinematics discriminate multidirectional falls and recoveries following large postural disturbances. Proceedings of the American Society of Biomechanics, Providence, RI. 2010.

Crenshaw, J.R., Kaufman, K.R., Grabiner, M.D.: Improving dynamic stability during the compensatory stepping response of a transfemoral amputee. Proceedings of the American Society of Biomechanics, State College, PA. 2009.

Cain, J.B., **Crenshaw, J.R.**, Kaufman, K.R., Grabiner, M.D.: Predicting an imminent fall using 3D trunk acceleration. Proceedings of the American Society of Biomechanics, State College, PA. 2009.

Crenshaw, J, Kaufman, K, Grabiner, M.: Trip-recovery strategies of a transfemoral amputee. Proceedings of the North American Congress on Biomechanics, Ann Arbor, MI. 2008.

Royer, T., **Crenshaw, J.**, Barrios, J., Davis, I.: Knee-joint loading variability during gait does not differ between individuals with and without knee osteoarthritis. Proceedings of the North American Congress on Biomechanics, Ann Arbor, MI. 2008.

Rosenblatt, N., **Crenshaw, J.**, Wenning, J., Grabiner, M: Contributions of active dorsiflexion to toe clearance in transtibial amputees: a case study. Proceedings of the North American Congress on Biomechanics, Ann Arbor, MI. 2008.

Barrios, J.A., Davis, I.M., **Crenshaw, J.R.**, Royer, T.D.: Frontal plane mechanics during walking in patients with lateral compartment tibiofemoral knee osteoarthritis with and without a laterally wedged orthosis. Proceedings of the American Society of Biomechanics, Blacksburg, VA. 2006.

Royer TD, **Crenshaw, J.R.**, Barrios J, Davis IM. Wedged shoe orthoses reduce peak medial ground reaction force. *National American College of Sports Medicine Conference*. Denver, CO. 2006.

Crenshaw, J.R., Royer TD, Davis IM, Barrios JA. Long-term effects of wedged orthoses on function and WOMAC scores in subjects with knee osteoarthritis. *National American College of Sports Medicine Conference*, Denver, CO. 2006.

Barrios, J.A., Davis, I.M., **Crenshaw, J.R.**, Royer, T.D.: Effect of laterally wedged orthoses on frontal plane knee mechanics in subjects with medial compartment tibiofemoral osteoarthritis. National American College of Sports Medicine Conference, Denver, CO. 2006.

Crenshaw, J.R., Royer, T.D., Davis, I.M., Crenshaw, S.J., Butler, R.J.: The effect of laterally wedged orthoses on talus angle. Proceedings of the International Society of Biomechanics, Cleveland, OH. 2005.

Crenshaw, J.R., Royer, T.D.: Effects of unilaterally reduced ankle motion on intrasubject gait variability. National American College of Sports Medicine Conference, Nashville, TN. Mid-Atlantic Regional ACSM, Harrisburg, PA. 2005.

Crenshaw, J.R., Johnson, A.M., Bird, M.: Gender differences in change of direction maneuvers with long axis rotation: a preliminary investigation. International Symposium on Biomechanics in Sports, Ottawa, ON, Canada. 2004.

Invited Lectures

Hurt, C.P., **Crenshaw, J.R.** Gait analysis and human walking. Schwab Rehabilitation Institute, Chicago, IL. 2007.

LABORATORY INSTRUMENTATION AND SOFTWARE

Motion Analysis Systems (Cortex and Orthotrak software)

Simbex ActiveStep (operation and disturbance profile development)

Noraxon telemetered electromyography system

AMTI force platform

MATLAB software

LabVIEW software

SPSS software

Microsoft Office software

Vicon motion analysis system

PheoniX Technologies motion analysis system

Basler Pilot high speed digital camera

SERVICE

2008-2010	Grades 5 and 6 science fair judge. Chicago Public Schools.
2008-2009	UIC anatomy laboratory tour coordinator for Chicago area high schools.