BIOMECHANICS OF LEANING AND DOWNWARD REACHING TASKS IN YOUNG AND OLDER WOMEN

by

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Age is an issue of mind over matter. If you don't mind, it doesn't matter.

~Mark Twain

DEDICATION

This doctoral dissertation is dedicated to my wife, family, and friends that helped me along this journey. Thank you for everything! In particular, my wife, Mia, who provided me with a constant reminder of the goal at hand, and was particularly instrumental in my pursuit of a meaningful research topic, with her insightful questions. My family, and mom in particular, who provided steadfast and unwavering support every time we talked of grad school, and friends, whose company made the trip an adventure.

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PREFACE

This dissertation includes five separate chapters (2-6) intended for independent publication in various scholarly journals. For this reason, some repetition of material occurs in the Introduction and Methods of each chapter.

TABLE OF CONTENTS

DEDICATION……………………………………………………….……………………ii

ACKNOWLEDGEMENTS……………………………………………………..………iii

PREFACE…………………………………………………….……………………..……v

LIST OF FIGURES……………………………………………………..………………ix

LIST OF TABLES………………………………………………….…………………xiii

LIST OF APPENDICES………………………………………………….………...xiv

ABSTRACT………………………………………………………….…………………xv

CHAPTER

1. INTRODUCTION……………………………………………………………1

1.1. Epidemiology..……………………………………………….1
1.2. Stooping, Crouching, or Kneeling and Fall Risk…….……….2
1.3. Stooping, Crouching, or Kneeling Movements….…………...3
1.4. Postural Control…………………………...……………….4
1.5. Speed-Accuracy Tradeoffs in COP Movements and Submovements…………………………………………….6
1.6. Biomechanical Modeling..……………………………...7
1.7. Summary………………………………………………...9

2. CHARACTERISTICS OF OLDER ADULTS WITH SELF-REPORTED STOOPING, CROUCHING, OR KNEELING DIFFICULTY…………………………………………………….....18

2.1. Abstract…..…………………………………………………18
2.2. Introduction…………………………………………………19
2.3. Methods…..…………………………………………………20
2.4. Results……………………………………….………………23
2.5. Discussion……………………………..…………………26

2.6. Conclusion……………………………………………………..29
2.7. Significance…………………………………………………29
2.8. References…………………………………………………...30

3. **THE EFFECT OF AGE, MOVEMENT DIRECTION, AND TARGET SIZE ON THE MAXIMUM SPEED OF TARGETED COP MOVEMENTS IN HEALTHY WOMEN…...…………………………....33**
3.1. Abstract…..…………………………………………………33
3.2. Introduction…………………………………………………34
3.3. Methods…..…………………………………………………37
3.4. Results……………………………………….………………43
3.5. Discussion…………………………..…………………….…47
3.6. Conclusion……………………………………………………..52
3.7. Significance…………………………………………………52
3.8. References…………………………………………………...53

4. **AGE-RELATED CHANGES IN SPEED AND ACCURACY DURING RAPID TARGETED COP MOVEMENTS NEAR THE POSTERIOR LIMIT OF THE BASE OF SUPPORT…...………….........................58**
4.1. Abstract…..…………………………………………………58
4.2. Introduction…………………………………………………59
4.3. Methods…..…………………………………………………62
4.4. Results……………………………………….………………70
4.5. Discussion…………………………..…………………….…81
4.6. Conclusion……………………………………………………..85
4.7. Significance…………………………………………………85
4.8. References…………………………………………………...86

5. **AGE-RELATED CHANGES IN POSTURAL CONTROL WHEN PERFORMING DOWNWARD REACH AND PICK-UP MOVEMENTS WITH A LIMITED BASE OF SUPPORT…...……...90**
5.1. Abstract…..…………………………………………………90
5.2. Introduction…………………………………………………91
5.3. Methods…..…………………………………………………93
5.4. Results……………………………………….……………..100
5.5. Discussion…………………………..…………….......106
5.6. Conclusion………………………………………………..…..112
5.7. Significance……………………………………………...113
5.8. References…………………………………………………114

6. **THEORETICAL ANALYSIS OF FACTORS AFFECTING DYNAMIC STABILITY DURING THE MOMENTUM ARREST PHASE OF A DOWNWARD REACH AND PICK-UP TASK: SIMULATIONS WITH A FORWARD MODEL…...………………..119**
6.1. Abstract…..…………………………………………………...119

6.2. Introduction……………………………………………………..120
6.3. Methods….…………………………………………………...122
6.4. Results………………………………….…………………..127
6.5. Discussion…………………………….………………….....130
6.6. Conclusion…………………………………...……….133
6.7. Significance………………………………………………...133
6.8. References………………………………………………….134

7. GENERAL DISCUSSION…...……………………….………………137
7.1. Factors Underlying the Control of SCK Movements……...138
7.2. Speed-Accuracy Tradeoffs in COP Movements and Submovements……………………………………………..140
7.3. Biomechanical Modeling…..………………………………142
7.4. Clinical Implications…..…………………………………...143
7.5. Strengths…..……………………………………………….144
7.6. Limitations…..……………………………………………..146
7.7. References…..……………………………………………...149

8. CONCLUSIONS…...……………………………………………………154

9. RECOMMENDATIONS FOR FUTURE WORK…...………………156

APPENDICES……………………………………...…...…………………..…….158

LIST OF FIGURES

Figure 1.1: Conceptual postural control model of stooping, crouching, or kneeling (SCK) movements, considering feedback control and the center of pressure (COP) as system output. The dashed square box represents the focus of this thesis……………………………..5

Figure 1.2: A) Inverted pendulum model depicting center of mass (COM) and center of pressure (COP) estimation. B) Schematic of COP and COM changes in a quiet stance (Winter, 1995)………………………6

Figure 3.1: A) Illustration of discrete COP movement initiation using auditory cues. B) Exemplary COP movement during movements from a target at a neutral standing posture (dashed light gray lines, home target zone) to a target with a fixed movement amplitude of 6 cm, within the anterior edge of the functional base of support (dashed dark gray lines, anterior target zone) and back (solid light gray line). C) COP velocity trajectory illustrating the zero acceleration crossings used to define submovements. Submovements were defined as a pair of zero acceleration crossings (i.e., COP velocity peaks or throughs) between the movement's peak velocity and arrest of movement……………………………………..…...……………….42

Figure 3.2: COP position and velocity data illustrating the mean (solid line) and standard deviation (shaded area) of anterior (solid center line) and posterior (dashed center line) movements during discrete 2-cm trials, representative of young and older women. Both the anterior target (dashed dark gray lines) and target at a neutral standing posture (dashed light gray lines) are presented in plots of COP position) ……………………………………..44

Figure 3.3: Influence of target size on anterior and posterior COP movements for young and older women. Mean (S.D.) movement speed, number of submovements, and ratio of peak-to-average COP velocity were found to be significantly affected by movement direction and target size in linear mixed

models. Age was found to have a significant effect on movement speed and submovements. * and † indicate $P < .008$ in post-hoc t-tests using Hochberg's step-up method to evaluate age and movement direction differences, respectively……………………..45

Figure 3.4: Effect of age on the relationship between COP movement speed and number of compensatory submovements. Correlations between movement speed and submovements for young and older women ($r^2 = -.72$ to $-.80$) were all statistically significant ($P < .01$) …………………………………………………..46

Figure 4.1: A) Illustration of participant at a neutral standing posture and their representative rigid link model. B) Illustration of experimental setup showing an exemplary small-amplitude COP movement from the anterior target zone (dashed black lines), to the center target zone (dashed dark gray lines) located at a neutral standing posture. Dashed light gray lines indicate the posterior target zone used in large- amplitude movements. Hatched area indicate 6-cm targets, whereas gray shaded areas indicate 2-cm targets…………………………………………………..66

Figure 4.2: Position, velocity, and acceleration of a typical COP movement, decomposed by its primary and secondary submovement. V_1 represents the threshold frequency, equivalent to 10% of the peak posterior velocity, V_{max}. Pairs of acceleration zero crossings after Vmax represent a corrective submovement. Primary submovements are defined by the onset time, T_{on}, and the offset time, T_{off}……………………………………..69

Figure 4.3: Mean ± SD values of A) primary submovement (PSM) speed and B) COP endpoint position during small-amplitude (S2 & S6) and large- amplitude movements (L2 & L6) to 2-cm and 6 cm targets, respectively…………………………………………………..72

Figure 4.4: A) Correlations between speed of the overall movement and its primary submovement (PSM), and B) correlations between movement time and primary submovement endpoint position in 2-cm and 6-cm targets. Data for both hand support conditions are aggregated……………………………………………………………75

Figure 4.5: Mean ± SD COP A) movement speed, B) number of submovements, and C) COP endpoint position for targeted movements with and without hand support. S2 and S6 refer to small-amplitude movements to 2-cm and 6-cm targets, respectively, whereas L2 and L6 refer to large-amplitude

movements to 2-cm and 6-cm targets……………………………………78

Figure 4.6: Distribution of COP endpoint along the center and posterior target for accuracy-constrained COP trials without hand support of 2-cm and 6-cm targets for both young (Y) and older (O) women. Bin sizes for histograms have a 4-mm width. The leftmost dashed line represents the posterior limit of the FBOS………...80

Figure 5.1: A) Illustration of symmetric two-handed downward reaches to a target placed on the floor under challenging reach conditions. B) Full base of support (BOS) and forefoot BOS conditions used for experiments……………………………………………………..96

Figure 5.2: A) Illustration of forefoot base of support (BOS) as defined by the minimal distance between the toes and the posterior edge of the platform during a successful maximal toe reach. B) Maximal forward reaching distance on floor, defined as the distance between the anterior edge of the base of support at an upright stance and the most anterior position of fingertips at the floor level. C) Mean (SD) values of forefoot BOS and D) maximal forward reaching distance on floor. Test results indicated by * (P < .01) …………………………………………………97

Figure 5.3: A) Schematic of virtual time-to-contact (VTC) calculation to instantaneous geometric boundary of the base of support (BOS). B) Exemplary VTC data while standing on either a full or forefoot BOS…………………………………………………………101

Figure 5.4: Left) Mean (SD) VTC values. Right) Mean (SD) percentage of movement with an infinite VTC value in full and forefoot base of support (BOS) conditions……………………………………………103

Figure 5.5: Left) Maximum anterior or minimum posterior COP excursion (Mean [SD]) in full or forefoot base of support (BOS) conditions. Right) Mean (SD) COP root mean square (RMS) error demonstrating interaction effects between age and base of support condition……………………………………………………105

Figure 5.6: Left) Mean (SD) movement times. Right) Number of submovements in full and forefoot base of support (BOS) conditions……………………………………………………………105

Figure 5.7: Correlation between COP submovements and movement time during (Left) downward reach and (Right) upward recovery phases……………………………………………………………..106

Figure 6.1: Illustration of sagitally-symmetric double-inverted pendulum with 2 degrees of freedom consisting of active ankle and hip joint torques……………………………………………………………..125

Figure 6.2: Schematic of feedforward control model used to describe the arrest of downward reaching movements………………………………126

Figure 6.3: Effects of age and SCK difficulty on maximal initial COM velocities. Data showed a 12% reduction in maximal, initial COM velocities for healthy older females and a 21% reduction in older women with SCK difficulty, when compared to healthy young women………………………………………………..128

Figure 6.4: Effect of base of support length (FULL vs LIMITED [LIM]) on simulated COM phase plot of healthy young and older females and impaired older females with SCK difficulty………………128

Figure 6.5: Sensitivity analysis demonstrates a greater impact of ankle and hip rate of torque development (RTD) on the maximal initial COM velocity when values are varied ± 30% from values of 300 N & 300 Nm, respectively, for peak torque and rate of torque development…………………………………………………..129

Figure 6.6: Effects of decreased ankle (RTD) rate of torque development on maximal initial COM velocities……………………………………..130

LIST OF TABLES

Table 2.1: Characteristics Associated With Difficulty in SCK (n=184) ……………25

Table 2.2: Characteristics Independently Associated With SCK Difficulty (n=142) …………………………………………………………………..26

Table 4.1: Mean ± SD participant characteristics…………………………………...62

Table 4.2: Mean ± SD participant baseline measurements…………………………71

Table 4.3: Mean and standard deviation (SD) of raw dependent variables for healthy young (Y) and older (O) women, organized by task condition…………………………………………………………………73

Table 4.4: Effect of age, hand support, movement amplitude, and target size on primary submovement (PSM) characteristics……………………74

Table 4.5: Effect of age, hand support condition, movement amplitude (small- amplitude versus large-amplitude movement), target size, and their interactions with age on overall COP movement characteristics……………………………………………………………79

Table 5.1: Mean ± SD subject characteristics………………………………………95

Table 5.2: Mean ± SD subject capacities and performance………………………..102

Table 6.1: Average subject body segment parameter data…………………………124

Table 6.2: Nominal values of musculoskeletal parameters………………………...127

LIST OF APPENDICES

Appendix A: Decreased Muscle Strength Relates to Self-Reported Stooping, Crouching, or Kneeling Difficulty in Older Adults……………………159

Appendix B: Changes in Distal Postural Control Accuracy Near the limits of the Base of Support……………………………………………………………182

Appendix C: Downward Reach and Pick-Up Strategies in Older Females With Self-Reported Difficulty…………………………………………….193

ABSTRACT

Biomechanics of Leaning and Downward Reaching Tasks in Young and Older Women

by

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Stooping crouching or kneeling (SCK) difficulty is prevalent among older adults yet few studies have explored the mechanisms underlying downward reaching and pick-up difficulty. During targeted movements tradeoffs are expected between the speed and accuracy of center of pressure (COP) movements as balance is maintained. Thus, this research focused on how age-related changes in COP control strategies affect the performance of tasks with a large range of truncal motion and momenta. It was hypothesized that while performing leaning and downward reaching movements, older women, compared to younger women, would exhibit slower but more frequent COP submovements in order to accomplish the task and regain the upright posture.

First, we investigated the limiting factors in downward reach and pick-up movements. Using an age-adjusted proportional odds model, increased SCK difficulty was found to be independently associated with balance confidence, leg joint limitations, and knee extension strength.

Secondly, we explored the age-related changes in COP control in healthy women. Despite being 27% slower, older women rely on nearly twice as many submovements to maintain a similar level of endpoint accuracy in volitional COP movements, particularly when moving posteriorly. Furthermore, older women used slower primary submovements that more often undershot their target, in comparison to young women, particularly as movement amplitude increased.

Lastly, healthy older women were found to lose their balance more often than young women in downward reaching tasks, but rely on similar COP control strategies when successful. Modeling results suggest that a simple forward dynamic model that accounts for changes in musculoskeletal factors may distinguish between healthy young and healthy older women with and without SCK difficulty.

We conclude that biomechanical factors can distinguish between older women with and without SCK difficulty. Given the significance of the rate of torque development in arresting downward reaching movements, changes in COP control may be effective tools in evaluating early signs of physical impairment. Undershooting primary submovements and increased secondary submovements are indicative of an increasingly conservative strategy used by older adults near the limits of the base of support that may explain their slower speeds during whole body movements to maintain balance.

CHAPTER 1

INTRODUCTION

1.1. Epidemiology

Stooping, crouching, or kneeling (SCK) is a common daily activity, often needed when gardening, doing laundry, or putting groceries away in low cabinets. However, difficulty in downward reach and pick-up tasks is a prevalent phenomenon among older adults in modern society. Based on the Established Populations for Epidemiologic Studies of the Elderly (EPESE) baseline data, more than half (52%) of older adults over the age of 65 years reported some or more difficulty stooping, crouching, or kneeling (Taylor et al., 1998). The prevalence of older adults with significant SCK difficulty (Liao et al., 2001, Whetstone et al., 2001) may even be underestimated given the percentage of older adults (39%) who restrict or avoid tasks that require stooping, crouching, or kneeling due to health or physical problems (Wolinsky et al., 2005). Severe limitations in downward reach and pick-up tasks, requiring SCK movements, increase the likelihood of difficulty in other functional tasks such as lifting or carrying, and prolonged standing (Long & Pavalko, 2004). Few data explain why older adults have SCK difficulty, even though this difficulty may significantly impact the overall mobility and independence of older adults.

1.2. Stooping, Crouching, or Kneeling and Fall Risk

Falls are among the most significant causes of mortality and serious injury in older adults over 65 years of age (Bell, 2000, Del Corso, 1994, Kingma, 2000, Riley, 1992, Rubenstein, 2002). Furthermore, specific groups, such as women are at higher risk for injurious falls (Norton et al., 1997, Tinetti et al., 1995). However, considering that among older adults, the oldest of the old (80+ years of age) constitutes the fastest growing segment (UN, 2000), and that the risk of injurious falling increases with age (Lord, 1993), falls are an increasingly significant problem in the world. Thus, fall prevention interventions and fall-risk assessment tools, will be increasingly important to reduce the number of impaired older adults in the world.

Difficulty in downward reach and pick-up tasks may be indicative of an increased risk of falling among older adults. Second to walking, activities that involve getting in and out of a bed or chair and stooping, bending or reaching have been found to be associated with 37% of nonsyncopal falls, with an incidence rate of 37% (Nevitt et al., 1991). Difficulty bending down to pick up an object from the floor has been associated with a higher risk of falls in community-dwelling older adults (O'Loughlin et al., 1993). Furthermore, older adults who regularly perform gardening report significantly better balance and mobility, as well as fewer falls (Chen & Janke, 2012). Picking up a slipper from the floor, along with standing on one leg, have been found to be one of the items that best distinguishes fallers from non-fallers (Chiu et al., 2003). The importance of stooping, crouching, or kneeling is evidenced by the use of functional and self-reported downward reach and pick-up tasks in clinical fall risk batteries such as the Activities-specific Balance Confidence Scale, Berg Balance Scale, and the Physical Performance Test (Berg et al., 1992, Reuben & Siu, 1990, Powell & Myers, 1995).

1.3. Stooping, Crouching, or Kneeling Movements

SCK movements require coordination of the whole body over a wide range of postures in order to prevent a loss of balance. Moving from an upright stance into crouching and kneeling involves significant ankle, knee, and hip range of motion, whereas stooping movements are characterized by a reduction of knee movement (Burgess-Limerick et al., 2001). As with standing up from a chair, large whole-body linear and angular momenta must be arrested when reaching down to the floor or recovering to an upright stance (Riley et al., 1991). SCK movements may constitute a significant challenge to the balance or strength capacities of older adults (Kingma et al., 2004, Puniello et al., 2000), particularly when stooping (Bazrgari et al., 2007). Given that the ability to limit excessive trunk flexion and velocity are important elements of balance recovery (Grabiner et al., 1993, Owings et al., 2001), trunk control might be of importance during downward reach and pick-up movements. During SCK movements, both balance and body configuration must be simultaneously coordinated to account for pending interactions with the environment (Toussaint et al., 1995, 1997). Thus, as the position where all resultant forces act on the body, the center of pressure (COP) can provide insight into the central nervous system's strategy to control the body's center of mass during SCK movements (Winter, 1990).

Considering the control of SCK movements as a closed loop process (**Figure 1.1**), musculoskeletal changes in strength or joint range of motion are not the only factors expected to limit SCK strategies (Edmond & Felson, 2003, Janssen et al., 2002, Puniello et al., 2000, 2001). Age-related changes in sensory function and low balance confidence may also affect postural performance in downward reaching movements (Carpenter et al., 2006). Furthermore, prior experience can significantly affect the postural control of older adults (Lamoth & van Heuvelen, 2011), but in this thesis we shall focus on body dynamics, as strength and mobility are crucial in numerous functional tasks.

1.4. Postural Control

Balance during upright stance requires constant adjustments of the location of the COP under the feet (Winter, 1990), particularly with increased age (Freitas et al., 2005, Prado et al., 2007, Prieto et al., 1996). Considering a person at an upright stance, the center of mass (COM), or the point in a body at which the whole mass is said to be concentrated, and the COP can be measured at a given instant of time (**Figure 1.2**). Looking further at a schematic of COM and COP changes in quiet stance (Winter, 1995), we can consider an analogy given by David Winter where the COM can be thought of a sheep that we need to keep contained within a certain area, while the COP is the sheep dog. Where, if the latter sees the former straying too far from where it should be, it has to round it up and push it back. Movements of the COP, in turn, could be achieved via corrective reactions consisting of ankle or hip strategies with the feet-in-place (Luchies, et al., 1994, Rogers, 1996). Balance during downward reach and pick-up movements may be achieved primarily by ankle or hip strategies up to the point that a loss of balance is eminent and a protective reaction is required. Stepping or grasping would then be an effective strategy to avoid a fall (Maki & McIlroy, 1997, McIlroy & Maki, 1993, Luchies et al., 1994, Rogers et al., 1996), and be indicative of a loss of balance.

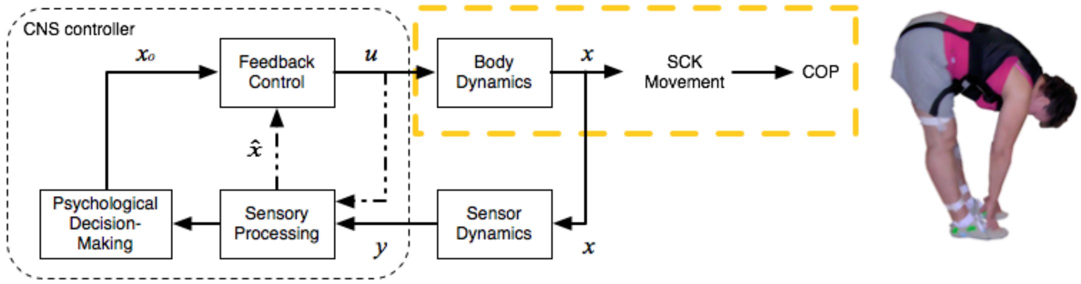

Figure 1.1: Conceptual postural control model of stooping, crouching, or kneeling (SCK) movements, considering feedback control and the center of pressure (COP) as system output. The dashed square box represents the focus of this thesis.

To prevent a loss of balance while leaning forward, the COP must be moved rapidly towards the toes, thereby keeping the COM from straying beyond the base of support. During rapid untargeted continuous COP movements, decreases in speed have been shown due to increased age and fall-risk (Tucker et al., 2008, Tucker et al., 2009). In contrast to Tucker et al. (2008, 2009), we propose to evaluate age-related changes in rapid discrete COP movements directed to a target. Considering the inevitable tradeoffs that occur when the COP is moved rapidly (Duarte & Freitas, 2005), limitations on the accuracy or maximal speed of compensatory movements while bending down to the floor would be expected as we age, due to neuromuscular changes (Galganski et al., 1993, Roos et al., 1997, Tracy & Enoka, 2002). Older adults demonstrate a decrease in the number of postural changes during standing in comparison to young, consistent with impairment in the somatosensory system (Lafond et al., 2008, Prado et al., 2010). Thus, the evaluation of COP control strategies should provide insight into the mechanisms underlying age-related and functional impairment-related changes in the postural control of downward reach and pick-up movements.

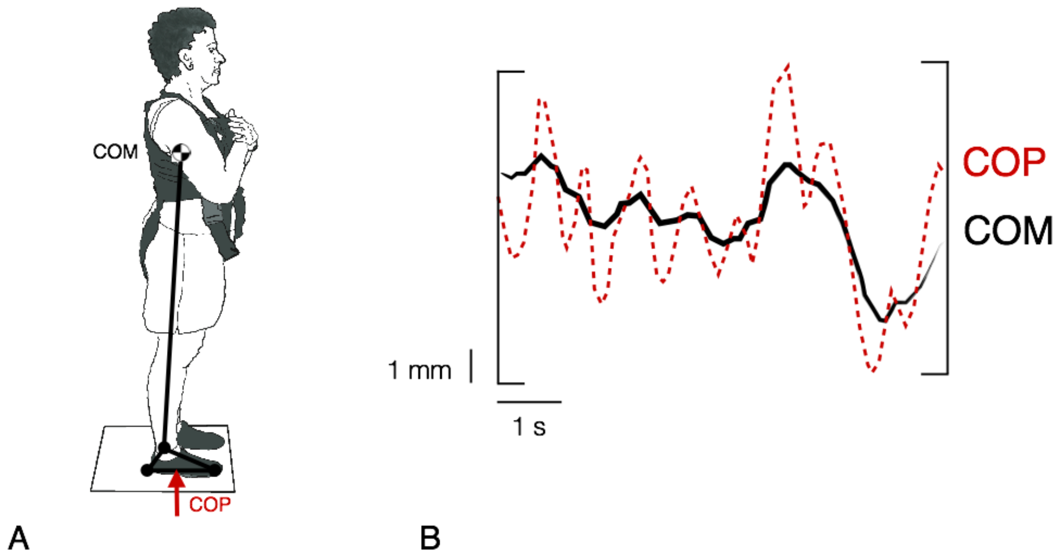

Figure 1.2: A) Inverted pendulum model depicting center of mass (COM) and center of pressure (COP) estimation. B) Schematic of COP and COM changes in a quiet stance (Winter, 1995).

1.5. Speed-Accuracy Tradeoffs in COP Movements and Submovements

Rapid COP movements are often required to avoid falls. However, movement accuracy must often be sacrificed during fast movements. Speed-accuracy tradeoffs have been observed in a wide range of tasks (Fitts, 1954) and can be characterized by the following relationship:

$$MT = a + b \cdot \log_2 (2MA/TS)$$

where *MT* is the movement time, *MA* is the movement amplitude, *TS* is the target size, and *a* and *b* are experimentally derived constants. When extending these principles to center of pressure movements, we find that additional factors such as the body's moment of inertia, musculoskeletal properties, or intrinsic variability while standing may limit the application of Fitts' Law (Danion et al, 1999, Duarte & Freitas, 2005). To account for the changes seen in postural sway along the length of the foot (Duarte & Zatsiorsky, 2002) and help explain the differences seen in continuous whole body movements, an alternate formulation of speed-accuracy tradeoffs is introduced (Meyer et al., 1982, Schmidt et al., 1979):

$$W_e = a + b \cdot S$$

where S is the mean movement speed, W_e is the effective target size, and a and b are experimentally derived constants. The stochastic optimized submovement model represents movements as a compromise between primary and secondary submovements that are dependent on the underlying motor control noise and spatial variability of submovement endpoints (Meyer et al., 1988). Thus, the optimized submovement model is useful for furthering the understanding of age-related changes, as it allows us to make predictions about the underlying submovement structure in the first phase of downward reach and pick-up movements, the leaning.

1.6. Biomechanical Modeling

In modeling reaching tasks from an upright stance, a simplification to motor control is suggested by the use of a common kinematic waveform across reaching movements to varied target positions and speeds (Thomas et al., 2003, 2005). In contrast to the symmetrical bell-shaped velocity curve found in the planar arm movements of young adults, older adults have been found to be temporally asymmetrical, spending more time decelerating (Cooke et al., 1989). Another commonly used motor control strategy is to control the center of mass position, however, evidence suggests that angular motions about the hip and ankle are not to keep center of mass at a constant position, but rather to minimize acceleration of the center of mass (Aramaki et al., 2001). The selection of any one criterion to quantify the control of balance while reaching down to the floor may be unfeasible given the numerous strategies available to achieve this task. Thus, one strategy to characterize the control of balance is to examine the maximal feasible whole body center of mass momenta that can be arrested safely, without exceeding the limits of stability, as defined by the boundaries of the base of support.

Previous biomechanical modeling studies have examined the feasible ranges in velocity, acceleration, or torque so as to maintain balance in an upright stance (Gordon, 1995, Iqbal & Pai, 2000, Kuo, 1995, Kuo & Zajac, 1993a, Kuo & Zajac, 1993b, Nashner, et al., 1989, Patton et al., 1999, Simoneau & Corbeil, 2005, Yang, et al., 1990, Zajac, et al., 1984). In these studies, the idea that a range of feasible movements can be influenced by the interaction between physical and task constraints was introduced. Based on an adaptation of these methods, we can examine how physical constraints in peak torque and rate of torque development can affect the maximal initial horizontal COM velocity that can be arrested without losing balance. The use of physical constraints to estimate the maximal horizontal COM momentum that can be generated has several practical implications. First, the biomechanical model assesses the relative importance of musculoskeletal factors on the ability to arrest horizontal linear momentum, which is a critical factor in the control of balance in common transfer tasks. In addition, the use of a double inverted pendulum allows for the comparison between ankle and hip joint parameters to evaluate the significance of distal versus proximal joints. Finally, the current literature has limited information on the characteristics of older adults with stooping, crouching, or kneeling difficulty, and thus the use of musculoskeletal parameters allows us to evaluate if any differences emerge due to healthy aging or stooping, crouching, or kneeling impairment.

1.7. Summary

The most important factors contributing to downward reach and pick-up difficulty in older women include limitations in lower extremity function. Motor control of both proximal and distal joints is expected to be important for balancing during downward reach and pick-up tasks. COP movement characteristics that reflect a speed-accuracy tradeoff and include submovements can provide insight into the performance of leaning and downward reaching tasks. Given the known changes in rate of ankle torque development and strength, this thesis will focus on how age-related changes in COP control strategies affect the performance of a task with a large range of truncal motion and momenta, and test the following overarching hypothesis: While performing leaning and downward reaching movements towards a target, older women, compared to younger women, will exhibit slower but more frequent COP submovements in order to accomplish the task and regain the upright posture.

1.7.1. Working Hypotheses

1. Older adults with increasing self-reported stooping, crouching, or kneeling difficulty will have greater impairments in musculoskeletal function (Chapter 2).
2. In comparison to young adults, healthy older adults will exhibit a decrease in COP movement speed and an increase in number of submovements and ratio of peak-to-average COP velocity, particularly in posterior versus anterior movements (Chapter 3).

3. In comparison to young women, older women will display a disproportionate decrease of speed and accuracy (e.g., increased number of submovements and increased incidence of COP undershooting) in their COP primary submovements, as movement amplitude increases (Chapter 4).
4. In comparison to young women, healthy older women will exhibit a higher incidence of losses of balance during a downward reach pick-up (Chapter 5).
5. Decreasing the length of the base of support, from the whole foot to just the forefoot, will lead to a disproportionate decrease in COP control in older women when compared to young, as evaluated by decreased virtual time-to-contact and increased COP excursion, postural sway, movement time and number of submovements (Chapter 5).
6. A simple biomechanical model can distinguish between healthy young, healthy older women, and older women with SCK difficulty (Chapter 6).
7. Rate of torque development will be the most significant factor in successfully arresting horizontal linear momentum during a downward reach (Chapter 6).

1.7.2. Overview

This dissertation consists of nine chapters and three appendices. Overall, this dissertation incorporates the use of experimental and analytical methods in a broad spectrum of participants, consisting of young and older adults with and without functional impairments. In Chapter 2, a regression model is used to identify characteristics independently associated with stooping, crouching, or kneeling difficulty amongst a broad spectrum of measures in a large sample of older adults. Appendix A builds upon the findings of Chapter 2 and correlates stooping, crouching, or kneeling difficulty with normalized strength and clinical balance measures. Given the association between lower extremity strength capacity and functional impairment among older adults, Chapters 3 & 4 explore the effects of healthy aging on distal postural control, as it may provide an outcome measure of lower extremity function. In Chapter 3 healthy young and older participants are asked to lean their entire bodies about their ankles, so as to move their COP as fast and as accurate as possible between a pair of targets well within their limits of stability. Using accuracy-constraints, the effect of age and movement direction on the speed of volitional COP movements is evaluated. Chapter 4 builds upon the findings of Chapter 3, by using large-amplitude targeted COP movements to investigate the role of the primary submovement in age-related changes in COP control strategies. Furthermore, Appendix B validates the use of COP control measures in young adults and explores the role that target position and foot placement have on distal postural control performance. Chapter 5 introduces downward reach and pick-up tasks requiring maximal exertions in balance control or body configuration control. Chapter 6 builds upon Chapters 2 & 5 by exploring the role of musculoskeletal parameters in the postural control of downward reach and pick-up movements by using a forward dynamic model of a planar double-inverted pendulum. General discussion of all these chapters is presented in Chapter 7, Chapter 8 presents the conclusions, while Chapter 9 outlines several recommendations for future research.

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CHAPTER 2

CHARACTERISTICS OF OLDER ADULTS WITH SELF-REPORTED STOOPING, CROUCHING, OR KNEELING DIFFICULTY

2.1. Abstract

Stooping, crouching, and kneeling (SCK) are fundamental components of daily living tasks, and nearly a quarter of older adults report a lot of difficulty or inability to perform these movements. This study examined characteristics associated with SCK difficulty to explore underlying mechanisms and remediation strategies. One hundred eighty-four older adults with no, low, or high SCK difficulty underwent a comprehensive laboratory visit at the University of Michigan. Twenty-one percent of participants (n = 39) reported a lot of difficulty or inability to stoop, crouch, or kneel. Characteristics independently associated with increasing SCK difficulty were self-reported leg joint limitations, (odds ratio [OR] = 3.84; 95% confidence interval [CI], 1.64-9.01), Activities-specific Balance Confidence Scale score (OR = 0.97; 95% CI, 0.95–0.99), and knee extension strength (OR = 0.72; 95% CI, 0.55–0.94). Increasing SCK difficulty is associated with balance confidence as well as leg limitations. Remediation of SCK

difficulty will likely require a program that encompasses both behavioral and physical issues.

2.2. Introduction

Based on the Established Populations for Epidemiologic Studies of the Elderly (EPESE) baseline data, almost one quarter of community-living older adults (24%) have a lot of difficulty or are unable to stoop, crouch, or kneel (Taylor et al., 1997). Stooping, crouching, or kneeling (SCK) movements are an integral component of many daily living tasks, including picking up an object from the floor, reaching to low-lying shelves, and gardening. Limitations in stooping, crouching, or kneeling are associated with the increased likelihood of limitations in other lower-body functional tasks such as lifting and prolonged standing (Long & Pavalko, 2004) and are associated with fall risk (O'Loughlin et al., 1993). Although SCK difficulty may significantly impact the overall mobility and independence of older adults, few data explore the mechanisms underlying SCK difficulty.

SCK movements require coordination of the whole body over a wide range of postures while maintaining balance. SCK movements in older adults may be limited by obesity as well as musculoskeletal impairments such as decreased lower extremity strength, and pain or stiffness-induced leg joint limitations (Edmond & Felson, 2003, Han et al., 1998, Janssen et al., 2002). These movements may also be more difficult for older adults with decreased sensation and who have low balance confidence (i.e., confidence in the ability to maintain balance and avoid a fall) as both of these factors are associated with fall risk (Carpenter et al., 2006). Tasks requiring stooping, crouching or kneeling can constitute a substantial postural threat, particularly for those who have physical impairments.

Although SCK difficulty is prevalent and affects daily function, factors contributing to SCK difficulty have not been studied in a comprehensive manner. The goal of this cross-sectional study is to identify these factors and thus inform specific interventions that might impact SCK difficulty. We hypothesized that, compared to those without SCK difficulty, those with increased SCK difficulty would have greater impairments in musculoskeletal, sensory, and balance function, as well as in balance confidence.

2.3. Methods

2.3.1. Subjects

The present study utilized baseline data from older adults enrolled in a 10-week balance training study (n=184) (Nnodim et al., 2006). In that study, participants were recruited at local senior housing facilities and senior centers in Southeastern Michigan. Inclusion criteria were: (a) age 65 or greater; (b) at least mild balance impairment, as defined by a unipedal stance time (UST) of < 25 seconds or more than one error during a tandem walk; and (c) ability to stand and take at least one step unsupported by a device or person. Potential participants were screened by a nurse practitioner to exclude those with significant cognitive impairment (< 24 on Mini-Mental State Examination [MMSE]), unstable medical conditions, or severe pain upon weight bearing precluding participation. For the present study, enrollees were categorized based on their response to a single question on the EPESE questionnaire (Smith et al., 1990) in which they rated their ability to stoop, crouch, or kneel according to a five point difficulty scale: no difficulty (n=45), a little (n=50), some (n=50), a lot (n=22), or unable to do (n=17). For analysis, participants were categorized into three groups: (i) no SCK difficulty (if they reported no difficulty), (ii) low SCK difficulty (if they reported a little, or some difficulty), or (iii) high SCK difficulty (if they reported having a lot of difficulty or were unable to perform SCK tasks).

2.3.2. Data Collection

Factors contributing to SCK difficulty were determined from general descriptors (such as age, gender, and obesity), and both self-reported health and performance-based physical function. Obesity was operationalized as a Body Mass Index (BMI) > 30 kg/m^2. Self-report (interview-based) measures were taken from the medical screening administered by a nurse practitioner. The total number of chronic medical conditions was ascertained by asking participants if they had a previous history of osteoarthritis, rheumatoid arthritis, osteoporosis, myocardial infarction, stroke, joint replacement, Parkinson's Disease, or peripheral neuropathy. The presence of dizziness was determined by asking participants if they had a current episode of lightheadedness or vertigo, which would affect their balance, and the total number of prescribed medications was recorded from their medical screening. Depression was assessed by the short version of the Geriatric Depression Scale (GDS, range 0-15) and cognitive function was assessed by the Folstein MMSE (range 0-30) (Folstein et al., 1975). Self-reported balance confidence was determined by the Activities-Specific Balance Confidence Scale (ABC Scale; range 0-100) (Powell & Myers, 1995), and self-reported joint movement limitations were determined by the report of pain or stiffness in the back or leg (i.e., hip, knee, or ankle). Physical function measures included maximum isokinetic knee extension strength (hereafter referred to as knee extension strength), using an isokinetic dynamometer, at a velocity of 120 degrees per second (Nnodim et al., 2006), and two common functional tests of balance and mobility: the UST (Vellas et al., 1997) and the Timed Up and Go (TUG) test (Podsiadlo & Richardson, 1991). To normalize for body size, knee extension peak torque values were expressed as a percentage of the product of body weight (Newtons) and body height (meters). Participants who were unable to perform the UST (n = 63/184, 34%) were coded as having a UST of 0 seconds. Impaired position sense was reported if participants had an abnormal position sense in the big toe of either one or both legs, as defined by having one or more repositioning errors within five trials (Van den Bosch et al., 1995). Fall-related variables were obtained by asking participants whether they had fallen within the past year, (if yes) the number of falls incurred, and if medical treatment was sought for their injuries (i.e., injurious falls).

2.3.3. Data Analysis

All analyses were carried out in SPSS (version 15.0; SPSS Inc., Chicago, IL). The Kolmogorov-Smirnov test was used to test for normality when variables were continuous. Group differences were evaluated in continuous variables by either using a one-way analysis of variance when normally distributed (e.g., age and knee extension strength) or the nonparametric Kruskall-Wallis rank test when abnormally distributed. To determine group differences in dichotomous variables, the chi-square test for independence was performed. To examine statistical significance, both SCK Difficulty groups (e.g., low and high difficulty) were individually compared to the no SCK Difficulty group using Hochberg's step-up method. To identify characteristics independently associated with self-reported SCK difficulty among older adults, an age-adjusted proportional odds model was used with a forward selection procedure, so as to capture the ordinal nature of self-reported SCK difficulty. The model included covariates that were statistically significant at $p \leq 0.1$ and included only participants with complete data on all measures (n = 142/184, 77%). In a preliminary analysis, we found that none of the predictor variables were highly correlated (>0.5), thus, indicating a low chance of collinearity in the model.

To determine whether the inclusion of the 42 participants with missing data biased the primary analysis of characteristics associated with SCK difficulty, a secondary analysis was conducted using *t* tests and chi-square tests, comparing participants included and excluded in the proportional odds model. We found no significant increases in the percentage of missing data in the participants who reported a lot of difficulty or were unable to carry out SCK tasks (20%), in comparison to the mean for the entire sample (23%). No statistically significant differences were found between participants included and those excluded from the proportional odds model in all but one variable: age. Participants who were excluded from the final proportional odds model were older (80.3 ± 7.2 vs 77.5 ± 6.9, P=.021). Given a lack of systematic differences between groups, results from the reduced sample are presented in this article.

2.4. Results

Participants (n=184) had a mean age ± standard deviation of 78.5 ± 6.9 (range 65-92) and were predominantly female (78.3 %). Twenty-one percent of all study participants reported a lot of SCK difficulty or were unable to stoop, crouch, or kneel; 23% were obese; and 32% had two or more chronic conditions.

Characteristics of participants with no SCK difficulty, low SCK difficulty, and high SCK difficulty are shown in **Table 2.1**. Comparison of the three groups revealed differences in gender, the number of chronic medical conditions, dizziness, GDS-short version score, ABC Scale score, self-reported back and leg joint limitations, knee extension strength, UST, TUG time, and incidence of more than two falls in the past year. There was a trend toward a group difference in the number of medications. There were no significant SCK group differences in age, obesity, MMSE score, impaired position sense, and incidence of injurious falls. Hochberg's step-up post hoc analyses revealed that, in comparison to participants with no SCK difficulty, participants with low or high SCK difficulty had significantly lower ABC Scale score, more self-reported leg joint limitation, and decreased knee extension strength.

Based on the findings from the primary analyses, the following predictor variables were included in the age-adjusted proportional odds model: gender, chronic medical conditions, dizziness, number of medications, GDS-short version, ABC Scale score, self-reported back and leg joint limitations, knee extension strength, UST, TUG time and more than two falls in the past year. While controlling for age, the proportional odds model (n=142) showed that characteristics independently associated with increasing SCK difficulty were self-reported leg joint limitations, (odds ratio [OR] = 3.84, 95% confidence interval [CI], 1.64 – 9.01), ABC Scale score (OR = 0.97, 95% CI = 0.95 – 0.99), and knee extension strength (OR = 0.72, 95% CI = 0.55 – 0.94, **Table 2.2**).

Table 2.1: Characteristics Associated With Difficulty in SCK (n=184).

Characteristic	No SCK Difficulty		Low SCK Difficulty*		High SCK Difficulty*		
	N	Mean ± SD or %	N	Mean ± SD or %	N	Mean ± SD or %	p Value
Participant Description							
Age, y	45	77.9 ± 6.8	100	78.5 ± 7.1	39	78.5 ± 7.0	0.77
Gender, female (%)	45	80	100	72	39	92.3	0.032
Obesity (%)	45	22.2	100	20	39	33.3	0.243
Self-Reported Health							
Chronic medical conditions	44	0.7 ± 0.8	99	1.3 ± 1.0	39	1.3 ± 0.9	0.008
Dizziness (%)	44	34.1	99	62.6	38	60.5	0.005
No. of Medications	44	3.5 ± 2.8	99	4.2 ± 2.6	39	4.5 ± 2.5	0.052
GDS-short version	40	0.7 ± 1.6	84	1.5 ± 2.1	36	1.9 ± 2.2	0.024
MMSE	31	27.3 ± 1.3	71	27.8 ± 1.3	26	27.5 ± 1.4	0.107
Activities-Specific Balance Confidence Scale	42	82.9 ± 14.2	97	72.5 ± 18.7†	38	65.0 ± 17.9†	< .001
Back joint limitation (%)	45	4.4	100	13	39	28.2	0.007
Leg joint limitation (%)	44	2.3	100	24	39	46.2†	< .001
Physical Function							
Knee Extension Strength	40	3.9 ± 1.5	79	3.1 ± 1.3	30	2.5 ± 1.3†	< .001
UST, s	45	7.7 ± 7.7	100	5.3 ± 6.3	39	3.6 ± 5.2	0.007
TUG time, s	45	11.8 ± 4.9	100	15.1 ± 8.5	38	15.6 ± 6.6	< .001
Impaired Position Sense (%)	45	22.2	99	20.2	37	13.5	0.579
Fall-Related							
≥ 2 Falls (%)	43	2.3	97	17.5	38	13.2	0.047
≥ 1 Injurious Falls (%)	44	22.7	99	37.4	39	21.4	0.189

Notes: Means and standard deviations (SD) are presented for continuous variables and proportions are presented for categorical variables. For dizziness, self-reported back and leg joint limitations, knee extension strength and position sense, see Methods for exact calculation. For determining group differences, continuous measures were evaluated using either an analysis of variance (ANOVA) test or a Kruskall-Wallis test, and dichotomous measures were evaluated using a Pearson chi-square test.

* Hochberg's step-up post hoc comparisons between No SCK Difficulty and Low or High SCK Difficulty group, using a *t* test for continuous and a Pearson chi-square test for dichotomous variables.

†p ≤ .001 for Hochberg's step-up post hoc comparisons.

SCK = stooping, crouching, and kneeling; GDS = Geriatric Depression Scale; MMSE = Mini Mental Status Exam; UST = Unipedal Stance Time; TUG = Timed Up and Go.

Table 2.2: Characteristics Independently Associated With SCK Difficulty (n=142).

Characteristic	Adjusted Odds Ratio (95% CI)	p Value
Leg Joint Limitations	3.84 (1.64 – 9.01)	0.002
Activities-Specific Balance Confidence	0.97 (0.95 – 0.99)	0.002
Normalized Knee extension strength	0.72 (0.55 – 0.94)	0.016

Notes: Results from age-adjusted Proportional Odds model using a forward selection procedure. SCK = stooping, crouching, and kneeling; CI = confidence interval.

The results from the proportional odds model demonstrated that participants who report leg joint limitations have 3.84 times the odds of reporting increased SCK difficulty, and that a unit decrease in the ABC Scale score and knee extension strength (i.e., 1.15 Nm for the average participant in this study) led to a 3% and 28% decrease in the odds of reporting increased SCK difficulty, respectively.

2.5. Discussion

To the authors' knowledge, no studies have comprehensively evaluated the mechanisms underlying self-reported SCK difficulty, despite the high prevalence of SCK difficulty in older adults. We hypothesized that, compared to those without SCK difficulty, those with increased SCK difficulty would have greater impairments in musculoskeletal, sensory, balance function, as well as balance confidence. Results from the primary analyses are mostly in agreement with our hypothesis. However, obesity and impaired position sense did not differ between the groups.

Self-reported leg joint limitations, ABC Scale score, and knee extension strength were found to be independently associated with increased self-reported SCK difficulty in the age-adjusted proportional odds model. Most of these factors have been noted elsewhere to be associated with SCK difficulty as well as limitations in physical function. Leg joint limitations in older adults, as a result of knee and hip osteoarthritis, have been associated with greater limitations in physical function, including bending, kneeling, and stooping (Gelber et al., 2000). Balance confidence is strongly associated with self-reported physical function (Li et al., 2001) and physical performance (e.g., gait speed, timed hand function) measures (Tinetti et al., 1994). Reduced knee extension strength has been associated with limited function in daily activities such as rising from a chair or walking (Buchner et al., 1996, Scarborough et al., 1999). More specifically, the generation of both increased momentum and stable and controlled movements while rising from a chair may be limited by knee extension strength (Scarborough et al., 1999).

These findings may have implications for refining clinical interventions for older adults with SCK difficulty. Physical capabilities, such as knee extension strength, are often a focus of rehabilitation treatment to help people perform daily living tasks. This study suggests that interventions that address leg joint limitations and balance confidence may be especially important. Leg joint limitations can be addressed by teaching task-specific strategies such as using support surfaces for assistance in SCK movements or customized whole-body movement strategies. Practice of task-specific strategies is an important component of improving self-efficacy and thus could also positively impact low balance confidence (Bandura, 1986). Analogously, older adult congregate housing residents with observed performance difficulties in bath transfers were also more likely to have leg range-of-motion limitations and low balance confidence (Murphy et al., 2006). Finally, given the shared characteristics between older adults with SCK difficulty and older adults with a history of falls (including range-of-motion limitations in the back, and lower extremity strength) (Hurley et al., 1998, Tinetti et al., 1986, Tinetti et al., 2000), reducing SCK difficulty may also reduce fall risk. Thus, the present study suggests that increasing SCK difficulty is associated with behavioral as well as physical limitations, and that interventions to reduce SCK difficulty will require a broader program that goes beyond enhancing physical function, to include behavioral issues and strategy training (Murphy et al., 2006). We believe that interventions should be designed specifically for SCK difficulty, just as we designed interventions for rising from a bed or chair or from the floor (Alexander et al., 2001, Hofmeyer et al., 2002). SCK are key functional tasks that may be amenable to training. SCK is a very high level, complex, balance task that requires sufficient joint motion and strength, similar to sit-to-stand tasks.

2.5.1. Study limitations

Participants were predominantly female, did not have cognitive impairments, were selected for having functional balance impairments, and did not have objective measures of active joint ranges of motion. Because of the selection criteria for the balance training program, the fittest older adults who were participating in a regular exercise program and the frailest older adults with severe lower extremity pain or assistive device dependency were excluded. Despite limitations in generalizability, the recruited sample represents persons who are an ideal target group for strategy training, as they are at fall risk and have little cognitive impairment that would preclude learning and using new strategies. Future research should examine older adults with a broader range of physical function.

2.6. Conclusion

Increasing SCK difficulty is associated with balance confidence as well as leg limitations. Remediation of SCK difficulty will likely require a program that encompasses both behavioral and physical issues.

2.7. Significance

The importance of SCK difficulty is already evident in batteries of mobility and balance performance such as the Physical Performance Test (Reuben & Siu, 1990) and the Berg Balance Scale (Bogle-Thorbahn & Newton, 1996), which include bending over to pick up objects from the ground. Nevertheless, an objective measure of difficulty in bending over to pick up objects is still lacking, as these batteries consider only time or ability. A biomechanical assessment of stability while bending over to pick up objects may elucidate the mechanism underlying increased fall risk in older adults with SCK difficulty.

This chapter provided a broad examination of characteristics associated with SCK difficulty among older adults. As one of the first studies to go beyond the examination of the role of specific deficits such as muscle loss or back pain or stiffness on self-reported functional limitations, this research provides potential mechanisms, such as lower extremity strength and range of motion limitations, that may underlie the difficulty in downward reaching and pick-up movements.

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CHAPTER 3

THE EFFECT OF AGE, MOVEMENT DIRECTION, AND TARGET SIZE ON THE MAXIMUM SPEED OF TARGETED COP MOVEMENTS IN HEALTHY WOMEN

3.1. Abstract

Rapid center of pressure (COP) movements are often required to avoid falls. Little is known about the effect of age on rapid and accurate volitional COP movements. We hypothesized that COP movements to a target would be slower and exhibit more submovements in older versus younger adults, particularly in posterior versus anterior movements. Healthy older (n = 12, mean age = 76 years) and young women (n = 13, mean age = 23 years) performed anterior and posterior lean movements while standing on a force plate, and were instructed to move their COP ‘as fast and as accurately as possible’ using visual feedback. The results show that rapid posterior COP movements were slower and had an increased number of submovements and ratio of peak-to-average velocity, in comparison to anterior movements ($P < .005$). Moreover, older compared to younger adults were 27% slower and utilized nearly twice as many compensatory submovements ($P < .005$), particularly when moving posteriorly ($P < .05$). Older women also had higher ratios of peak-to-average COP velocity than young ($P < .05$). Thus,

despite moving more slowly, older women needed to take more frequent submovements to maintain COP accuracy, particularly posteriorly, thereby providing evidence of a compensatory strategy that may be used for preventing backward falls.

3.2. Introduction

Rapid center of pressure (COP) movements are often required to avoid falls. Balance during upright stance requires constant adjustments of the location of the COP under the feet (Winter, 1990), particularly with increased age (Freitas et al., 2005, Prado et al., 2007, Prieto et al., 1996). COP movements during upright stance also slow with increased age and fall-risk (Tucker et al., 2008, Tucker et al., 2009). The ability to recover from perturbations in specific directions is of significance, as the direction of a fall can be a significant predictor of the subsequent injury (Palvanen et al., 2000). Synchronous real-time visual feedback of COP movements has been used for patient rehabilitation as well as to assess the effect of spatial accuracy constraints on COP movement time (Danion et al., 1999, Duarte & Freitas, 2005, Hamman et al., 1992, Shumway-Cook et al., 1988). However, to our knowledge, the effect of age and movement direction on spatial accuracy-constrained volitional COP movements has not been well explored.

Purposeful COP movements in one direction to a given spatial target region under the foot can be termed 'discrete' movements. These differ from the continuous, often multi-directional COP movements normally used to maintain upright balance. It is possible that discrete COP movements, which can be observed during certain phases of gait initiation, and during turning and reaching (Mbourou et al., 2003, Meinhart-Shibata et al., 2005, Row & Cavanagh, 2007), are more challenging to postural control than continuous movements because of the spatiotemporal constraints, particularly in older adults. Community-dwelling older adults have demonstrated an increased risk of injury in falls when turning around or reaching (Nevitt et al., 1991). Furthermore, previous studies have shown age-associated delays in gait initiation, which is commonly associated with falls (Henriksson & Hirschfeld, 2005). The need for rapid, accurate, and discrete COP movement arises when given a limited base of support or limited time to initiate a postural response, as has been observed among older adults (Inglin & Woollacott, 1988, King et al., 1994).

The overall goal of this study was to determine the effect of age on discrete and rapid accuracy-constrained COP movements. Across a wide variety of tasks, Fitts' law has successfully predicted a tradeoff between accuracy and speed (for example, Fitts, 1954, Plamondon & Alimi, 1997). However, based on previous studies of whole-body COP movements (Danion et al., 1999, Duarte & Freitas, 2005), performance in these complex tasks are better predicted by the following equation:

$$W_e = a + b \cdot S$$

where S is the mean movement speed, W_e is the effective target size, and a and b are experimentally derived constants (Meyer et al., 1988, Schmidt et al., 1979). Mean movement speed can be derived from the mean movement amplitude and mean movement time, whereas the effective target size is a measure of the dispersion of the COP endpoint, defined by a multiple of the standard deviation (SD) of the COP endpoint position (i.e., $4 \cdot SD_{COP\text{-}ENDPOINT}$). Thus, movement speed is a key element of the present study as spatiotemporal parameters are constrained.

Age-related changes in COP control strategies are expected. Discrete movements may stress older versus young adult postural control disproportionately more than continuous movements. A possible mechanism is that discrete COP movements are apt to rely on sensory feedback for control (Smits-Engelsman et al., 2002), and thus may be susceptible to decreases in foot plantar sensation and optic flow threshold sensitivity due to age (Inglis et al., 2002, McKeon & Hertel, 2007, Wade et al., 1995). COP movement time and corrective COP movements should increase in older versus young adults, given that older adults exhibit increased movement times and corrective movements when performing an accuracy-constrained task in the upper extremity (Ketcham et al., 2002, Pratt et al., 1994), which is consistent with observations of declined motor coordination with increased age (Morgan et al., 1994). The ratio of peak-to-average velocity is an indicator of energy efficiency (Flash & Hogan, 1985, Hogan et al., 1987, Nelson, 1983). As age-related changes have been observed in the peak-to-average velocity of targeted upper extremity movements (Meulenbroek et al., 1998), increases in muscle force variability and antagonist muscle coactivation in older adults (Galganski et al., 1993, Tracy & Enoka, 2002) suggest they could have an increase in the ratio of COP peak-to-average velocity when compared to younger women. Moreover, given the greater limitations in limits of support in posterior movements (King et al., 1994), one would expect that discrete posterior motions would differ from anterior motions, again, disproportionately with age.

In this study, the effects of different target sizes and movement direction were assessed separately to explore their relative interaction with age. We hypothesized that in comparison to young adults, older adults would exhibit a decrease in COP movement speed and an increase in the number of compensatory submovements and ratio of peak-to-average COP velocity, particularly in posterior versus anterior movements. Thus, this study furthers the understanding of how healthy aging affects the control strategies of discrete, rapid, and accurate volitional COP movements.

3.3. Methods

3.3.1. Participants

Thirteen healthy young (mean ± SD age 23 ± 3 years, height 164 ± 6 cm, weight 63 ± 11 kg) and twelve healthy older (age 76 ± 6 years, height 159 ± 5 cm, weight 63 ± 11 kg) women participated in this study. Both healthy young and older women had similar sized feet and body mass index (foot length 26 ± 1 cm vs. 25 ± 3 cm, body mass index 23 ± 4 kg/m^2 vs. 25 ± 5 kg/m^2, respectively). Young participants were recruited from the University of Michigan campus and older participants were recruited from a database maintained by the University of Michigan Older Americans Independence Center Human Subjects and Assessment Core. All young women filled out a medical history questionnaire and all older women underwent a medical history screening and physical examination by a nurse practitioner, in order to exclude those with significant musculoskeletal or neurological findings. All participants signed a written informed consent form approved by the University of Michigan Medical School Institutional Review Board.

3.3.2. Set-up

COP data was collected while participants stood on a ground-level six-channel force plate (OR6-7-1000, AMTI, Watertown, MA). Three-dimensional ground reaction forces and moments were sampled at 100 Hz. A thin wooden platform mounted on top of the force plate was used to control foot placement (i.e., stance width and anterior edge of the base of support). Real-time anteroposterior COP location was recorded using a Certus Optotrak system and First Principles software (Northern Digital, Inc., Waterloo, Canada) and calculated using the standard equation $COP_x = -M_y/F_z$. Participants viewed a COP target on a computer monitor positioned directly in front of them. In order to control for the shoe-floor interface during testing, participants wore standardized canvas shoes with a thin rubber sole that varied in ½ size increments between U.S. women sizes 7 ½ to 11 ½. Targets of varying size (2, 3, and 4 cm) were centered at two locations. One was the location during a neutral standing posture (home target zone). The second location lay 6-cm anterior to the home target zone (anterior target zone) but well within the anterior limit of the functional base of support (FBOS), which is representative of the typical COP excursion in forward reaches in a parallel upright stance (Gillette & Abbas, 2003). A pair of horizontal lines defined the participant-specific anterior and home target zones that were used in all conditions.

3.3.3. Protocol

The COP movements under each participant were first measured in a neutral standing posture for 30 seconds in order to establish a reference point for COP target placement during testing. Participants were tested for the anteroposterior excursion of the COP during maximal sustained anterior and posterior leans in two separate 30-s trials, so as to define their FBOS. Participants then performed discrete accuracy-constrained COP movements to pseudo-randomized targets in 60-s trials on three separate laboratory test sessions. Each testing session consisted of a minimum of 6 practice trials that preceded 6 experimental trials of varying target size. In addition to submaximal trials, trials of maximal COP movement to the anterior and posterior limits of the functional base of support were used to reduce anticipation of experimental trials.

Participants were instructed to move their COP from their initial position to a target zone, after hearing a go signal consisting of a time-varying (1-3 s) auditory tone. Participants maintained their COP within the confines of an experimentally manipulated target using visual feedback from the computer monitor at a fixed magnification. Subsequent auditory cues were used to prompt a change in movement direction (**Figure 3.1**). Participants were instructed to lean their body and move their COP 'as fast and as accurately' as possible, while keeping their arms crossed. Subsequent movements continued until 60-s had elapsed. On average, 21 discrete COP movements were performed on target, per trial. The first anterior and posterior movement within a trial was not analyzed, to account for re-familiarization with the task.

3.3.4. Data Processing

Custom Matlab (v7.4, Natick, MA) data processing software routines were written to process the data. Raw force plate data were processed with a 4th order, zero-lag, low-pass Butterworth filter with a 10 Hz cutoff frequency. COP velocity and acceleration were calculated using a five-point finite difference derivative algorithm. Using an automated procedure, COP movements were extracted from each trial, using the trigger signal associated with the auditory cue to estimate the onset and offset of individual movements. COP velocity profiles were used to define the onset and offset of COP movements. The algorithm defined the onset of the COP movement by searching backward from the sample with the peak COP velocity and locating the first sample at which the velocity exceeded 10% of the maximum value (V_{max}), within the starting zone (Teasdale et al., 1993). The offset of COP movement was similarly found by moving forward from the sample with the peak COP velocity and identifying the first sample less than or equal to 10% of V_{max} within the target zone.

Dependent COP variables included movement speed, number of submovements, and ratio of peak-to-average velocity. In each trial, dependent COP variables were averaged across all anterior or posterior movements for use in further analysis. For each individual COP movement: COP movement speed was defined as the movement amplitude divided by the movement time (i.e., the elapsed time from the onset to the offset); submovements were defined as pairs of zero crossings in COP acceleration after V_{max}; and the ratio of peak-to-average COP velocity was defined as V_{max} divided by the average COP velocity. In addition, to verify the expected changes in accuracy-constraints, the effective target size was measured. As previously described, the effective target size was defined as the length that would accommodate 95% of all COP endpoints.

3.3.5. Statistical Analysis

Assumptions of normality were confirmed using the Kolmogorov-Smirnov test for each variable. Linear mixed-models, using a restricted maximum likelihood method, were used to examine the effect of age (young versus older), movement direction (anterior versus posterior), and target size (2 versus 3 versus 4 cm). Movement direction and target size were identified as repeated effects assuming a first-order autoregressive covariance structure. The relationship between movement speed and the number of submovements was evaluated using the Pearson correlation coefficient. To evaluate learning effects, a repeated-measures analysis of variance (RM-ANOVA) was used that controlled for age and target size on an aggregate of anterior and posterior COP movements. The reliability of COP measures over the three sessions was evaluated by using intraclass correlation coefficients (ICCs). Measures were aggregated for each session and the ICC equation for a 2-way random effects model of consistency was utilized in the current study. Considering the general guidelines offered in the literature, ICC values greater than 0.75 are assumed to correspond to excellent reliability (Fleiss, 1996). The Geisser-Greenhouse correction to the degrees of freedom was used to report univariate test results when violations to sphericity occurred in a RM-ANOVA. Post-hoc t-tests were carried out using Hochberg's step-up method to assess the most significant age and movement direction differences. $P < .05$ was considered statistically significant. All statistical analyses were carried out in SPSS 16.0 for Windows (SPSS Inc., Chicago, IL).

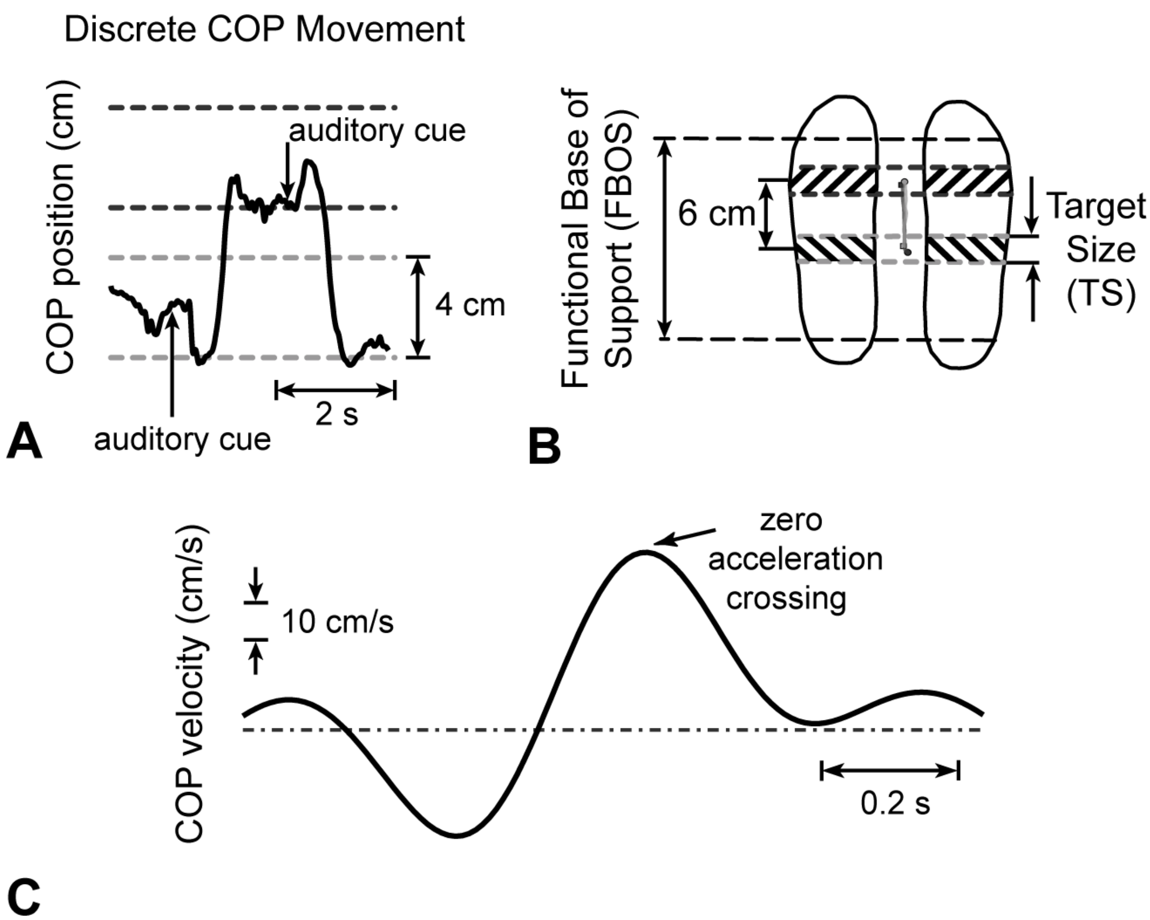

Figure 3.1: A) Illustration of discrete COP movement initiation using auditory cues. B) Exemplary COP movement during movements from a target at a neutral standing posture (dashed light gray lines, home target zone) to a target with a fixed movement amplitude of 6 cm, within the anterior edge of the functional base of support (dashed dark gray lines, anterior target zone) and back (solid light gray line). C) COP velocity trajectory illustrating the zero acceleration crossings used to define submovements. Submovements were defined as a pair of zero acceleration crossings (i.e., COP velocity peaks or throughs) between the movement's peak velocity and arrest of movement.

3.4. Results

All participants demonstrated no statistically significant changes in movement speed, number of submovements, and ratio of peak-to-average COP velocity ($P > .05$) across experimental sessions, based on a RM-ANOVA. The intraclass correlation coefficient (ICC) values of COP measures in this study, which ranged from 0.70-0.93, suggest good to excellent reliability. Thus, for ease of comparison, we considered only the results from the third and final test session for this study.

3.4.1. Effects of age

Overall, older women used slower speeds and an increased number of compensatory submovements as can be seen in data from a representative 60-s trial for one young and one older participant in **Figure 3.2**. In comparison to younger women, older women had 27% decreased COP movement speed, $F(1,23) = 10.9$, $P < .005$, twice as many compensatory submovements, $F(1,23) = 18.7$, $P < .001$, and had an 18% increase in the ratio of peak-to-average COP velocities, $F(1,23) = 5.9$, $P < .05$ (**Figure 3.3**). An interaction between age and movement direction was identified in the number of submovements, $F(1,45) = 5.0$, $P < .05$, evidenced by disproportionate increases in submovements posteriorly in the older adults. Post-hoc tests demonstrated that the most significant age-related changes occurred in the posterior movement speed in 2-cm trials and number of posterior submovements in 4 cm trials (t-test, $P < .008$).

The correlation between COP movement speed and number of compensatory submovements suggests a similar pattern of COP control for both young and older women (**Figure 3.4**, Pearson $r^2 = -.72$ to $-.80$). COP endpoint variability, as evaluated by the effective target size demonstrated no statistically significant differences due to age in a linear mixed model (e.g., mean difference in effective target size < 2 mm, P > .05). As expected, a decreased target size also led to a decrease in the effective target size, $F(2,69) = 16.1$, $P < .001$.

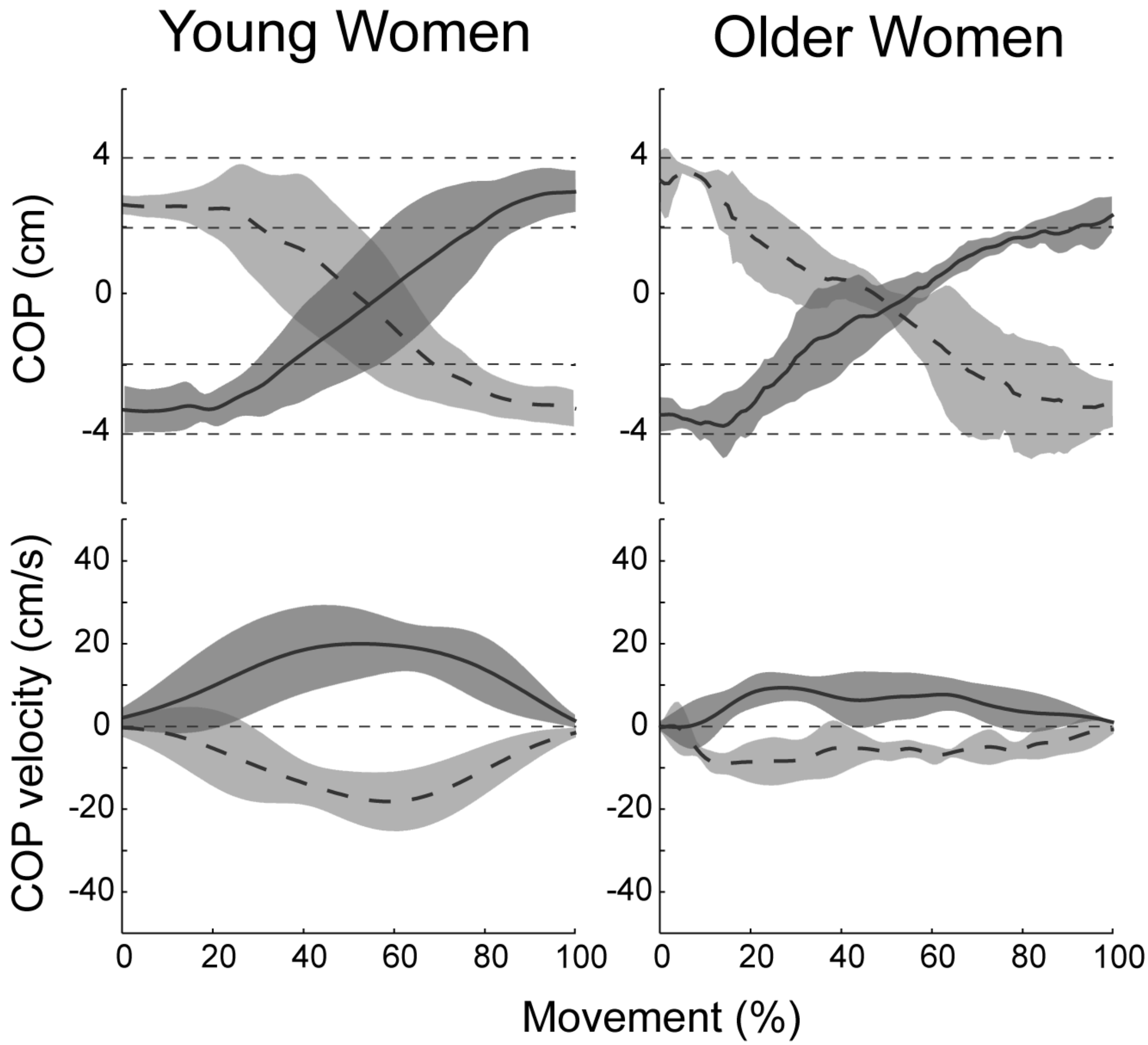

Figure 3.2: COP position and velocity data illustrating the mean (solid line) and standard deviation (shaded area) of anterior (solid center line) and posterior (dashed center line) movements during discrete 2-cm trials, representative of young and older women. Both the anterior target (dashed dark gray lines) and target at a neutral standing posture (dashed light gray lines) are presented in plots of COP position.

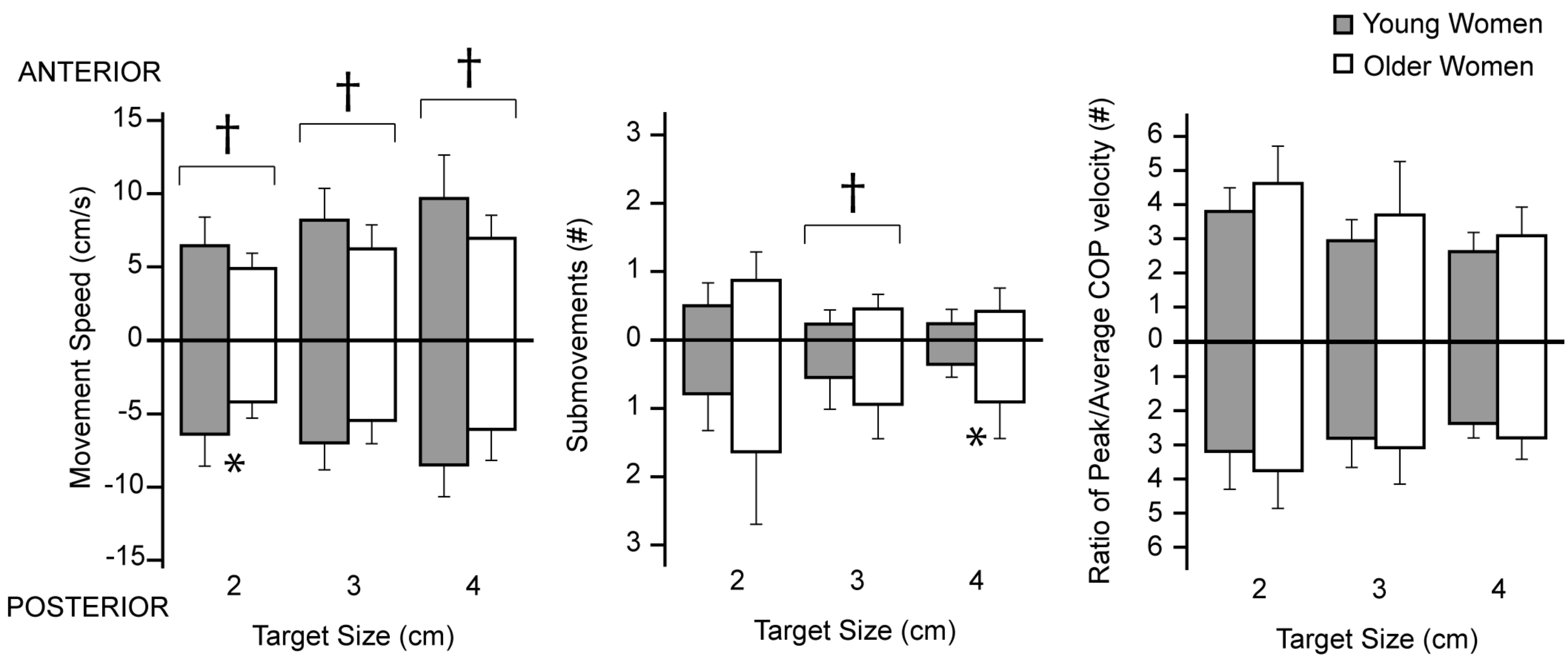

Figure 3.3: **Influence of target size on anterior and posterior COP movements for young and older women. Mean (S.D.) movement speed, number of submovements, and ratio of peak-to-average COP velocity were found to be significantly affected by movement direction and target size in linear mixed models. Age was found to have a significant effect on movement speed and submovements. * and † indicate P < .008 in post-hoc t-tests using Hochberg's step-up method to evaluate age and movement direction differences, respectively.**

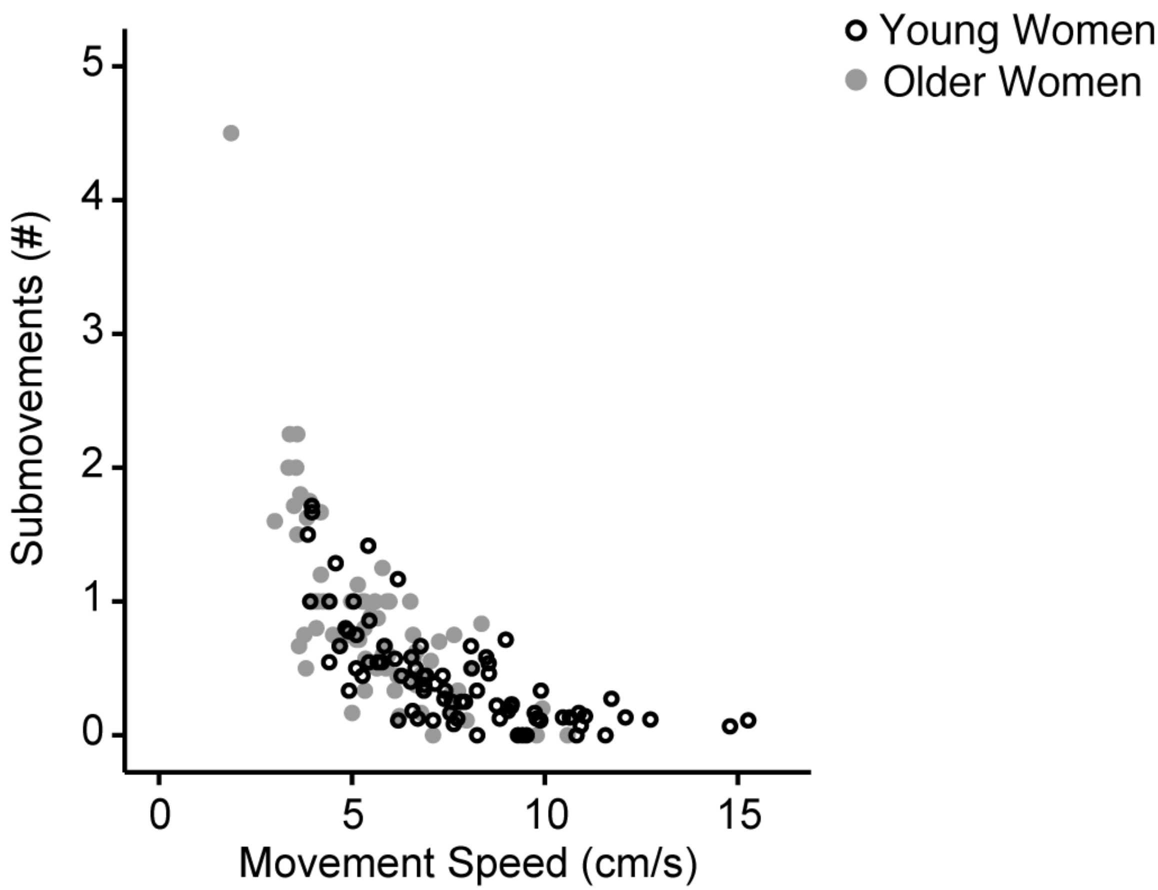

Figure 3.4: Effect of age on the relationship between COP movement speed and number of compensatory submovements. Correlations between movement speed and submovements for young and older women (r^2 = -.72 to -.80) were all statistically significant (P < .01).

3.4.2. Effects of target size

Movement speed slowed in all participants as target size was decreased, F(2,95) = 46.0, P < .001. Older women were significantly slower than younger women across all target sizes, but age-disproportional decreases in movement speed as target size decreased did not reach statistical significance. As target size decreased, the number of submovements, F(2,97) = 17.7, P < .001, and the ratio of peak-to-average COP velocity increased, F(2,96) = 27.1, P < .001.

3.4.3. Effects of movement direction

Posterior COP movements were found to be slower than anterior movements, $F(1,37) = 9.8$, $P < .005$. Furthermore, an increased number of compensatory submovements, $F(1,41) = 27.1$, $P < .001$, and higher ratios of peak-to-average COP velocities, $F(1,39) = 10.7$, $P < .005$, were seen in posterior versus anterior movements. In post-hoc tests, posterior movements were found to be significantly slower across all target sizes and found to have increased submovements in 3-cm trials ($P < .005$).

3.5. Discussion

The purpose of this study was to examine the effect of age, movement direction, and target size on targeted COP movements in healthy women. In doing so, we furthered the understanding of age-related changes on the control of rapid volitional movements. It was hypothesized that in comparison to young, older adults would exhibit slower COP movement speeds, and increased submovements and ratio of peak-to-average COP velocity, particularly in posterior movements. The results partly confirmed our hypothesis, as older women required more frequent submovements to maintain COP accuracy, despite moving slower than young women. Further supporting our hypothesis, is the novel finding of disproportionate increases in compensatory submovements in posteriorly directed COP movements with increased age: a finding consistent with older adults taking multiple steps more often than young adults when responding to posterior perturbations (Luchies et al., 1994, Schulz et al., 2005), as both represent an increase in the use of corrective actions.

3.5.1. Age-related changes in COP control

Our COP movement findings are consistent with planar arm movement studies showing that movement speed decreases and the number of compensatory submovements increases with age (Goggin & Meeuwsen, 1992, Ketcham et al., 2002). Considering the stochastic optimized submovement model of volitional movements (Bullock & Grossberg, 1988, Crossman & Goodeve, 1983, Meyer et al., 1988), age-related increases in neuromuscular noise (Galganski et al., 1993) would be expected to lead to decreased movement speed and increased submovements. Given the role that neuromotor noise is known to have in the production of volitional upper extremity movements (Fishbach et al., 2005, Novak et al., 2000, Schmidt et al., 1979, Smits-Engelsman et al., 2002), our findings of age-related changes in COP control are not unexpected. The present findings of increased ratios of peak-to-average COP velocities with age are also consistent with increased motor unit variability and antagonist coactivation in the submaximal force generation of older adults (Galganski et al., 1993, Roos et al., 1997, Tracy & Enoka, 2002).

When spatiotemporal demands were placed on rapid COP movements in this study, the age-related deficits in COP control performance, particularly in posterior movements, may be a result of decrements in energy storage capacity and motor control (Cavagna & Cittero, 1974, Grillner, 1985, Guiard, 1993). Tibialis anterior activity is required for generating posterior COP movements via dorsiflexion, while the much larger triceps surae can actively apply a moment to move the COP anteriorly. The muscle-tendon lengthening speed utilized by participants in this study may seem small, but based on a biphasic, ballistic-like control pattern of upright stance (Loram & Lakie, 2002) rapidly alternating torques can be used by participants to control an inverted pendulum. Thus, ankle plantarflexors such as the triceps surae in unison with the Achilles tendon might play a significant role in compensating for the effects of gravity while leaning anteriorly (Loram & Lakie, 2002). Furthermore, given the inverse relation between the ratio of peak-to-average velocity and energy efficiency (Nelson, 1983), increases in the ratio of peak-to average COP velocities among older adults, when compared to young, suggest a decrease in the efficiency of movements of older adults. Slow discrete movements may lead to a greater number of inflections (i.e., submovements) in older adults due to neuromuscular changes and central planning deficits (Galganski et al., 1993, Goggin & Meeuwsen, 1992). An increase in motor unit recruitment and demands for information processing in discrete movements, as suggested by decreases in efficiency, can lead to increases in neuromotor noise, and thereby to an increased number of corrective submovements (Meyer et al., 1988). The increased number of corrective submovements indicate that our older adult participants still had sufficient postural control "reserve" to vary their response despite possible decrements in sensorimotor capacities. The postural control "stressor" (i.e., target size and movement amplitude) in the present study was not challenging enough to cause a loss of balance in these healthy older women. However, if applied to a more clinically impaired older adult population, the stressor might uncover a further decline in the complexity of their postural response, consistent with the concept of impaired homeostasis with increasing frailty (Lipsitz, 2008, Manor et al., 2010).

3.5.2. The dependence of COP speed and accuracy on movement direction

Previous studies of COP movements in response to changes in posture had found no difference between the anterior and posterior phases of continuous movement (Danion et al., 1999, Duarte & Freitas, 2005). However, there are two significant differences in our study that may help explain why this was not found in the present study: (1) discrete, rather than continuous, movements were used in our study, which would increase movement times and corrective responses (Smits-Engelsman et al., 2002) amenable to the effects of movement direction; and (2) the use of a target at both the mid-foot, at a neutral standing posture, and at the toes, near the anterior limit of the FBOS, which would affect the underlying postural sway characteristics in the task (Duarte & Zatsiorsky, 2002) and present unique anatomical constraints.

Discrete COP movements to target are usually made in the presence of sensory feedback (Haken et al., 1985, Smits-Engelsman et al., 2002) and therefore may be susceptible to both changes in plantar cutaneous receptor sensitivity throughout the length of the foot and with increased age (Inglis et al., 2002, Wells et al., 2003). In particular, anteroposterior postural control of upright standing posture is significantly affected by diminished plantar cutaneous sensation (hypoesthesia) (McKeon & Hertel, 2007) and an increased dependence on distal sensory feedback has been shown with increased age (Brumagne et al., 2004). In addition, the increased number of submovements and decreased efficiency seen in posterior versus anterior COP movements may be due to intrinsic differences in structure between the midfoot and hallux, as foot structure has been shown to be a significant contributor to dynamic foot function (Cavanagh et al., 1997, Morag & Cavanagh, 1999). Thus, physiological and anatomical differences along the length of the foot may partly explain the observed directional difference of COP control.

3.5.3. Study limitations

First, a relatively modest sample size limited statistical power and the choice of female subjects means that the results may not be generalizable to men without further research. Second, despite our efforts to match groups for foot length and body mass index, older women were slightly shorter than young. This would not be expected to have biased the results significantly, as the shorter height should have led older women to have less angular inertia to overcome than young. Third, we did not control for how participants prioritized task instructions. Even though participants were instructed to place equal priority on accuracy and speed, accuracy may have been prioritized over speed in some older adults, leading to greater inter-subject variability. Fourth, the feedback involved in this experiment, using visual feedback from a computer display to help volitionally shift the COP, is artificial, hence care must be taken when extending findings to whole body movements such as stooping or reaching without COP feedback. It is a limitation that we did not evaluate more than two underfoot target positions, movements to the edge of the FBOS, or mediolateral COP movements. Future studies are required to examine the underlying mechanisms behind age-related deficits in posterior COP movement performance, and further explore whole body COP control during daily movements that present the older adult with a risk for falling, such as returning from reaching, stooping, and forward bending tasks. Lastly, while the use of lean movements simplified the postural control task it may limit the generalizability of the results to reaching movements.

3.6. Conclusions

Despite moving more slowly, older women needed to take more frequent submovements to achieve the same level of COP accuracy as younger women, particularly when moving posteriorly. This may provide evidence of a compensatory strategy used by older adults for preventing backward falls. However, the increased number of corrective submovements suggests one mechanism by which frailer older adults may not be able to maintain balance if hurried.

3.7. Significance

This study on whole body accuracy-constrained COP movements broadens the current literature of the effects of age on accuracy-constrained movements beyond planar arm movements to whole body movements. The use of more frequent and variable COP movements by older adults when maintaining balance, particularly posteriorly, is undesirable, as it would present a greater risk for a loss of balance when hurrying. Healthy older adults achieved comparable accuracy in their discrete volitional COP movements as healthy young adults. However, the increased number of corrective submovements, suggests that frailer older adults may not be able to be able to maintain balance if hurried. Future studies might explore underlying mechanisms behind age-related deficits in posterior COP movement performance, and further explore whole body COP control during daily movements that present the older adult with a risk for falling, such as returning from reaching, stooping, and forward bending tasks.

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CHAPTER 4

AGE-RELATED CHANGES IN SPEED AND ACCURACY DURING RAPID TARGETED COP MOVEMENTS NEAR THE POSTERIOR LIMIT OF THE BASE OF SUPPORT

4.1. Abstract

Backward falls are associated with injurious falls, particularly among older women. Older adults are more apt to take multiple steps when recovering from large posterior perturbations. The primary goal of this study was to examine the effects of age on the speed and accuracy of primary submovements (PSMs) in large and small posterior leaning tasks. We hypothesized that in comparison to young women, older women would display a disproportionate decrease of speed and accuracy (e.g., increased number of submovements and increased incidence of COP undershooting) in their COP PSMs, as movement amplitude increases. Ground reaction forces were recorded from healthy young (n=13) and older (n=12) women while performing rapid targeted COP movements of small and large amplitude in an upright stance. Repeated-measures linear mixed models were used in this analysis. Results showed that PSM speed was disproportionately decreased in older women when compared to young as movement amplitude increased. Even though older women achieved similar endpoint accuracy, they

demonstrated a 2 to 5-fold increase in the incidence rate of the PSM undershooting the desired target in large-amplitude movements. Overall, the posterior COP movements of older women were 41% slower and used 43% more secondary submovements than young women. We conclude that undershooting PSMs and increased secondary submovements are indicative of an increasingly conservative strategy used by older adults near the limits of the functional base of support (FBOS) that may explain their slower speeds during whole body movements to maintain upright balance.

4.2. Introduction

Falls are among the most significant causes of mortality and serious injury in adults over 65 years of age (Riley, 1997, Rubenstein, 2002), particularly for older women (Norton et al., 1997, Tinetti et al., 1995). Backward falls are of particular significance, as they are likely to lead to wrist fractures among older women (Nevitt & Cummings, 1993). To maintain balance during common daily activities such as standing up from a chair or getting up from the floor after stooping or crouching, large and rapid center of pressure (COP) movements in the anteroposterior direction are often required (Breniere & Do, 1986, McIlroy & Maki, 1993, Scholz, 2001). While leaning maximally, older adults have increased COP variability and reduced spatiotemporal stability margins in the anteroposterior direction, when compared to young adults (van Wegen et al., 2002). The use of more frequent and variable COP movements by older adults when maintaining balance near the anteroposterior limits of stability is undesirable, as it would present a greater risk for a loss of balance, particularly in response to large perturbations.

The need for rapid movements of the COP during a recovery from a potential loss of balance would be expected to lead to tradeoffs in accuracy, as suggested by Fitts' Law

(Fitts, 1954, Plamondon & Alimi, 1997). However, studies on accuracy-constrained whole body movements have identified violations to Fitts' law (Danion et al., 1999, Duarte & Freitas, 2005), which may be explained by the body's inherent variability in an upright stance (Duarte & Zatsiorsky, 2002). Considering the optimized submovement model (Meyer et al., 1988), based on an alternate definition for speed-accuracy tradeoffs (Meyer et al., 1982, Schmidt et al., 1979):

$$W_e = a + b \cdot S$$

where S is the mean movement speed, W_e is the effective target size, and a and b are experimentally derived constants, we can directly account for variability. Mean movement speed is derived from the mean movement amplitude and mean movement time, whereas the effective target size is defined as four times the standard deviation (SD) of the COP endpoint position. Depending on the operational demands of a task, individuals can employ low force, slow submovements to achieve a high level of accuracy or high forces so as to generate rapid movements. However, to compensate for the increased motor noise, rapid movements would necessitate an increased number of submovements to accurately reach a desired target. Thus, rapid targeted movements can be divided into either a primary, ballistic submovement (PSM) or a series of secondary, corrective submovements, which can provide insight into changes in whole body movement strategies.

Age has been shown to affect the tradeoffs between movement speed and accuracy. During spatially constrained arm movements, older adults use slower movements than young adults to achieve a similar level of accuracy (Goggin & Meeuwsen, 1992, Ketcham et al., 2002, Salthouse, 1988). However, few data exist regarding COP speed-accuracy changes with age under challenging balance conditions, such as those that might precede a loss of balance. During undisturbed upright stance, older adults have demonstrated greater delays before deploying feedback control of COP movements (Collins et al., 1995), which is consistent with findings of increased distal muscle latency in older adults during postural perturbations (Woollacott et al., 1986). Older adults have shown a higher variability in postural responses to more challenging balance conditions (Alexander et al., 1992). Thus, we would expect older adults, who tend to have greater variability during upright stance (Pyykkö et al., 1990) and slower voluntary movements when correcting postural perturbations (Stelmach et al., 1989), to use more conservative COP movements near the limits of stability, in contrast to young. On the other hand, the use of hand support has been shown to reduce postural sway with just a light touch (Jeka & Lackner, 1994, Jeka, 1997) and would be expected to reduce age-related changes in postural control. Thus, the provision of hand support would provide us with further insight into the benefits of additional support in the COP control of large amplitude movements.

The primary goal of this study was to examine the effects of age on the speed and accuracy of PSMs in large and small posterior leaning tasks. We hypothesized that in comparison to young women, older women would display a disproportionate decrease of speed and accuracy (e.g., increased number of submovements and increased incidence of COP undershooting) in their COP PSMs, as movement amplitude increased. We expect that the analysis of COP primary and secondary submovements may help explain the increasingly conservative strategy used by older adults near the limits of the FBOS to maintain safe upright stance. These data may ultimately provide insight into mechanisms underlying fall avoidance in older adults, particularly in the backward direction.

4.3. Methods

4.3.1. Participants

Thirteen young and twelve older healthy, community-dwelling, women were recruited for this study (**Table 4.1**). All young participants completed a medical history questionnaire and older participants were physically screened by a nurse practitioner, so as to exclude those with significant musculoskeletal or neurological findings. The two groups were of similar weight and body mass index, but older women were shorter and had a shorter functional base of support than young women. All participants provided written informed consent as approved by University of Michigan Medical School Institutional Review Board procedures.

Table 4.1: Mean ± SD participant characteristics.

	Younger Women (N = 13)	Older Women (N = 12)
Mean age (years)[b]	23 ± 3	76 ± 6
Height (cm)[a]	164 ± 6	159 ± 5
Weight (kg)	63 ± 11	63 ± 11
Body mass index (kg/m^2)	23 ± 4	25 ± 5
Foot length	26 ± 1	25 ± 3
Functional base of support (cm)[b]	20 ± 1	17 ± 2

[a] Indicates age group effect $P < .05$
[b] Indicates age group effect $P < .005$

4.3.2. Instrumentation

Participants stood on a single ground-level six-channel force plate (OR6-7-1000, AMTI, Watertown, MA) with data collected at a sampling rate of 100 Hz. Kinematic data were collected using a three camera, three-dimensional, motion capture system to verify movement strategies (two Optotrak 3020 and one Optotrak Certus Camera, Northern Digital, Inc., Waterloo, Canada). Infrared light emitting diodes were placed on the right leg over the lateral malleolus, heel, fifth metatarsalphalangeal joint, femoral epicondyle; on the greater trochanter; and on the right shoulder. Kinematic data were sampled at 25 Hz. All data were recorded using the Optotrak system and First Principles software (Northern Digital, Inc., Waterloo, Canada).

4.3.3. Protocol

Before performing accuracy-constrained COP movements, participants performed a series of 30-s calibration trials in an upright bipedal stance on the force plate with their arms crossed over their chest. Participants first stood in a neutral standing posture to establish the 'CENTER' target zone for testing. Participants then leaned as far anteriorly or posteriorly as they could without losing their balance (i.e., fall or step) over the course of a 30-s trial. The anterior and posterior limits of the participant's COP excursion were used to determine their functional base of support (FBOS), a quasi-static limit of stability (King et al., 1994), and set the 'ANTERIOR' and 'POSTERIOR' target zones for testing. Participants wore standardized canvas shoes to control for the shoe-floor interface during all tests. During hand support trials, participants used the side hand grips of a walker placed around the force plate at a fixed height from the floor (approximately hip height). A thin wooden platform, outlining the anterior and lateral edges of the feet was mounted on top of the force plate to control foot placement (i.e., stance width and anterior edge of the base of support). One of two different sized targets (2 and 6 cm) were centered at either the posterior edge of the FBOS (posterior target zone), the anterior edge of the FBOS (anterior target zone), or at the neutral standing posture (center target zone, **Figure 4.1**). During all trials, real-time, COP feedback was provided on a display monitor positioned directly in front of the participant. A pair of horizontal lines defined the participant-specific ANTERIOR, POSTERIOR, and CENTER target zones that were used in all conditions. At the conclusion of test trials, participants performed rapid COP movements without any spatial constraints for use in evaluating their COP slew rate, the maximal change in COP position at any point in time.

Volitional COP movements were directed towards pseudo-randomized targets on three separate laboratory test sessions. Performance measures of volitional COP movements were found to demonstrate good to excellent reliability (intraclass correlation coefficient [ICC] = 0.70-0.93). Thus, for ease of interpretation only data from the final session are presented. The final session consisted of 24 trials with a 60-s duration, out of which 8 trials, corresponding to two COP movement amplitudes (small-amplitude movement from ANTERIOR to CENTER target versus large-amplitude movement from ANTERIOR to POSTERIOR target), two target sizes (2 versus 6 cm), and two hand support conditions (with versus without hand support) were analyzed. Both the COP target position and target size were randomized within all trials.

In all accuracy-constrained movements, participants were instructed to move their COP as fast and as accurately as possible. A time-varying (1-3 sec) auditory tone was used to cue the start of the movement from the ANTERIOR target zone to the desired target and back. In discrete movements with hand support, participants were instructed to hold on to the walker to maintain balance, whereas in trials without hand support, participants were instructed to lean their body, while keeping their arms crossed.

4.3.4. Data processing and analysis

Only posterior COP movements from the ANTERIOR target zone were analyzed in this study. We focused solely on posterior movements from the limits of the FBOS to investigate the more challenging postural demands relevant to the study of backward falls. The first posterior movement of each trial was not analyzed so as to provide some time for re-familiarization at the start of each task. The range of COP movement amplitudes explored in this study encompass a large range of excursion feasible in a feet-in-place balance recovery strategy.

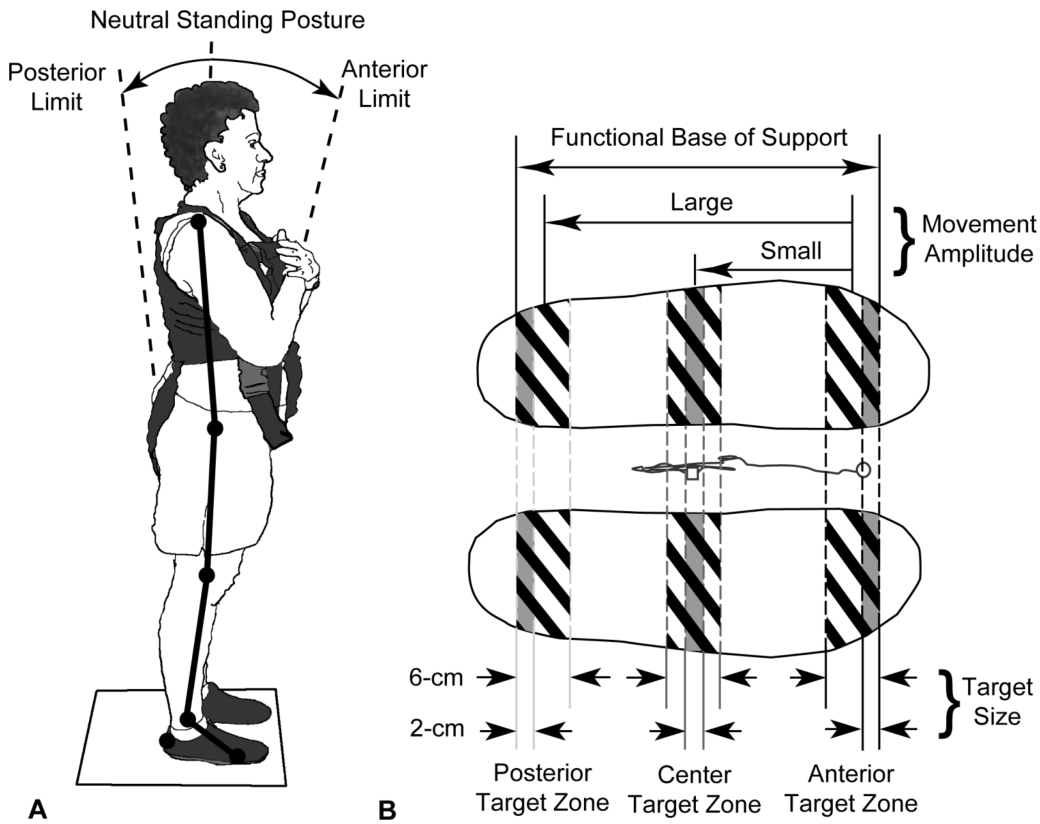

Figure 4.1: A) Illustration of participant at a neutral standing posture and their representative rigid link model. B) Illustration of experimental setup showing an exemplary small-amplitude COP movement from the anterior target zone (dashed black lines), to the center target zone (dashed dark gray lines) located at a neutral standing posture. Dashed light gray lines indicate the posterior target zone used in large-amplitude movements. Hatched areas indicate 6-cm targets, whereas gray shaded areas indicate 2-cm targets.

Customized Matlab (v7.4, Natick, MA) data processing software routines were used to process the data. A fourth order, zero-lag, low-pass Butterworth filter with a 10 Hz cutoff frequency was used in processing raw force plate data. A five-point finite difference derivative algorithm was used in calculating both COP velocity and acceleration. Using an automated procedure, overall COP movements were extracted from each test trial. Based on the 'soft symmetry' method (Fishbach et al., 2005, Fishbach et al., 2007, Novak et al., 2000), we adapted a similar definition to a primary COP submovement using two thresholds of the COP velocity, V_1 and V_2. The onset time, T_{on} was defined as the last time that the COP velocity exceeded V_1 before reaching V_2. Similarly, the offset time, T_{off} was defined as the first time that the COP velocity reaches V_1 after decreasing below V_2 (**Figure 4.2**). In this study, V_2 was set to V_{max}, the maximum COP velocity, and V_1 was set to 10% of V_{max}.

For the purpose of investigating discrete COP movement strategies, we calculated the movement speed and COP endpoint position of primary submovements. Primary submovement (PSM) speed and PSM COP endpoint position were calculated using the COP position at T_{on} and T_{off}. Overall movement speed was calculated by dividing the movement amplitude (e.g., effective distance between onset and offset of COP movement) by the movement time (e.g., time elapsed between onset and offset of movement). All posterior movements were arrested at either the CENTER or POSTERIOR target zones, corresponding to a neutral standing posture or the posterior limit of the FBOS, respectively. Movement speed and COP endpoint variability was evaluated using the standard deviation of all PSM and overall movements in each 60-s trial. The number of submovements was defined as the number of pairs of zero acceleration crossings in the COP trajectory, between V_{max} and the offset of the movement (**Figure 4.2**). COP endpoint position was defined as the position at the offset of movement within the target zone, relative to the nearest boundary of the target with respect to the center of the movement. Based on the COP endpoint position of a PSM, the incidence rate of undershooting (i.e., a PSM COP endpoint terminated before reaching the desired target zone) was further evaluated. For each trial, the mean movement speed and endpoint position of all PSMs and overall movements were calculated for further analysis, in addition to the mean number of submovements.

Baseline postural sway at an upright stance and COP slew rate were measured to quantify age-related changes in balance capacity. Using the 30-s evaluation of the neutral standing posture, the root mean square (RMS) values of COP excursion were calculated. COP slew rates with and without hand support were measured using volitional COP movements without accuracy constraints, when participants were instructed to move as fast and as far as possible. After identifying the peak COP velocity in the trial, the slew rate was calculated by measuring the COP excursion from 20% to 80% of the maximal velocity, and dividing by the elapsed time, so as to remain within the linear range of COP movement.

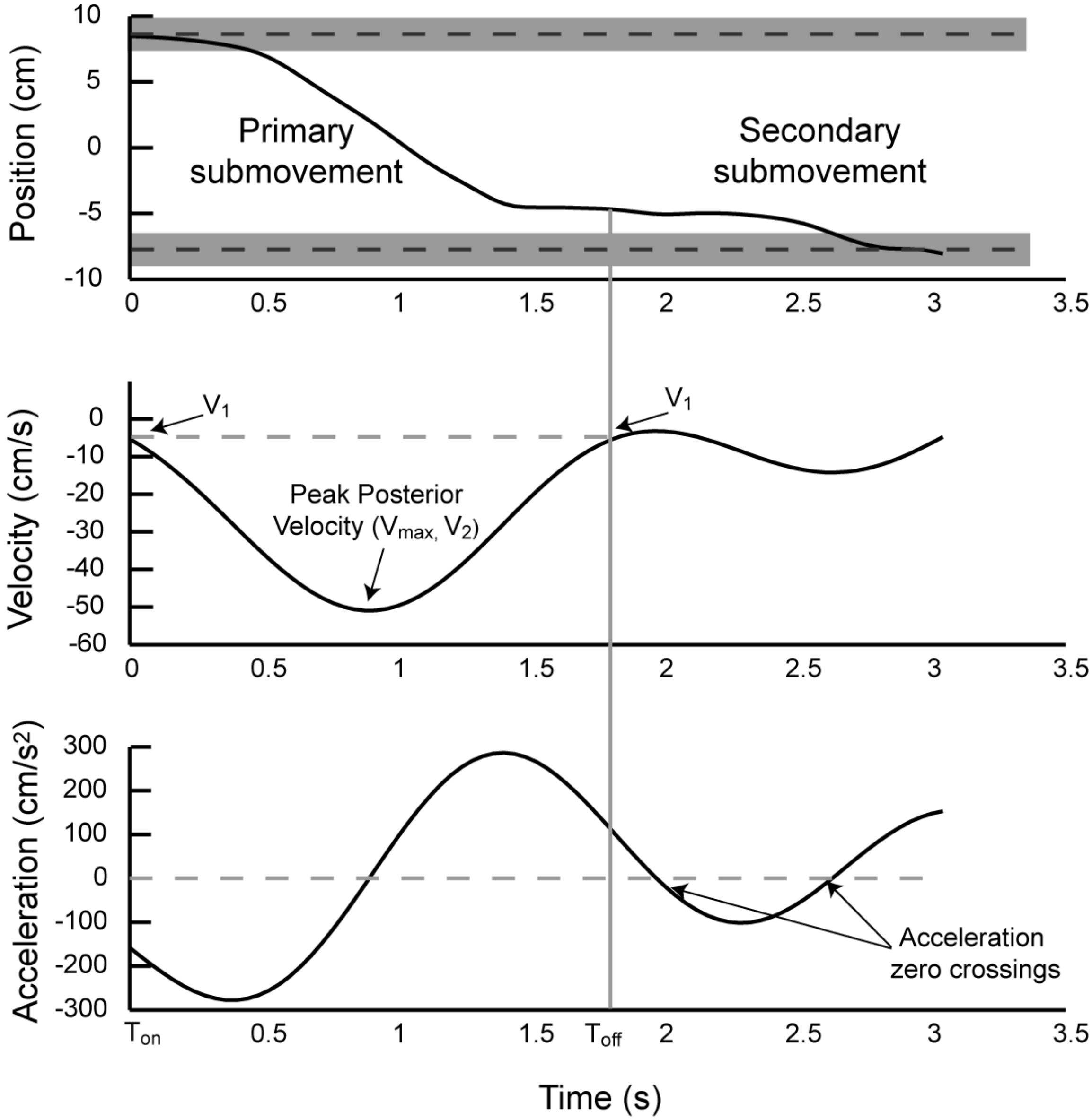

Figure 4.2: Position, velocity, and aceleration of a typical COP movement, decomposed by its primary and secondary submovement. V_1 represents the threshold frequency, equivalent to 10% of the peak posterior velocity, V_{max}. Pairs of acceleration zero crossings after V_{max} represent a corrective submovement. Primary submovements are defined by the onset time, T_{on}, and the offset time, T_{off}.

4.3.5. Statistical Analysis

All statistical analyses were carried out in SPSS 16.0 for Windows (SPSS Inc., Chicago, IL). To identify age group differences, independent sample t-tests were used for all subject characteristics. Repeated-measures mixed-model analyses of variance, using a restricted maximum likelihood method, were used to examine the effect of age, hand support condition (i.e., with or without hand support), movement amplitude (i.e., small or large-amplitude movement), and target size, as well as all second-level interactions with age. The linear mixed models assumed a first-order autoregressive covariance structure. To account for multiple comparisons, post-hoc tests were carried out using Hochberg's step-up method, and $P < .05$ was used for statistical significance.

4.4. Results

Older women demonstrated no significant differences in their baseline postural sway (i.e., COP RMS error) with eyes open in an upright stance when compared to younger women (**Table 4.2**). Overall, older women had 25% slower COP slew rates than younger women without hand support and statistically significant 42% slower slew rates with hand support ($P < .05$). **Table 4.3** presents discrete COP performance characteristics. The means and standard deviations of movement speed and COP endpoint position of primary submovements (PSMs) and overall movements are presented in addition to the mean and standard deviation of the overall number of submovements.

4.4.1. Primary Submovement Characteristics

Effects of Age. Older women demonstrated slower PSMs compared to young women ($F = 21.4$, $P < .001$, **Table 4.4**), particularly in large-amplitude movements versus small-amplitude movements ($F = 20.9$, $P < .001$, **Figure 4.3**). PSM COP endpoint position tended to decrease in older versus younger women, and a borderline interaction between age and movement amplitude was observed, but neither were statistically significant after correcting for multiple comparisons (**Table 4.4**). The incidence rate of undershooting (i.e., a PSM COP endpoint terminated before reaching the desired target zone) significantly increased in older versus younger women (Pearson $\chi^2 = 17.5$, $P < .001$). In particular, a 2 to 5-fold increase in the incidence rate of undershooting was found in the large-amplitude movements of older women, when compared to young women.

Table 4.2: Mean ± SD participant baseline measurements.

	Younger Women	Older Women
COP slew rate (cm/s)	73 ± 27	55 ± 24
COP slew rate w/HS (cm/s)[a]	177 ± 93	102 ± 36
COP RMS error (mm)	4 ± 1	5 ± 2

Notes: COP = center of pressure, HS = hand support.
[a] Indicates age group effect $P < .05$.

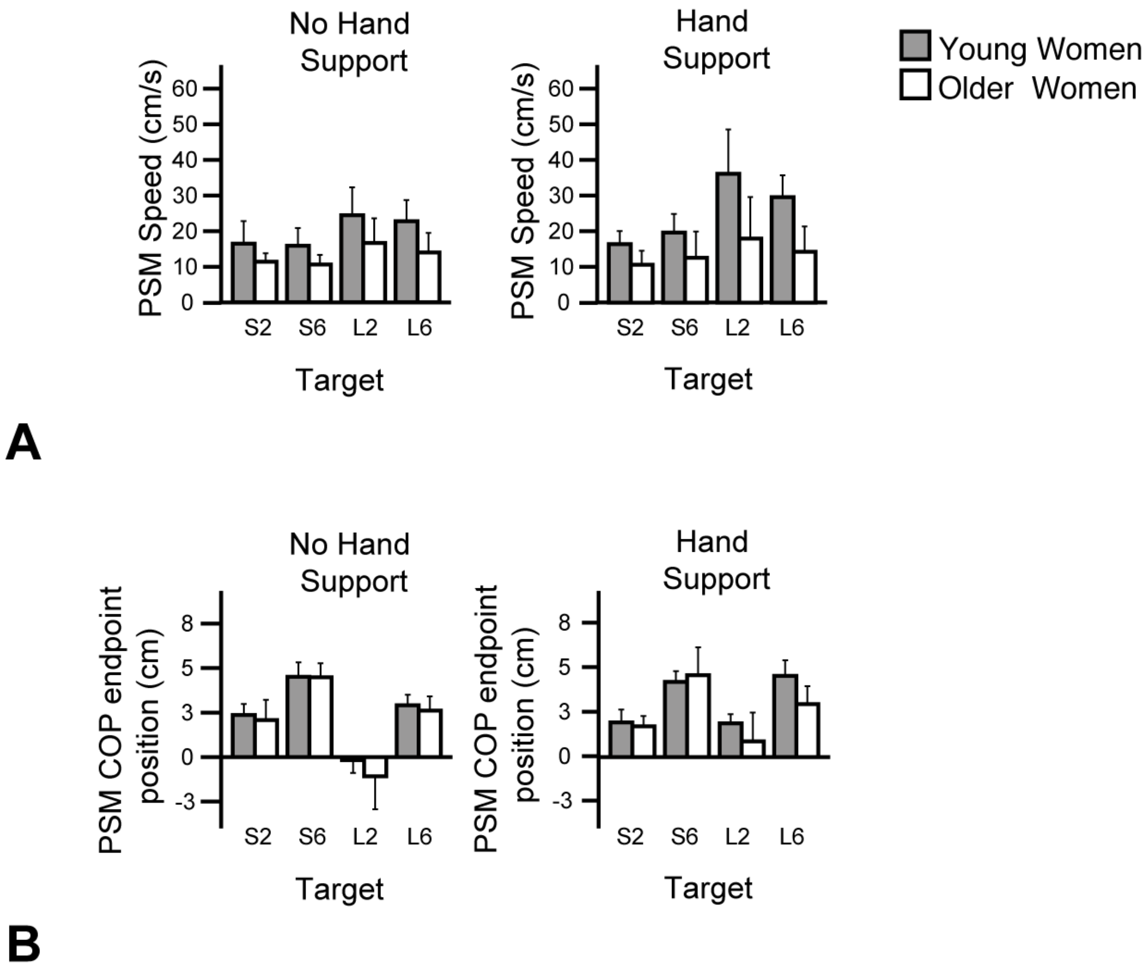

Figure 4.3: Mean ± SD values of A) primary submovement (PSM) speed and B) COP endpoint position during small-amplitude (S2 & S6) and large-amplitude movements (L2 & L6) to 2-cm and 6-cm targets, respectively.

Table 4.3: Mean and standard deviation (SD) of raw dependent variables for healthy young (Y) and older (O) women, organized by task condition.

Condition	No Hand Support		Hand Support		Small-Amplitude Movement		Large-Amplitude Movement		2-cm Target Size		6-cm Target Size	
	Mean	(SD)	Mean	(SD)	Mean	(SD)	Mean	(SD)	Mean	(SD)	Mean	(SD)
COP movement parameter, by group												
PSM Speed (cm/s)												
Y	21	5	26	5	18	4	29	6	23	5	23	5
O	14	4	15	3	12	3	16	4	15	3	14	4
PSM COP endpoint position (cm)												
Y	2.5	1.3	3.3	1.0	3.3	1.3	2.4	1.0	1	1.0	4	1.0
O	2	1.4	2.7	1.1	3.4	1.3	1.4	1.3	0.8	1.2	3.7	1.4
Number of Submovements (#)												
Y	1.1	0.7	0.9	0.4	1.1	0.6	0.9	0.5	1.4	0.9	0.7	0.3
O	1.5	0.8	1.4	0.8	1.4	0.9	1.6	0.8	1.9	1.1	1.1	0.7
Movement speed (cm/s)												
Y	14	5	19	5	12	4	21	6	15	5	18	5
O	9	3	10	3	8	3	11	4	9	3	10	3
COP endpoint position (cm)												
Y	2.3	0.8	3	0.8	2.7	0.8	2.6	0.9	1.2	0.5	3.9	1.1
O	2.2	1.0	2.5	0.8	2.7	0.9	1.9	0.9	1	0.5	3.4	1.2

Notes: COP = center of pressure, PSM = primary submovement.

Table 4.4: Effect of age, hand support, movement amplitude, and target size on primary submovement (PSM) characteristics

Effect	F	p-value
	PSM Speed (cm/s)	
Age	21.438	<.001 [a]
Hand Support	6.210	.015
Movement Amplitude	106.488	<.001 [a]
Target Size	1.267	.263
Age * Hand Support	4.986	.029
Age * Movement Amplitude	20.948	<.001 [a]
Age * Target Size	.007	.935
	PSM COP endpoint position (cm)	
Age	6.111	.022
Hand Support	11.022	.001 [a]
Movement Amplitude	81.607	<.001 [a]
Target Size	355.089	<.001 [a]
Age * Hand Support	.395	.532
Age * Movement Amplitude	7.874	.006
Age * Target Size	.619	.434

[a] Denotes a significant difference, as determined by Hochman's step-up test

Other Effects. Overall, PSM speeds increased in large-amplitude movements versus small-amplitude movements ($F = 106.5$, $P < .001$), while demonstrating a 36% decrease in COP endpoint position ($F = 81.6$, $P < .001$). Decreasing the target size from 6 cm to 2 cm led to decreased PSM endpoint positions ($F = 355.1$, $P < .001$). Furthermore, the incidence rate of undershooting was found to significantly differ due to the effect of hand support, movement amplitude, and target size (Pearson $\chi^2 = 26.5$ - 363.5, $P < .001$). Overall, a significant correlation between PSM and overall movement speed was found (Pearson correlation $R^2 = 0.8$). As seen in **Figure 4.4**, PSM correlated well with overall movement speed in 6-cm trials (Linear fit, $R^2 = 0.6$-0.8), but less well in 2-cm trials as fast PSMs led to more variable overall movement speeds.

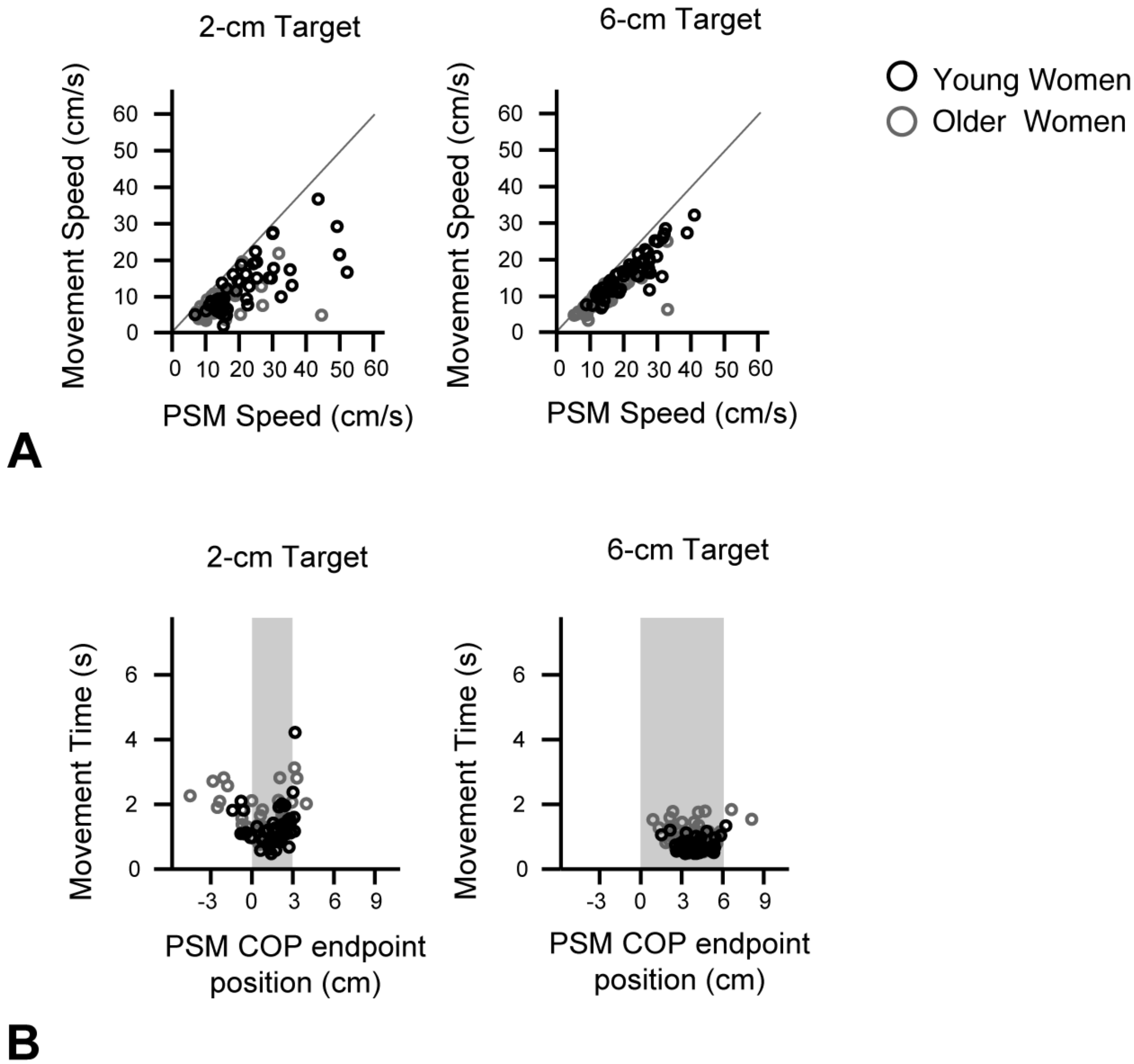

Figure 4.4: A) Correlations between speed of the overall movement and its primary submovement (PSM), and B) correlations between movement time and primary submovement endpoint position in 2-cm and 6-cm targets. Data for both hand support conditions are aggregated.

4.4.2. Overall Movement Characteristics

Effects of Age. Overall, the COP movements of older women were 41% slower than young women ($F = 39.7$, $P < .001$, **Table 4.5**). The effects of age were particularly significant in large versus small-amplitude movements ($F = 29.1$, $P < .001$), and in hand support versus no hand support trials ($F = 11.5$, $P = .001$). Overall, older women had a 43% increase in the number of secondary corrective submovements, when compared to young women ($F = 12.5$, $P = .002$, **Figure 4.5**). Older women tended to terminate their COP movements 13% closer to the initial target boundary ($F = 8.0$, $P = .008$), when compared to young women, and particularly when undergoing large-amplitude movements ($F = 6.2$, $P = .015$). After controlling for multiple comparisons, changes in COP endpoint position due to age and the interaction between age and movement amplitude were not found to be statistically significant (**Figure 4.6**).

Other Effects. The use of hand support led to a 32% increase in the mean movement speed of targeted COP movements, when compared to trials without hand support ($F = 26.5$, $P < .001$). COP speeds were 62% faster in large-amplitude versus small-amplitude movements ($F = 154.5$, $P < .001$). Decreased target size was shown to significantly decrease COP speed ($F = 22.9$, $P < .001$), as seen in **Figure 4.5**.

Overall, compensatory submovements were decreased 56% in COP movements aimed at 6-cm targets as compared to 2-cm targets ($F = 62.3$, $P < .001$, **Table 4.5**). COP endpoint position was increased 10% in movements with hand support versus no hand support ($F = 13.1$, $P = .001$). COP endpoint positions in large-amplitude movement trials decreased 9% compared to small-amplitude movement trials ($F = 18.9$, $P < .001$ **Table 4.5**). The mean COP endpoint position of movements to 2-cm targets laid 70% closer to the initial boundary of the target in comparison to 6-cm targets ($F = 937.3$, $P < .001$). Overall, COP movements were terminated near the center of all targets, at a distance of 55% from the initial target boundary for the 2-cm target, and 61% for the 6-cm target.

4.4.3. Quantifying the use of hand support

We found that both young and older women supported a similar amount of their body weight (BW) through the use of hand support in small-amplitude movements (7 ± 4 versus 5 ± 3 % BW, respectively, $P > .05$) and large-amplitude movements (7 ± 5 versus 5 ± 3 % BW, respectively, $P > .05$). Similarly, evaluating the maximal and minimal shear forces revealed no significant age-related changes in all trials ($P > .05$).

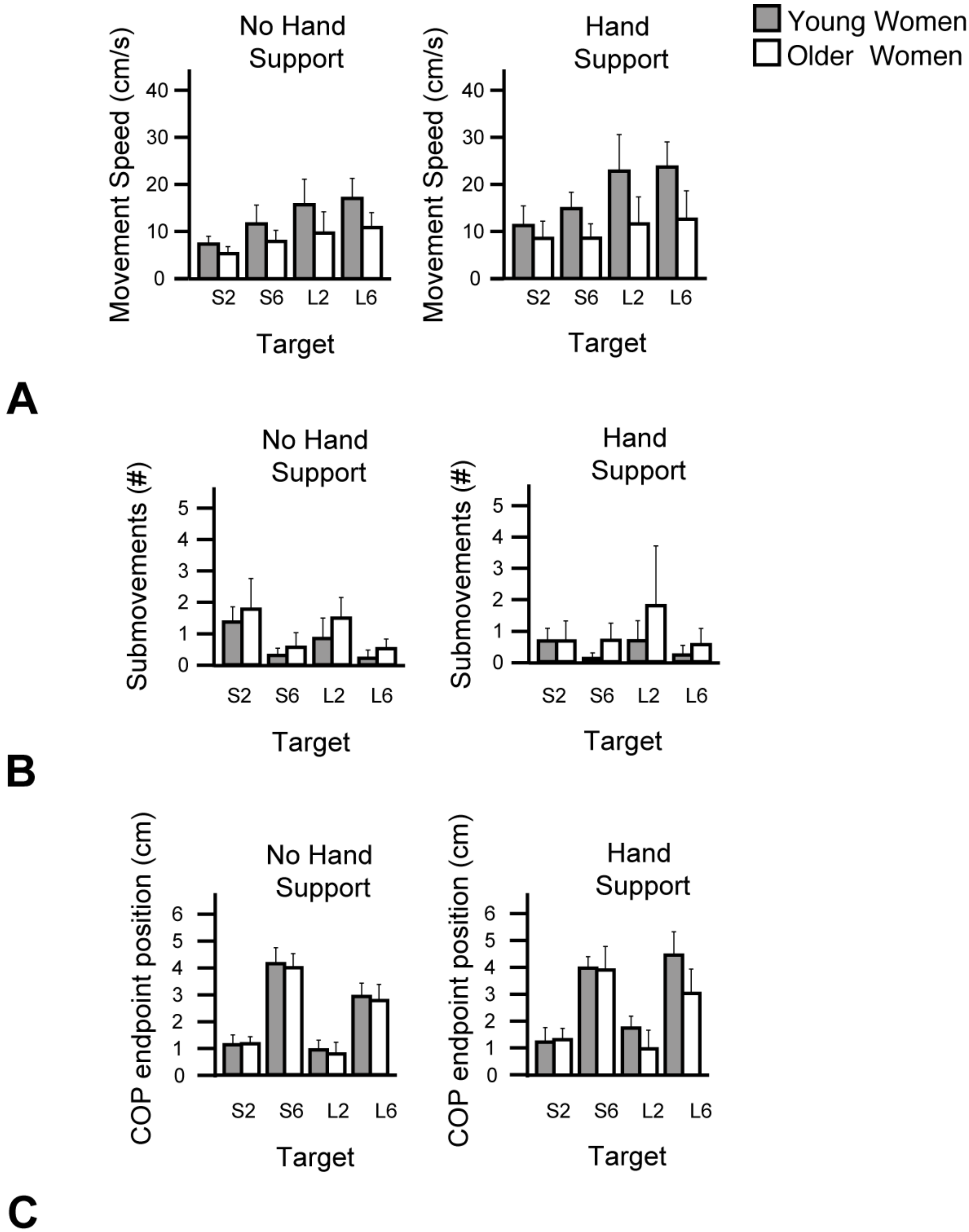

Figure 4.5: Mean ± SD COP A) movement speed, B) number of submovements, and C) COP endpoint position for targeted movements with and without hand support. S2 and S6 refer to small-amplitude movements to 2-cm and 6-cm targets, respectively, whereas L2 and L6 refer to large-amplitude movements to 2-cm and 6-cm targets.

Table 4.5: Effect of age, hand support condition, movement amplitude (small-amplitude versus large-amplitude movement), target size, and their interactions with age on overall COP movement characteristics.

Effect	F	p-value
	Speed (cm/s)	
Age	39.675	<.001 [a]
Hand Support	26.471	<.001 [a]
Movement Amplitude	154.527	<.001 [a]
Target Size	22.885	<.001 [a]
Age * Hand Support	11.537	.001 [a]
Age * Movement Amplitude	29.060	<.001 [a]
Age * Target Size	1.372	.245
	COP endpoint position (cm)	
Age	7.992	.008
Hand Support	13.075	.001 [a]
Movement Amplitude	18.908	<.001 [a]
Target Size	937.333	<.001 [a]
Age * Hand Support	1.939	.169
Age * Movement Amplitude	6.246	.015
Age * Target Size	1.494	.225
	Submovements (#)	
Age	12.523	.002 [a]
Hand Support	1.210	.276
Movement Amplitude	.016	.901
Target Size	62.330	<.001 [a]
Age * Hand Support	.896	.348
Age * Movement Amplitude	2.830	.095
Age * Target Size	.409	.525

[a] Denotes a significant difference, as determined by Hochman's step-up test.

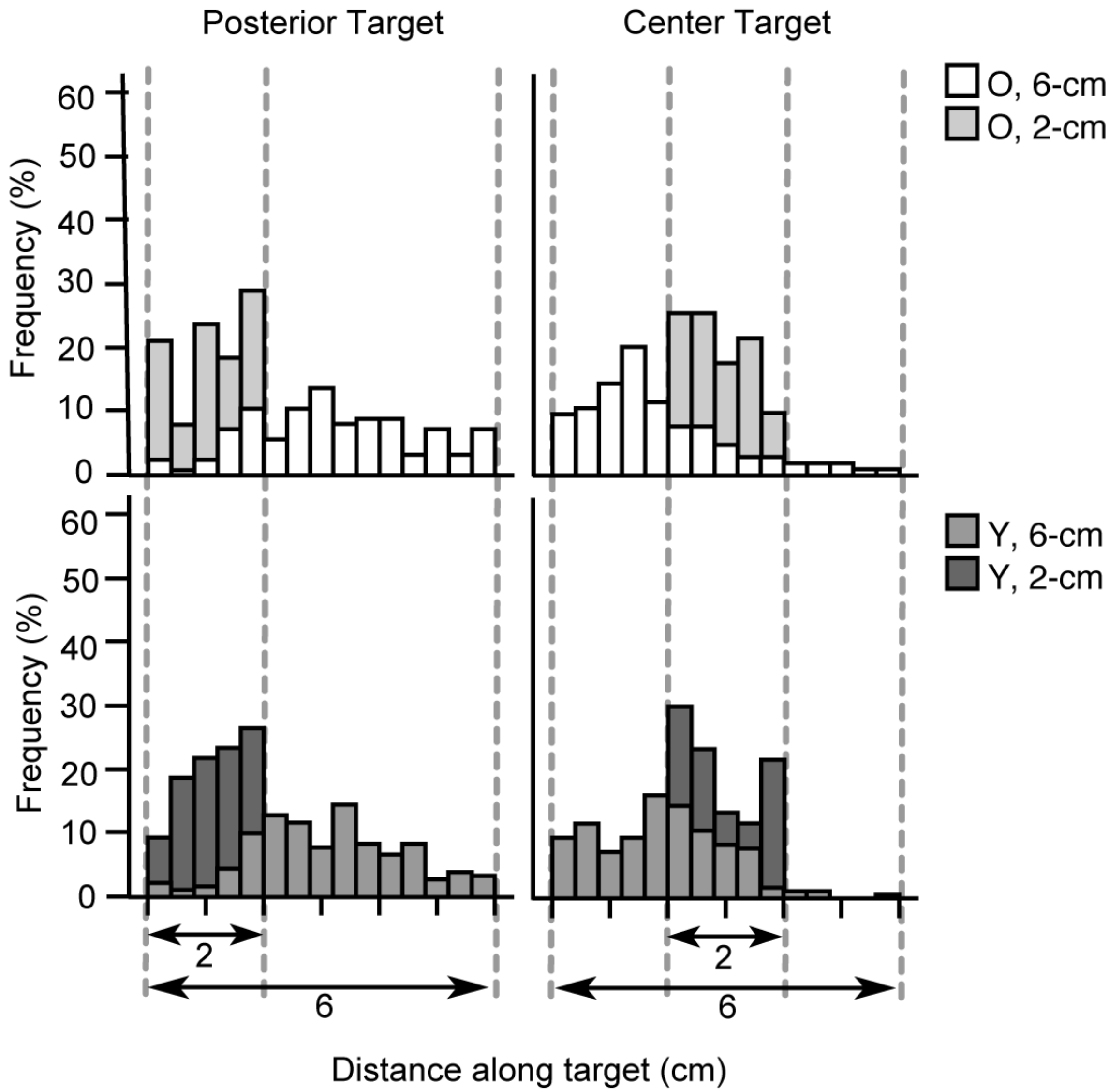

Figure 4.6: Distribution of COP endpoint along the center and posterior target for accuracy-constrained COP trials without hand support of 2-cm and 6-cm targets for both young (Y) and older (O) women. Bin sizes for histograms have a 4-mm width. The leftmost dashed line represents the posterior limit of the FBOS.

4.5. Discussion

To our knowledge, this is the first study to examine the effects of age and movement amplitude on discrete and large accuracy-constrained COP movements in upright bilateral stance. A novel finding was that primary submovement (PSM) speed was disproportionately decreased in older women when compared to young as movement amplitude increased. Even though older women achieved similar endpoint accuracy, they demonstrated a 2 to 5-fold increase in the incidence rate of the PSM undershooting the desired target in large-amplitude movements. Consistent with earlier studies of accuracy-constrained arm movements (Goggin & Meeuwsen, 1992, Ketcham et al., 2002, Salthouse, 1988) and submaximal whole-body movements (Hernandez et al., 2011), older adults were 41% slower and used 43% more secondary submovements than young women. Discrete PSM characteristics were found to be consistent across varying hand support conditions and target size constraints, indicating a potential generalizability to more realistic conditions.

4.5.1 Factors underlying changes in COP speed

The mean speed of targeted COP movements demonstrated significant changes due to age, hand support, movement amplitude, and target size. Older women demonstrated difficulty in increasing speeds of their volitional COP movements near the limits of the FBOS when compared to young women. The lack of movement speed changes by older women for different balance demand conditions suggests a more conservative, error adverse strategy that is consistent with previous motor control studies (Walker et al., 1997, Ketcham et al., 2002). The ability of older women to produce sufficient force is evident by the increases in COP movement speed when no accuracy constraints were in place. However, the decrease in mean COP speed seen in older women, in comparison to young women may be partly attributed to differences in the rate of torque development of ankle plantarflexors and dorsiflexors (Thelen et al., 1996). Furthermore, older women, when compared to young women, terminated a greater number of PSMs well before reaching the desired target. The increased incidence rate of undershooting, suggests that a greater proportion of older women's overall movements required corrective submovements, consistent with previous work (Seidler-Dobrin & Stelmach, 1998).

The use of hand support led to a faster COP movement speed, consistent with findings of enhanced postural stability with hand support (Reginella et al., 1999, Schultz et al., 1992). Even with a small loading, young women took advantage of hand support to produce significantly faster COP slew rates and accuracy-constrained movements, consistent with findings of stabilizing hand reaction forces when the base of support is increased (Bateni & Maki, 2005, Maki & McIlroy, 1997, Maki et al., 2003). The use of a walker for weight support has been associated with high demands on elbow extensors, in addition to wrist flexors and shoulder flexors and adductors (Bachsmidt et al., 2001, Bateni & Maki, 2005, Pardo et al., 1993). Due to age and gender-related changes in upper and lower extremity musculature (Frontera et al., 1991, Metter et al., 1997), older women may find themselves in trouble using hand rails or bars for postural recovery.

Increasingly difficult tasks, as defined by greater movement amplitude and accuracy demands, also led to slower volitional COP movements. Increases in the magnitude of COP excursion, as seen when going from small-amplitude to large-amplitude movement trials, led to faster movements, even as movements were performed towards the anteroposterior limits of stability. Consistent with findings in continuous volitional COP movements among young adults, larger movement amplitudes lead to faster movements, irrespective of target size (Duarte & Freitas, 2005) and movements towards the limits of stability were faster (Appendix B). Decreased target size was also found to significantly decrease mean COP speed as seen in previous Fitts' Law tasks (Duarte & Freitas, 2005, Fitts, 1954, Plamondon & Alimi, 1997).

4.5.2 Movement Accuracy

Older adults have shown a reduced capacity to propel their limbs to a designated target within the initial ballistic phase of their movements and thus utilize a greater number of corrective submovements to successfully reach a target (Bellgrove at al., 1998, Darling et al., 1989). When controlling for strategic differences, older adults can perform accuracy-constrained movements with comparable accuracy but greater hesitancy and more frequent corrective submovements (Morgan et al., 1994). Consistent with previous studies (Ketcham et al., 2002), older women used a greater number of online corrections during volitional movements to achieve similar levels of end-point accuracy, as evaluated by their effective target size. Thus, changes due to natural aging can lead to a more complex movement control strategy when accuracy constraints are placed on COP movements.

The effective target size, mean number of submovements, and COP endpoint position of accuracy-constrained movements were all influenced by target size. The effective target size and COP endpoint position when compared to the number of secondary corrective submovements represent two different aspects of movement accuracy: endpoint accuracy versus on-line control. As seen in previous studies, increased accuracy demands led to an increased number of submovements (Ketcham et al., 2002), and also decreased effective target sizes and decreases in COP endpoint position. Furthermore, decreased target size led to the earlier termination of the COP PSM relative to desired target. Thus, when asked to place an equal priority on speed and accuracy, participants use PSMs that undershot their desired target more often when target size decreased.

Decreased balance demands (i.e., hand support or smaller movement amplitudes) also led to COP endpoint positions closer to the limits of the FBOS. Postural sway increases as a standing adult leans farther away from their comfortable upright stance (Duarte & Zatsiorsky, 2002), and thus we would expect increases in postural sway near the limits of the FBOS or in movements without hand support, which would lead to the use of larger margins of stability to account for expected changes in postural sway. This change in strategy is evidenced by the decreased PSM COP endpoints in large-amplitude movements.

4.5.3 Limitations

As target positions were customized for each participant, changes in outcome measures due to age and types of trial may be partly explained by individual differences in FBOS lengths and desired movement amplitudes. The limited number of disproportionate age effects seen in this study may have been attributable to the small sample size. Given that the number of trials that can be run without fatiguing a subject is limited, we were unable to fully isolate the effects of target position or movement direction on speed-accuracy tradeoffs. Furthermore, the use of a 2-cm target size as the most challenging target, may have limited the power for this study to detect interactions between age and target size. Use of a smaller target size might have brought out differences better. Another limitation of this study is that hand support was unable to be fully quantified, as the actual shear forces used by the hands were not measured by an instrumented hand rail.

4.6. Conclusions

The increased incidence rate of undershooting by the PSM and increased secondary submovements are indicative of an increasingly conservative strategy used by older adults near the limits of the FBOS that may explain their slower speeds during whole body movements to maintain upright balance. These data may ultimately provide insight into mechanisms underlying fall avoidance in older adults, particularly in the backward direction.

4.7. Significance

The accuracy constrained tasks examined in this chapter involved COP movements from the anterior limit of the FBOS to the center of the foot, at a comfortable upright stance and from the anterior to the posterior limit of the FBOS. The use of the FBOS to define the target positions is clinically relevant for large body movements, such as stooping to reach for something on the floor or getting out of a chair, because it provides natural and familiar limits within which the central nervous system must make accurate, as well as rapid, movements to arrest a loss of balance.

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CHAPTER 5

AGE-RELATED CHANGES IN POSTURAL CONTROL WHEN PERFORMING DOWNWARD REACH AND PICK-UP MOVEMENTS WITH A LIMITED BASE OF SUPPORT

5.1. Abstract

Stooping, crouching, or kneeling difficulty is prevalent among older adults but few studies have explored its underlying mechanisms. This study assessed the effect of age on the postural control and losses of balance when performing downward reach and pick-up movements with a limited base of support. Healthy young (n=13) and older (n=12) women reached as fast as possible to a target placed at their maximal forward reaching distance along the floor, while standing on their whole foot (i.e., full base of support [BOS]) or forefoot (i.e., forefoot BOS). Older women required a 50% larger BOS when stooping down to touch their toes, and had a 25% decrease in their maximal forward reaching distance along the floor. Older women were twice as likely to lose their balance as young women while performing rapid downward reach and pick-up movements ($\chi^2(2) = 3.9$, $P < .05$, relative risk = 1.91, 95% CI = 0.99 - 3.72). Regarding COP excursions and COP RMS error, compared to the young, older women swayed similarly anteriorly, but posteriorly, sway was decreased in full BOS and increased in

forefoot BOS (age x BOS condition, P < .01). We conclude that even healthy older women, when reliant upon their forefoot for balance in a downward reaching task, demonstrate poorer performance than young women by requiring a longer base of support, swaying more, losing their balance more often and having a decreased reaching distance.

5.2. Introduction

Difficulty bending down to pick up an object from the floor is associated with increased fall risk in older adults (O'Loughlin et al., 1993). Stooping, crouching, or kneeling difficulty is prevalent among older adults but few studies explore the mechanisms underlying downward reaching and pick-up difficulty (Puniello et al., 2000, Puniello et al., 2001, Kuo et al., 2011). Rapid and accurate control of the center of pressure (COP) is expected for maintaining balance. Moreover, older adults with stooping, crouching, or kneeling difficulty have lower extremity dysfunction, particularly in decreased lower extremity torque capacities (Hernandez et al., 2008, Hernandez et al., 2010).

The strategies used by older adults to maintain balance while performing common daily activities with large ranges of motion at the trunk or hip (e.g., rising from a chair or ascending and descending stairs) have been examined in previous studies (Hughes et al., 1994, Lee & Chou, 2007). During lifting, posture and movement control must be simultaneously coordinated to account for pending interactions with the environment (Toussaint et al., 1995, Toussaint et al., 1997, Commissaris et al., 2001). Multi-segmental movements involve control of the angular momentum of the entire body through appropriately directed ground reaction forces (Toussaint et al., 1995).

Anticipatory postural adjustments, specified in advance of a lifting movement, form an integral part of movement control (Toussaint et al., 1997). Thus, evaluation of COP control strategies may provide some insight into age-related changes in downward reaching and lifting (e.g., upward recovery) movements.

Spatial and temporal COP measures have been previously used to evaluate postural control. Measures such as COP sway magnitude, root mean square error, and virtual time-to-contact (VTC) represent complementary aspects of COP control characteristics. The VTC represents a measure of the temporal proximity of the COP trajectory, at any given moment, with the two-dimensional base of support boundary (Slobounov et al., 1997). In upright stance, there are age-related differences in postural control responses to induced sway on balance-demanding surfaces (Maki et al., 1990, Pykko et al., 1990). Because VTC is dependent on age and balance demands (Slobounov et al., 1998), VTC can provide insight into age-related postural control changes.

This experiment was designed to assess the effect of healthy aging on downward reaching and pick-up performance. An important experimental parameter was to limit the length of the base of support by having participants stand on their forefoot. This configuration ultimately simulates typical downward reach and lean forward tasks where the heel lifts off the ground and also limits leg torque output, thereby simulating the decreased lower extremity strength that might be seen in older adults with stooping, crouching, or kneeling difficulty. Body configuration limitations when reaching down to targets at the reaching envelope were used to elicit faster and more frequent compensatory COP movements in young and older women, including the possibility of loss of balance. This study should provide insight into how COP movement speed and distal torque amplitude affect balance control strategies in young and older women.

In this study, we tested the following hypotheses: First, (H1) in comparison to young women, healthy older women will exhibit a higher incidence of losses of balance during a downward reach pick-up. Second, (H2) decreasing the length of the base of support, from the whole foot to just the forefoot, will lead to a disproportionate decrease in COP control in older women when compared to young, as evaluated by decreased virtual time-to-contact and increased COP excursion, postural sway, movement time and number of submovements.

5.3. Methods

5.3.1. Participants

Healthy young (mean ± standard deviation [SD], age 23 ± 3, n = 13) and healthy older women (age 76 ± 6, n = 12) were recruited from the local community (**Table 5.1**). All young women completed a medical history questionnaire and older women were physically screened by a nurse practitioner, so as to exclude those with significant musculoskeletal or neurological findings. The two groups were of similar weight and foot length, but older women were shorter, than young women (159 ± 5 vs. 164 ± 6 cm). Participants wore standardized canvas shoes during all testing to control for the friction coefficients within tests. All participants provided written informed consent as approved by University of Michigan Medical School Institutional Review Board procedures.

5.3.2. Instrumentation

Participants stood on a single ground-level six-channel force plate (OR6-7-1000, AMTI, Watertown, MA) with data collected at a sampling rate of 100 Hz. Kinematic data were collected using a three camera, three-dimensional, motion capture system (two Optotrak 3020 and one Optotrak Certus Camera, Northern Digital, Inc., Waterloo, Canada). Infrared light emitting diodes were placed on the right leg over the lateral malleolus, heel, fifth metatarsalphalangeal joint, femoral epicondyle; greater trochanter; over the right acromion; on the right arm over the humeral lateral epicondyle; ulnar styloid process; third metacarpalphalangeal joint; and over the nail of the middle finger. On the left hand side of the body, markers were placed over the left medial malleolus, heel, first metarsalphalangeal joint, left acromion and over the nail of the middle finger. In addition, a set of three technical markers were placed over the middle of the left thigh and left forearm to use in estimating joint movements during experimental trials. Kinematic data were sampled at 25 Hz. All data were recorded using the Optotrak system and First Principles software (Northern Digital, Inc., Waterloo, Canada). Isometric peak torque of the knee extensors and ankle plantar flexors and dorsiflexors was evaluated using a Biodex System 3 isokinetic dynamometer (Biodex Medical Systems, Inc., Shirley, NY) Torque data was sampled at a rate of 100 Hz. In addition, passive repositioning errors of the ankle joint were measured using the Biodex System 3.

Table 5.1: Mean ± SD subject characteristics

	Young Women (n=13)	Older Women (n=12)
Mean age (years)[b]	23 ± 3	76 ± 6
Height (cm)[a]	164 ± 6	159 ± 5
Weight (kg)	63 ± 11	63 ± 11
Body mass index (kg/m^2)	23 ± 4	25 ± 5
Foot Length (cm)	26 ± 1	25 ± 3

[a] Indicates age group effect P < .05.
[b] Indicates age group effect P < .005.

5.3.3. Protocol

This experiment was part of a larger study and consisted of two test sessions. On the first visit to the laboratory, the testing session included calibration trials to assess participants' maximal forward reaching distance on the floor, as well as their minimal base of support at their toes. Participants provided a series of self-reported measures, which included their activities-specific balance confidence (Powell & Myers, 1995), self-reported downward reaching performance, and the significance of positive and negative consequences during a downward reach pick-up task, given their potential impact on physical performance (Bandura, 1986). On the second test session, participants performed symmetric two-handed downward reaches to a target placed on the floor. All experimental trials were normalized to the maximal forward reaching distance on the floor, which was defined as the distance between the anterior edge of the base of support (BOS) at an upright stance, and the most anterior position of the fingertip at the floor level (**Figure 5.1**). Experimental trials were also performed at each participant's forefoot BOS. To establish the forefoot BOS, participants first bent down to touch their toes on a full BOS, with as little knee flexion as possible. Participants then moved posteriorly in quarter inch (.635 cm) increments until they were unable or unwilling to perform the maximal toe reach. The forefoot BOS was then defined as the minimal distance between the toes and the posterior edge of the platform during a successful maximal toe reach (**Figure 5.2**).

Practice downward reach trials were first performed at a comfortable speed, and then followed by trials at a fast speed. Participants were instructed to move 'as fast as possible' toward and from a target positioned at the maximal forward reaching distance on floor (mean values [SD] shown in **Figure 5.2**) at a pseudo-randomized base of support condition. Three trials were performed at either full BOS or forefoot BOS.

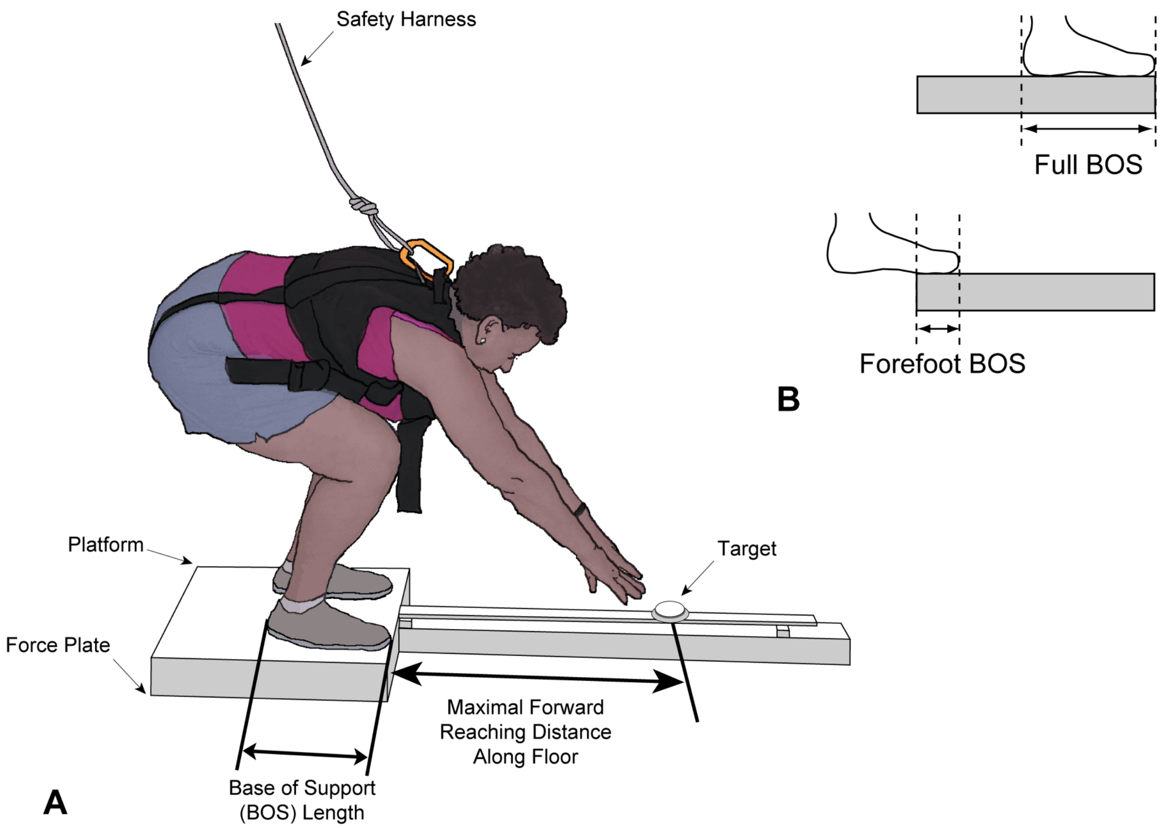

Figure 5.1: A) Illustration of symmetric two-handed downward reaches to a target placed on the floor under challenging reach conditions. B) Full base of support (BOS) and forefoot BOS conditions used for experiments.

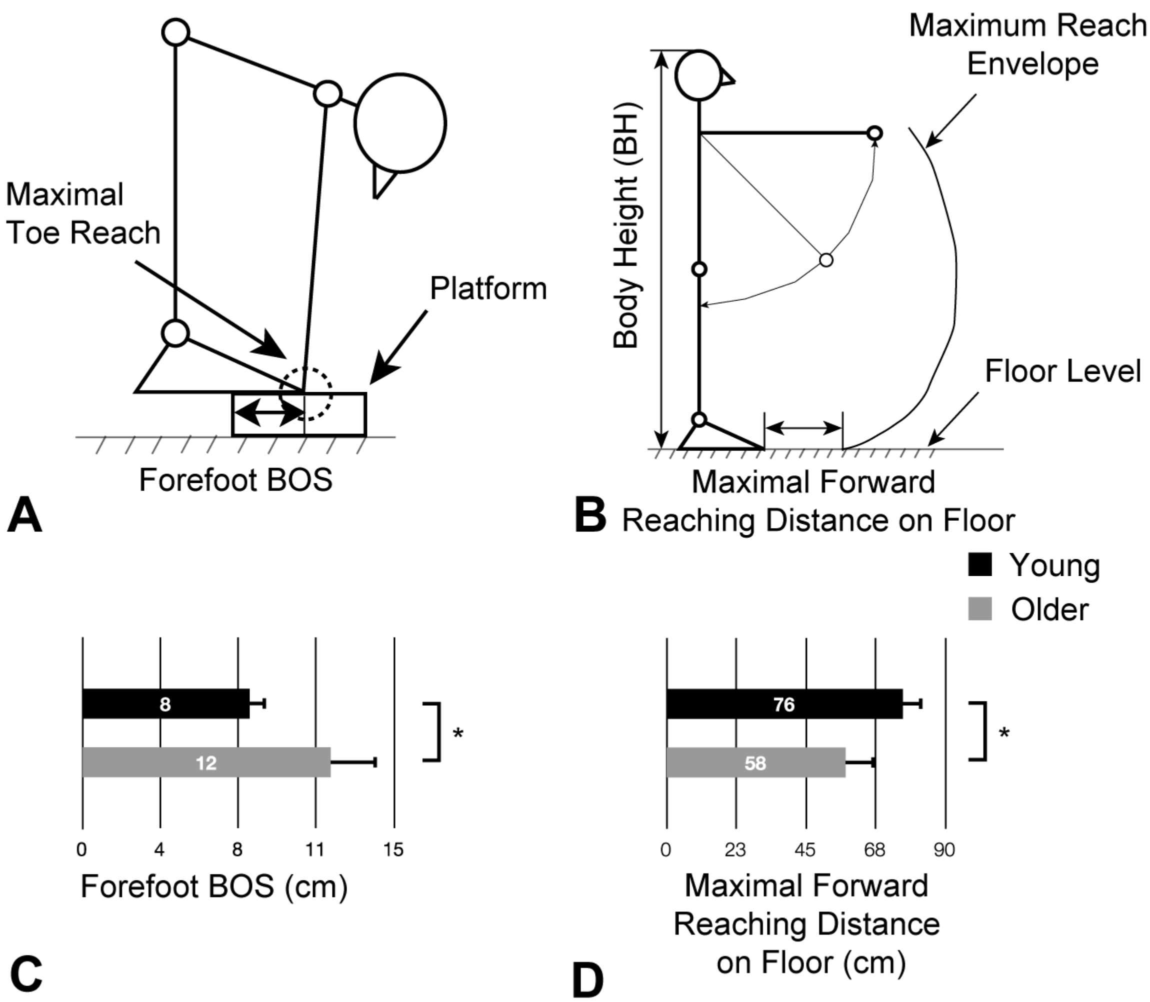

Figure 5.2: A) Illustration of forefoot base of support (BOS) as defined by the minimal distance between the toes and the posterior edge of the platform during a successful maximal toe reach. B) Maximal forward reaching distance on floor, defined as the distance between the anterior edge of the base of support at an upright stance and the most anterior position of fingertips at the floor level. C) Mean (SD) values of forefoot BOS and D) maximal forward reaching distance on floor. Test results indicated by * (P < .01).

5.3.4. Data processing and analysis

Custom Matlab (v7.4, Natick, MA) data processing software routines were used to process the data. Raw force plate data were processed with a 4th order, zero-lag, low-pass Butterworth filter with a 10-Hz cutoff frequency. COP velocity and acceleration were calculated using a five-point finite difference derivative algorithm. Using an automated procedure, downward reaching and upward recovery movements were calculated for each trial. Marker velocity profiles were used to identify the peak velocity in the downward reach and upward recovery movements. The onset of downward reaching movement was defined by having the software algorithm trace backward from the sample with the peak velocity to locate the first sample at which the velocity exceeded 10% of the maximum value (V_{max}), within the starting zone (Teasdale et al., 1993). The end of the upward recovery movement was similarly found by tracing forward from the sample with the peak velocity to identify the first sample less than or equal to 10% of V_{max}. The transition from the downward reach to the upward recovery movement was identified by the change in overall marker velocity nearest in time to the maximal forward distance, or minimal vertical height of the selected marker.

To determine whether the effect of age on the COP control of downward reaching movements becomes more pronounced in tasks with a limited base of support, we examined the incidence of losses of balance, virtual time-to-contact, COP excursion, movement time and number of COP submovements. Analysis of joint motion using optoelectronic cameras provided an assessment of the mean number of losses of balance, defined by the execution of a change in base of support strategy, namely a stepping maneuver, occurring during the downward reach or upward recovery phase of movement.

The virtual time-to-contact (VTC) was determined by calculating the estimated time for each sample point in the center of pressure trajectory to make contact with the two-dimensional boundary defined by the border of the base of support created by the feet and the elevated platform (**Figure 5.3**), based on the velocity and acceleration of the COP sample point. Estimated virtual time-to-contact times without a solution were considered to be infinite and were disregarded in calculation of the average VTC. The overall percentage of the movement spent with an infinite VTC was also calculated to account for stable COP movements. COP excursion was defined by subtracting the mean COP position in a movement from the actual COP position at a given time, so as to use for calculating the minimum posterior, maximum anterior, and mean root mean square error. Movement time was defined by measuring the elapsed time from the onset to the offset of a downward reach or upward recovery movement. The number of submovements was defined by the pairs of zero crossings in the trajectory of COP acceleration after V_{max}, as defined previously in the methods for Chapter 3.

5.3.5. Statistical analyses

All statistical analyses were carried out in SPSS 16.0 for Windows (SPSS Inc., Chicago, IL). Independent sample t-tests and chi square tests were performed to assess age differences in self-report measures, functional reaching and peak torque performance and rate of losses of balance during downward reach and pick-up movements. Linear mixed models using a restricted maximum likelihood method were used to examine the effect of age (i.e., young vs. old), BOS condition (i.e., full vs. forefoot BOS), and movement phase (i.e., downward reach or upward recovery) on outcome measures of postural control. BOS condition and movement phase were identified as repeated effects assuming a first-order autoregressive covariance structure. $P < .05$ was used for statistical significance.

5.4. Results

In comparison to young women, older women participating in this study reported no statistically significant differences in passive ankle repositioning accuracy, balance confidence, self-reported downward reaching and pick-up performance, or significance placed on positive and negative consequences (**Table 5.2**). Older women, compared to the young, had lower extremity peak isometric torque measures (e.g., knee extensor, ankle dorsiflexor, and ankle plantarflexor) and most rate of torque development measures (e.g., knee extensor and ankle plantarflexor, **Table 5.2**). No significant differences were seen in ankle dorsiflexor rate of torque development or in their normalized functional base of support length.

In terms of the downward reach performance, older women demonstrated nearly a 25% decrease in their maximal reaching distance along the floor in comparison to young women. Older women also required a 50% increase in their base of support (BOS), in comparison to young women, to successfully bend down to touch their toes (i.e., forefoot BOS, **Figure 5.2)**. Even after normalizing for body height and foot length, older women demonstrated significant decreases in their maximal downward reaching distance along the floor and significant increases in their forefoot BOS ($P < .005$, **Table 5.2**).

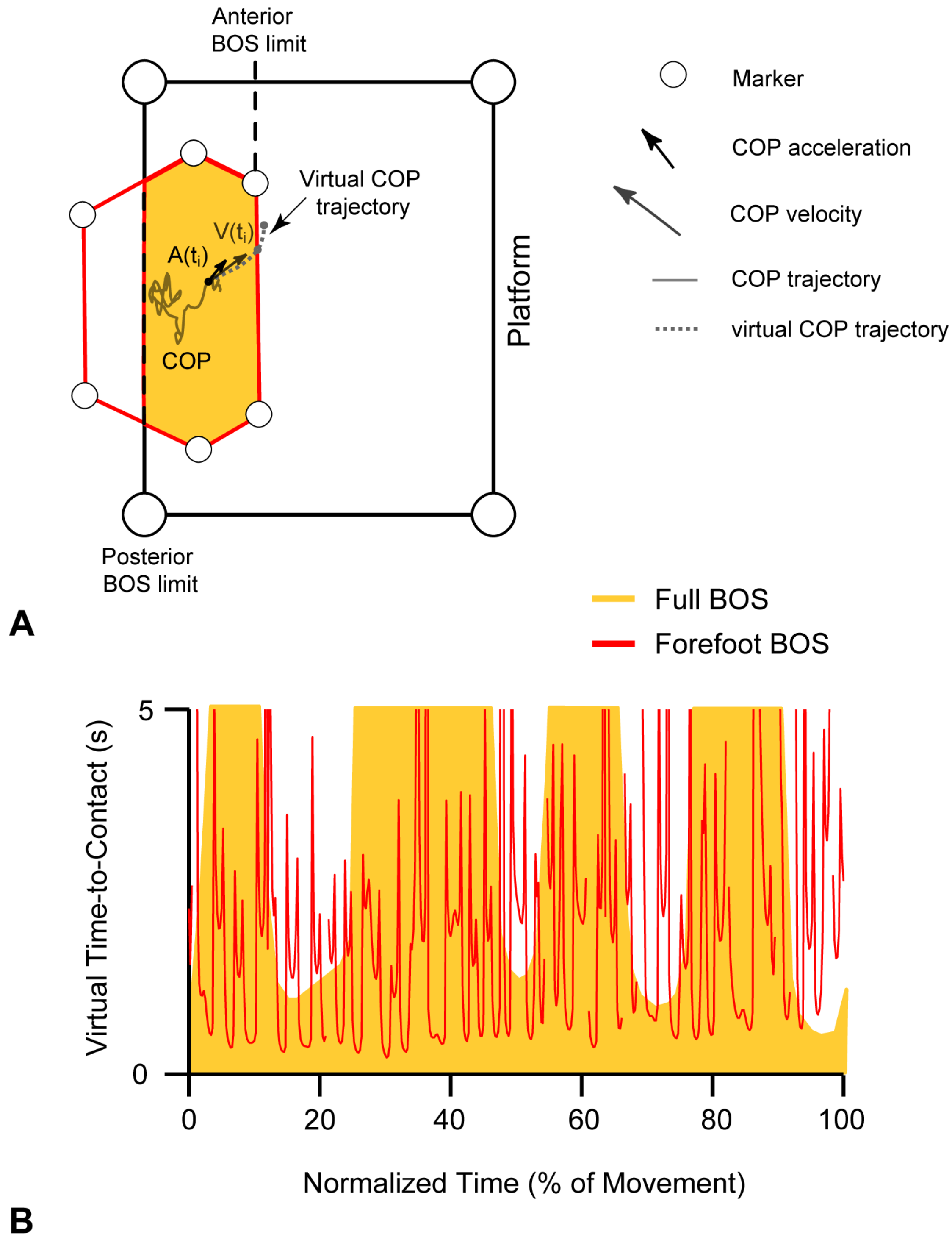

Figure 5.3: A) Schematic of virtual time-to-contact (VTC) calculation to instantaneous geometric boundary of the base of support (BOS). B) Exemplary VTC data while standing on either a full or forefoot BOS.

5.4.1. Incidence rate of losses of balance

Overall incidence rates of losses of balance during fast movements increased from 16.5% to 30.7% in older women, when compared to young women. Chi-square tests suggest that older women lose their balance at higher rates than young women ($\chi^2(2) = 3.9$, $P < .05$), with a relative risk of 1.91 (95% CI = 0.99 - 3.72).

Table 5.2: Mean ± SD subject capacities and performance

	Young Women	Older Women
Physical Capacity		
Knee Extensor Strength (% BH*BW)[a]	11 ± 4	7 ± 2
Knee Extensor RTD (% BH*BW/s)[a]	33 ± 16	20 ± 10
Ankle PF Strength (% BH*BW)[b]	6 ± 2	4 ± 2
Ankle PF RTD (% BH*BW/s)[b]	9 ± 4	4 ± 2
Ankle DF Strength (% BH*BW)[b]	2 ± 1	1 ± 1
Ankle DF RTD (% BH*BW/s)	7 ± 3	5 ± 5
Ankle Repositioning Error (deg)	3 ± 2	4 ± 2
Self-Reported Measures		
Activities-Specific Balance Confidence (0-100 scale)	97 ± 2	96 ± 4
DRPU Performance (0-100 scale)	72 ± 11	74 ± 15
Positive Consequences (0-100 scale)	41 ± 16	54 ± 24
Negative Consequences (0-100 scale)	31 ± 11	36 ± 18
Functional Performance		
Maximal Forward Reaching Distance on Floor (% BH)[b]	46 ± 3	36 ± 5
Minimal Toe Base of Support (% FL)[b]	31 ± 3	46 ± 8
Functional base of support (% FL)	77 ± 4	70 ± 14

Notes: BH = Body Height; BW = Body Weight; FL = Foot Length; RTD = Rate of Torque Development; PF = plantarflexor; DF = dorsiflexor; DRPU = Downward Reach Pick-Up.

[a] Indicates age group effect P < .05.

[b] Indicates age group effect P < .005.

5.4.2. Virtual time-to-contact

An exemplary virtual time-to-contact (VTC) calculation for a participant during both a full and a limited base of support condition is illustrated in **Figure 5.3**. A linear mixed model analysis demonstrated no significant differences due to age, base of support condition, or movement phase on VTC. Considering the percentage of movement with an infinite VTC, a linear mixed model analysis revealed significant decreases in the percentage of downward reach and pick-up movements with an infinite VTC when going from a full to a minimal toe BOS condition, $F(1,79) = 14.2$, $P < .001$ (**Figure 5.4**). A disproportionate decrease in the percentage of trials with an infinite VTC was seen during the downward reaching phase when participants were standing on their minimal toe BOS, $F(1,53) = 4.2$, $P < .05$. No other significant effects were found.

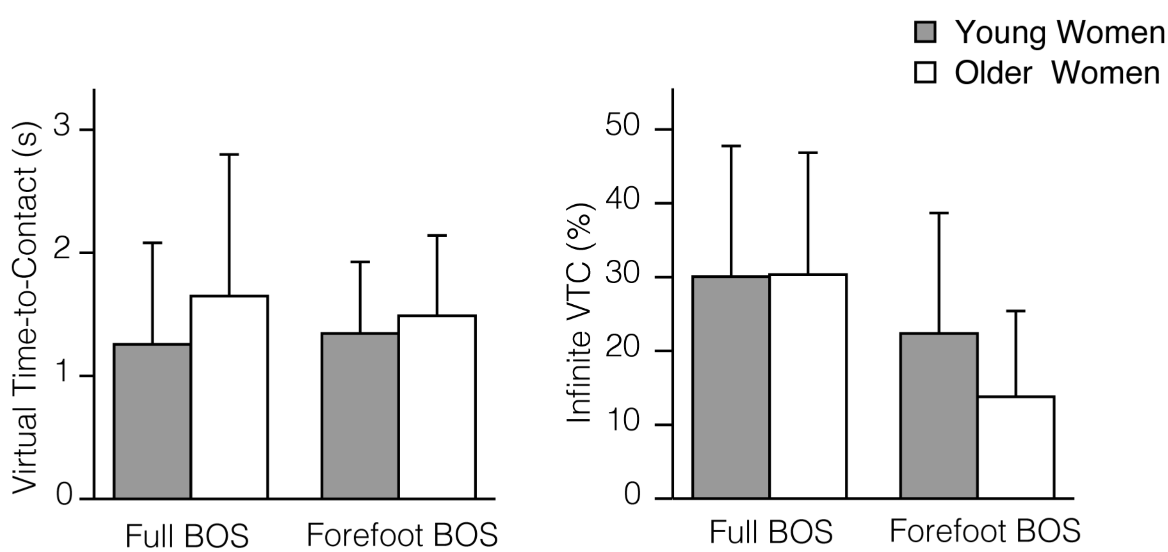

Figure 5.4: Left) Mean (SD) VTC values. Right) Mean (SD) percentage of movement with an infinite VTC value in full and forefoot base of support (BOS) conditions.

5.4.3. COP excursion

COP excursion was quantified through the measure of maximum anterior and minimum posterior COP excursion and the COP root mean square (RMS) error during downward reach and pick-up movements. The maximum anterior COP excursion significantly increased in full BOS trials in comparison to forefoot BOS trials, $F(1,41) = 17.9$, $P < .001$, (**Figure 5.5**) and during the downward reaching versus upward recovery phase, $F(1,30)= 8.5$, $P = .007$. The minimum COP excursion was smaller in forefoot BOS trials, in comparison to full BOS trials, $F(1,24) = 190.4$, $P < .001$. A significant interaction between age and BOS condition was found, $F(1,24) = 8.6$, $P < .01$, as well as a significant interaction between movement phase and BOS condition, $F(1,50) = 9.5$, $P < .005$. Considering the COP RMS error, significant decreases were observed in forefoot BOS trials versus full BOS trials, $F(1,36) = 112.5$, $P < .001$. Furthermore, an interaction between age and BOS condition was observed, $F(1,36) = 6.4$, $P < .05$. No other significant differences were observed.

5.4.4. Movement time and COP submovements

No significant differences were observed in either movement time or in the number of submovements in successful downward reach and pick-up movements due to age, BOS condition, or movement phase (**Figure 5.6)**. Trends towards an interaction between the BOS condition and movement phase were seen in movement time, $F(1,56) = 4.3$, $P < .05$, and number of submovements, $F(1,57) = 3.9$, $P < .10$. Trends towards an interaction between age and BOS condition were also seen in the number of submovements, F[(1,34) = 3.9, $P < .10$. During successful trials, a significant correlation between movement time and number of submovements was observed during downward reaching movements (Pearson correlation coefficient, r = .858, P < .01) but not in upward recovery movements (P > .05, **Figure 5.7**).

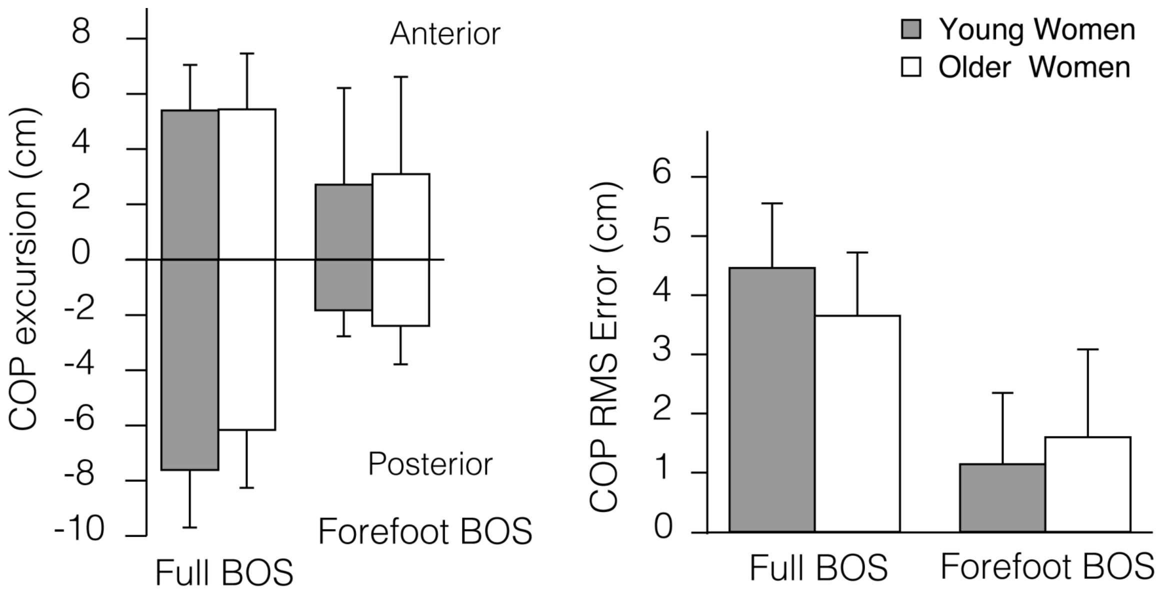

Figure 5.5: **Left) Maximum anterior or minimum posterior COP excursion (Mean [SD]) in full or forefoot base of support (BOS) conditions. Right) Mean (SD) COP root mean square (RMS) error demonstrating interaction effects between age and base of support condition.**

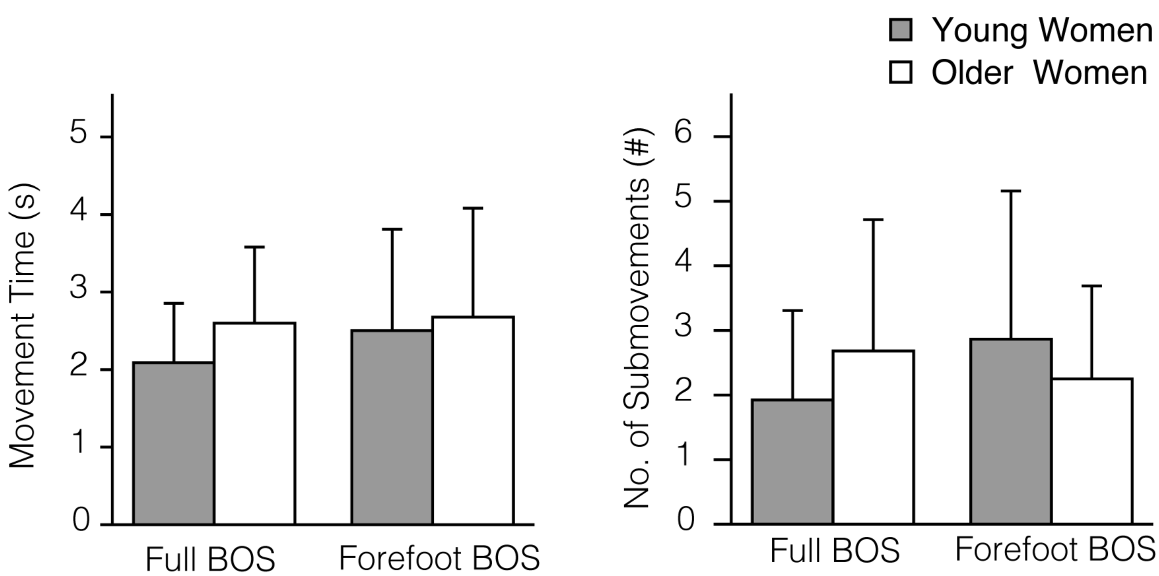

Figure 5.6: **Left) Mean (SD) movement times. Right) Number of submovements in full and forefoot base of support (BOS) conditions.**

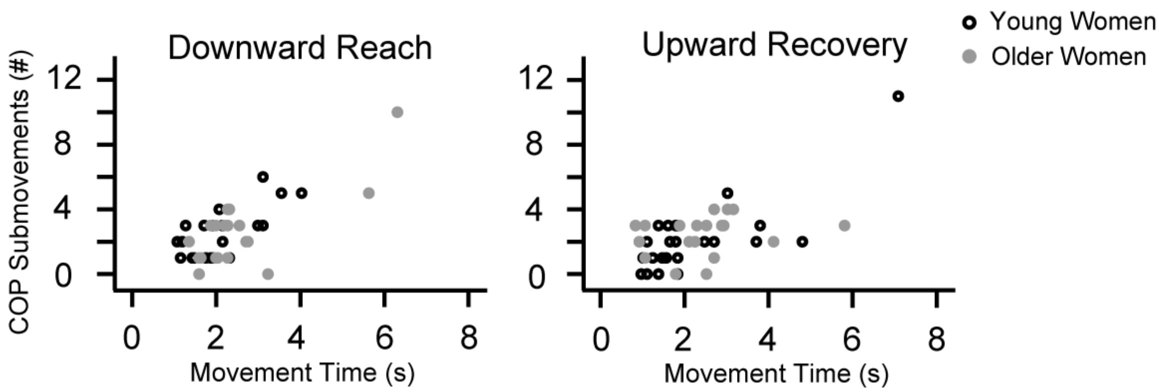

Figure 5.7: Correlation between COP submovements and movement time during (Left) downward reach and (Right) upward recovery phases.

5.5. Discussion

As hypothesized, older women were found to have nearly twice the risk of losing their balance than young women when reaching down to a target at the maximal forward reaching distance on the floor and returning to an upright stance. In addition, consistent with previous reaching studies (Duncan et al., 1990, Kozak et al., 2003, Row & Cavanagh, 2007), older women demonstrated nearly a 25% decrease in their maximal reaching distances in comparison to young, and required a 50% increase in their base of support (BOS) length to successfully bend down to touch their toes, as evaluated by their minimal toe BOS. Secondly, decreasing the length of the base of support, from the whole foot to just the forefoot, led to a disproportionate decrease in COP control in older women when compared to young, as evaluated by an increased posterior COP excursion and COP RMS. In this study, we found no significant changes in the virtual-time-to-contact (VTC), in contrast to previous studies (Haddad et al., 2010, Sloubonov et al., 1998, van Wegen et al., 2002), which may be due to the current study of a dynamic task with significant constraints on body configuration and stability.

5.5.1. Age-Related Changes in Incidence of Losses of Balance

Older women struggled to maintain their balance when reaching to a target placed at their maximal forward reaching distance along the floor while standing on their forefoot. The use of a parallel stance and symmetrical lifting movements in this study may have induced a complex postural control strategy, where the COP and COM have to be carefully coordinated to perform counter movements, contrary to a single inverted pendulum model (Kollmitzer et al., 2002). Even though no significant differences in self-reported measures of downward reach and pick-up performance were observed between young and older women, the use of a limited BOS constituted a postural threat that may have led to the use of a stiffening strategy (Carpenter et al., 2006), and may help explain age-related increases in losses of balance.

Older adults were found to have no significant impairments in passive repositioning ankle sense, which would have been detrimental in participants' performance of an appropriate postural control strategy (Horak et al., 1990, Riley et al., 1995, van der Kooij et al., 2001). However, sub-clinical sensory capacity deficits may be sufficient to put an older adult at a higher risk for a loss of balance, as dynamic sensory reweighing must occur to maintain adequate postural control (Day et al., 2002, Peterka & Loughlin, 2004). The postural threat provided by the use of a limited BOS may have similarly led to increased muscle spindle sensitivity in all participants (Davis et al., 2011). Given that the modulation of afferent feedback is reduced with advancing age (Baudry et al., 2010, Klass et al., 2011), the increased postural sway seen in older adults during the use of a limited BOS may have been required for eliciting sufficient somatosensory information (Thompson et al., 2011, Trimble et al., 1998), but also have led to the higher incidence of losses of balance.

Despite using a larger BOS and reaching to closer targets, healthy older women still had a greater number of losses of balance than young women. The decreases in knee extensor and ankle plantarflexor and dorsiflexor strength seen in older women, when compared to young, may have also led to the increased incidence rate of losses of balance. Reduced lower extremity strength is associated with self-reported stooping, crouching, or kneeling difficulty (Hernandez et al., 2008, 2010), and leg strength has also been associated with functional performance (Brown et al., 1995, Bean et al., 2002). Older adults require a greater relative amount of effort to perform functional tasks, in comparison to young adults (Hortobagyi et al., 2003, Kuo et al., 2011, Madhavan et al., 2009). Thus, the limited strength reserves available to older women, in this dynamic and balance demanding task, likely contributed to their increased incidence of losses of balance in comparison to young women.

5.5.2. Postural Control of Downward Reach and Pick-Up Movements

Decreasing the BOS from the entire foot to just their forefoot significantly decreased COP excursion and postural sway (i.e., COP RMS error) in both young and older women, consistent with previous findings (Aruin et al. 1998, Fautrelle et. al., 2010, Nouillot et al. 1992, Yiou et al. 2007). Compared to the young, older women used similar anterior COP excursions but tended to decrease posterior COP excursion with the full BOS and increase COP excursion with forefoot support (i.e., the interaction of age x BOS condition), suggesting some limitations in how far posterior COP excursions were allowed in a full BOS, and limitations in how tightly posterior COP excursions could be constrained in a forefoot BOS by older women.

Anteroposterior postural sway, as evaluated by the minimum COP excursion also demonstrated a significant interaction between BOS condition and movement phase. The contrast between the downward reach and upward recovery phase may arise from the use of 'forefoot BOS' conditions, which allow for the normal use of toe flexor musculature when moving anteriorly, but are limited when moving posteriorly, as the heels are not in contact with the raised platform. The constraints on posterior COP movements may be responsible for the similarity in COP control strategies between young and older women, as the available postural control strategies to reach down to a target and return to an upright stance may be highly constrained.

Contrary to previous studies of postural sway (Haddad et al., 2010, Sloubonov et al., 1998, van Wegen et al., 2002), older women and more challenging postural conditions did not demonstrate significant decreases in VTC. Coupled with the similar percentage of movements with an infinite VTC between young and older women, healthy aging appears to lead to the use of similar strategies of COP control during rapid whole body movements with large ranges of truncal motion. As diminished plantar cutaneous sensation and the presence of chronic ankle instability can negatively affect the VTC (McKeon and Hertel, 2007, McKeon and Hertel, 2008), healthy older women in this study may have possessed similar structural and somatosensory capacities to younger women in their lower extremities, as evidenced by a similar magnitude of passive ankle repositioning error.

Considering the percentage of a downward reach and pick-up movements with an infinite VTC, we found a significant decrease when comparing full BOS to forefoot BOS trials, consistent with prior work (Sloubonov et al., 1998). Furthermore, the disproportionate decrease in the percentage of trials with an infinite VTC seen during the downward reaching phase when participants were standing on their forefoot, suggests that configuration demands by themselves might not have been enough to identify age-related changes in VTC characteristics in this task.

5.5.3. Balance versus Configuration Demands

Downward reach and pick-up movements lead to unique challenges. An adequate amount of muscle control coordination is needed to arrest movement, while an accuracy requirement exists at the end point effector in unison with postural control demands. Upward recovery movements might necessitate less control, but movement generation requires greater use of muscle rate of torque development capacities and increased muscle strength demands. These underlying differences between downward reaching and upward recovery movements help explain some of the interactions seen between the effect of BOS conditions and movement phase (i.e., downward reaching vs. upward recovery) on minimum COP excursion, movement time, and percentage of infinite VTC.

In trunk bending movements, synergistic strategies have been identified that seek to stabilize the center of gravity shift created in the anteroposterior direction by trunk bending with lower extremity movements (Alexandrov et al., 1998). These postural control movements, often termed ankle or hip strategies (Horak and Nashner, 1986), may be sufficient to maintain balance on a platform under external perturbations. However, in whole-body reaching movements requiring both configuration control (e.g., coordination of multiple joints) and stability control (e.g., the preservation of postural stability), observed anterior displacements of the body's center of mass suggest that stability control is subordinate to configuration control (Pozzo et al., 2002).

In performing the downward reaching movements in this study, participants were challenged in both aspects of postural control, as fingertips had to reach to the limits of their reaching envelope along the floor, and their base of support was limited to the shortest distance a toe reach could be performed without losing balance. Thus, limitations in the number of available postural control strategies at the relative limits of configuration and balance capacity of each participant may explain some similarities between healthy young and older women postural control performance. In addition, simplification of movement control strategies during lifting may explain the similarities between young and older women. Even though simple speed scaling has not been strictly adhered to in standing reaching tasks among young adults (Thomas et al., 2003), the postural control of downward reaching and pick-up tasks may be simplified by the use of other constraining kinetic and kinematic relationships (Thomas et al., 2005).

5.5.4. Clinical Relevance

Difficulty with performing downward reach and upward recovery movements may be indicative of an increased risk of falls among older adults (O'Loughlin et al., 1993), as picking up a slipper from the floor has been found to be a significant task in identifying fallers from non-fallers (Chiu et al., 2003). Existing clinical tests using downward reach and pick-up items have focused on sub-maximal performance, unlike forward reaching tests (Chevan et al., 2003, Duncan et al., 1990, Newton et al., 2001). Thus, this study further supports the use of maximal performance measures for downward reach and pick-up tasks, as the maximal forward reaching task along the floor was found to elicit significant age-related changes, even after accounting for body height. Older women have been found to use increased anterior margins of safety than younger women while reaching upward (Row and Cavanagh, 2007), similar to this study's finding of an increased forefoot BOS in older women versus young. The increased postural sway seen in older women when standing on a limited BOS, suggests that older adults have more difficulty than the young lifting their heels off the floor for fear they might fall. This observation is in contrast to the commonly used heel-off strategy seen in older adults when crouching and reaching to a target (Kuo et al., 2011), and supports the increased fall-risk found in older adults with downward reaching difficulty.

5.5.5. Limitations

The exclusive use of women in this study limits generalizability, as does the excellent health of the older participants. The small sample size is a limitation, as more statistically significant differences may have arisen with the inclusion of additional participants. The inclusion of additional trials would also have been beneficial to this study, to better control for intra-subject variability. The focus on postural control measures limits the scope of this study, as kinematic changes might provide further insight to the nature of age-related changes. Furthermore, trials where a loss of balance occurred were not analyzed in this study.

5.6. Conclusion

We conclude that even healthy older women, when reliant upon their forefoot for balance in a downward reaching task, demonstrate poorer performance than young women by requiring a longer base of support, swaying more, losing their balance more often and having a decreased reaching distance. Future studies should further examine kinematic and COM control differences due to aging and their movement phase dependencies in complex whole-body movements, as well as contributing factors to actual losses of balance in downward reach pick-up movements.

5.7. Significance

Functional downward reach and pick-up tasks suggest significant age-related changes in the ability to maintain balance. This study furthers the findings in Chapter 2 & Appendix A, as decreased lower extremity strength was one of the few identified differences between healthy young and older women. In contrast to leaning tasks in Chapters 3 & 4, the use of similar postural control strategies by young and older women when performing downward reaching movements suggests a constraint on the possible strategies for a successful downward reach. The correlation found between the number of COP submovements and overall movement time suggests that in leaning and downward reaching movements, COP control may play a significant role in modulating the speed of the movement.

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CHAPTER 6

THEORETICAL ANALYSIS OF FACTORS AFFECTING DYNAMIC STABILITY DURING THE MOMENTUM ARREST PHASE OF A DOWNWARD REACH AND PICK-UP TASK: SIMULATIONS WITH A FORWARD MODEL

6.1. Abstract

Falling while performing daily tasks is a significant problem among older adults. Downward reaching requires coordination, muscle control, and balance, particularly when arresting momentum in a downward reaching task. Using a double inverted pendulum model and optimization techniques, this simulation study mathematically identified the maximal horizontal linear momentum that could be arrested before the body's center of gravity breached the limits of stability. The effects of age-related and impairment-related changes in musculoskeletal factors (i.e., decreased ankle and hip peak torque and rate of torque development) on dynamic stability were investigated. Findings suggest that the rate of torque development is a significant factor in limiting the maximal momentum that can be arrested, and that the estimated changes in strength and rate of torque development due to older age or difficulty with stooping, crouching, or kneeling seen in Appendix II, can lead to 12-21% reductions in momentum. These findings suggest that a simple forward model that accounts for changes in musculoskeletal factors

may distinguish between healthy young and healthy older women with and without stooping, crouching, and kneeling difficulty.

6.2. Introduction

The control of balance while changing the body's configuration is a critical component of daily tasks such as turning, reaching, lifting, or standing up from a bed or chair. Similarly to rising from a chair, reaching down to the floor requires movements with significant truncal or knee range of motion and knee extension strength. The selection of any one criterion to quantify the control of balance while reaching down to the floor may be unfeasible given the numerous strategies available to achieve this task. Thus, one strategy to characterize the control of balance is to examine the maximal feasible whole body center of mass momenta that can be arrested safely, without exceeding the limits of stability, as defined by the boundaries of the base of support.

Among community-dwelling older adults, difficulty bending down to the ground is associated with an increased fall risk (O'Loughlin et al., 1993). Older adults with stooping, crouching, or kneeling (SCK) difficulty have been characterized by decreased lower extremity strength, in comparison to healthy older adults (Hernandez et al., 2008, Hernandez et al., 2010). In addition, a decrease in balance confidence and an increase in the incidence rate of lower extremity range of motion limitations has been observed in older adults with a history of falls and those with SCK difficulty (Hernandez et al, 2008). Thus, it is of interest to investigate if a simple biomechanical model of a downward reaching task, can distinguish among healthy young, healthy older women with and without SCK difficulty, based on solely musculoskeletal parameters.

Previous studies have examined the feasible ranges in velocity, acceleration, or torque so as to maintain balance in an upright stance (Gordon, 1995, Iqbal & Pai, 2000, Kuo, 1995, Kuo & Zajac, 1993a, Kuo & Zajac, 1993b, Nashner, et al., 1989, Patton et. al., 1999, Yang, et al., 1990, Zajac, et al., 1984). In these studies, the idea that a range of feasible movements can be influenced by the interaction between physical and task constraints is introduced. Based on an adaptation of these methods, we will examine how physical constraints in peak torque and rate of torque development, based on findings in the literature (Thelen et al., 1996, Dean et al., 2004, Hernandez et al., 2010), can affect the maximal initial horizontal center of mass (COM) velocity that can be arrested without losing balance. As traditionally defined, we confine our control of balance to movements that successfully arrest the anteroposterior position of the center of gravity (i.e., the projection of the whole body COM) within the anteroposterior boundaries of the base of support.

The use of physical constraints to estimate the maximal horizontal COM momentum that can be generated has several practical implications. First, the biomechanical model assesses the relative importance of musculoskeletal factors on the ability to arrest horizontal linear momentum, which is a critical factor in the control of balance in common transfer tasks. In addition the use of a double inverted pendulum allows for the comparison between ankle and hip joint parameters to evaluate the significance of distal versus proximal joints. Finally, the current literature has limited information on the characteristics of older adults with SCK difficulty, and thus the use of musculoskeletal parameters allows us to evaluate if any differences emerge due to healthy aging or SCK impairment.

The primary goal of this modeling study was to address two questions: (a) what is the maximal initial COM velocity that healthy young women can arrest without losing balance and how does it differ with healthy older women with and without SCK difficulty? And (b) what is the most significant factor in successfully arresting horizontal momentum during a downward reach? The model identified the most influential factors in limiting the dynamic stability of a simple biomechanical model, and compared the effect from estimated changes in musculoskeletal parameters due to age or SCK difficulty on dynamic stability.

6.3. Methods

6.3.1. Model

The model consists of a two degree-of-freedom (2-DOF) double-inverted pendulum, nonlinear, system, which includes two point masses, one on the leg and one on the trunk, and two active joint torques at the ankle and hip (**Figure 6.1**). The model was created in Matlab (Mathworks), using the equations of motion that were derived by Kane's Method in Mathematica (Wolfram Research). The biomechanical model utilized body weight and height characteristics of the 50th percentile female (Chaffin et al., 1999) that were used to scale non-dimensional segment masses, moment of inertias, and lengths, accordingly (Contini et al., 1963; Winter, 1990). The compact form of the equations of motion was expressed as:

$$M(\dot{\theta})\ddot{\theta} = V(\theta,\dot{\theta}) + G(\theta) + T,$$

where the mass matrix, velocity vector, gravity vector, and torque vector consist of functions of the segment angles and segment angular velocities. In this model, the ankle and hip segment angles were defined with respect to the normal to the ground (**Figure 6.1**). The model's symmetric mass matrix was defined as follows:

$$M = \begin{bmatrix} M_{11} & M_{12} \\ M_{21} & M_{22} \end{bmatrix}$$

$$M_{11} = \left[I_1 + I_2 + 2 * l_1 * l_{c2} * M_2 * \cos(\theta_2) + M_2 * l_1^2 + M_1 * l_{c1}^2 + M_2 * l_{c2}^2 \right]$$

$$M_{12} = \left[I_2 + l_1 * l_{c2} * M_2 * \cos(\theta_2) + M_2 * l_{c2}^2 \right]$$

$$M_{21} = M_{12}$$

$$M_{22} = \left[I_2 + M_2 * l_{c2}^2 \right],$$

where M_1 and M_2 represent the mass and I_1 and I_2 represent the moment of inertia of the leg and head, arms, and trunk. In addition, l_1 and l_2 represent segment lengths, l_{c1} and l_{c2} represent segment center of masses, and θ_1 and θ_2 represent segment angles of the leg and trunk, respectively. The velocity vector was defined as follows:

$$V = \begin{bmatrix} V_1 \\ V_2 \end{bmatrix}$$

$$V_1 = \begin{bmatrix} 2 * l_1 * l_{c2} * M_2 * \dot{\theta}_1 * \dot{\theta}_2 + \ldots \\ \left(l_1 * l_{c2} * M_2 - l_{c1} * l_{c2} * M_2 + \left(-l_1 + l_{c1} \right) * l_{c2} * M_2 \right) * \dot{\theta}_1^2 + \ldots \\ l_1 * l_{c2} * M_2 * \dot{\theta}_2^2 \end{bmatrix} * \sin\left(\theta_2\right)$$

$$V_2 = \left[\left(-l_1 * l_{c2} * M_2 + \left(-l_1 + l_{c1} \right) * l_{c2} * M_2 \right) * \dot{\theta}_1^2 \right] * \sin\left(\theta_2\right)$$

where $\dot{\theta}_1$ and $\dot{\theta}_2$ represent leg and trunk segment angular velocities. The gravity vector is defined as:

$$G = \begin{bmatrix} G_1 \\ G_2 \end{bmatrix}$$

$$G_1 = \left[-M_1 * l_{c1} * \cos(\theta_1) - M_2 * l_1 * \cos(\theta_1) - M_2 * l_{c2} * \cos(\theta_1 + \theta_2)\right] * g$$

$$G_2 = \left[-M_2 * l_{c2} * \cos(\theta_1 + \theta_2)\right] * g$$

and the torque vector is defined as:

$$T = \begin{bmatrix} T_1 \\ T_2 \end{bmatrix}$$, where $T_1 = T_{ANKLE}$ and $T_2 = T_{HIP}$.

The nominal modeling parameters are defined in **Table 6.1**.

Table 6.1: Average subject body segment parameter data.

M_1	18.133	(kg)
M_2	42.104	(kg)
L_1	0.786	(m)
L_2	0.544	(m)
C_1	0.434	(m)
C_2	0.341	(m)
I_1	1.189	($kg*m^2/s^2$)
I_2	1.743	($kg*m^2/s^2$)
M	62.1	(kg)
H	1.6	(m)

Notes: M_1 = leg mass, M_2 = head, arms, and trunk (HAT) mass, L_1 = leg length, L_2 = upper body length, C_1 = leg center of mass length, C_2 = HAT center of mass length, I_1 = leg moment of inertia, I_2 = HAT moment of inertia, M = body mass, H = body height.

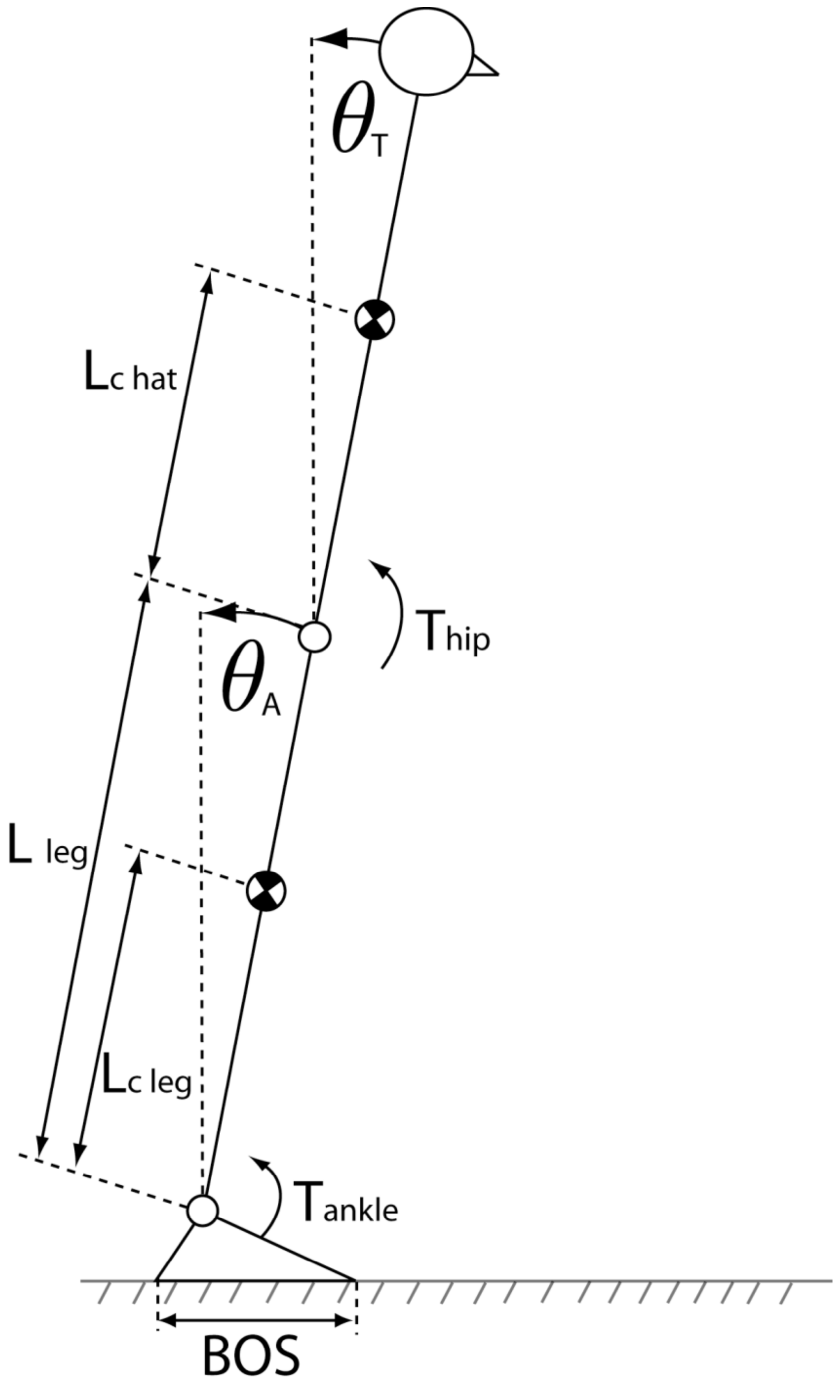

Figure 6.1: Illustration of sagitally-symmetric double-inverted pendulum with 2 degrees of freedom consisting of active ankle and hip joint torques.

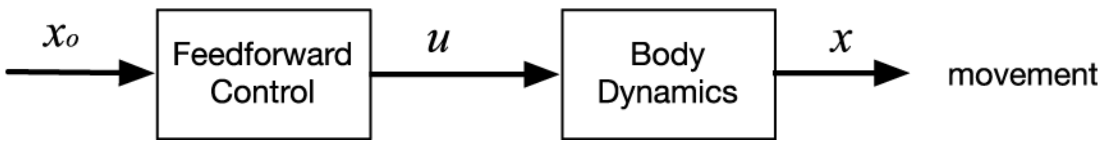

Figure 6.2: Schematic of feedforward control model used to describe the arrest of downward reaching movements.

The objective of the nonlinear optimization was to identify the maximal horizontal COM velocity, or linear horizontal momentum that could be arrested without exceeding the boundaries of the base of support. The deceleration phase of downward reaching movements was modeled in this simulation, using a simple feed-forward controller to model the arrest of momentum (**Figure 6.2**). The feed-forward controller used in the model ramped up from zero at the onset of the simulation at a fixed rate of torque development up to the physiological torque limit for each joint. The peak torque and rate of torque development for ankle and hip torque were based on previously reported values (Thelen et al., 1996; Dean et al., 2004; Hernandez et al., 2010). An estimate of rate of torque development changes due to SCK difficulty was based on the relative difference in peak isometric strength between older women with and without SCK difficulty. Using this controller, an optimal COM velocity was obtained iteratively by adjusting the initial trunk velocity, while using an initial body configuration consisting of the leg segment at a vertical position, and the trunk flexed anteriorly. The initial hip flexion angle was adjusted so as to place the vertical projection of the body's COM from directly above the ankle joint to the anterior limit of the base of support (BOS). Thus, at any given initial COM position, the boundary of feasible linear momentum that can be arrested before the COM reaches the boundary of the BOS is defined for a model of healthy young and older women with and without SCK difficulty. These simulations maximized the initial COM velocity that still permitted the body's COM to a complete rest before reaching the anterior limit of the BOS. The equations of motion of this model were integrated using a fifth order Range-Kutta method with a fourth order step-size control.

6.4. Results

6.4.1. Modeling of SCK difficulty in older women

Findings suggest that the estimated changes in strength and rate of torque development due to older age or difficulty with stooping, crouching or kneeling (SCK) difficulty can lead to 12-21% reductions in the maximal horizontal linear momentum that can be arrested (**Figure 6.3**). The three cohorts of women were modeled using age-related changes and SCK difficulty related changes in the four musculoskeletal parameters defined in **Table 6.2**. Simulated COM trajectories demonstrated how older women with and without SCK difficulty had a greater deficit in initial COM velocity under a full base of support in comparison to healthy young women, as seen in **Figure 6.4**. Furthermore, the phase plots demonstrate how simulations of increased age and SCK impairment have a slower rate of COM momentum arrest at the onset of the arrest phase of a downward reach movement.

Table 6.2: Nominal values of musculoskeletal parameters.

Variable	Symbol	Y	O	SCK Diff
Max. Ankle Torque (Nm)	T ankle	132	92	65
Max. Hip Torque (Nm)	T hip	107	73	54
Max. Ankle RTD (Nm/s)	RTD ankle	615	405	286
Max. Hip RTD (Nm/s)	RTD hip	494	322	236

Notes: Y = healthy young women, O = healthy older women, SCK diff = older women with SCK difficulty, T = torque, RTD = rate of torque development.

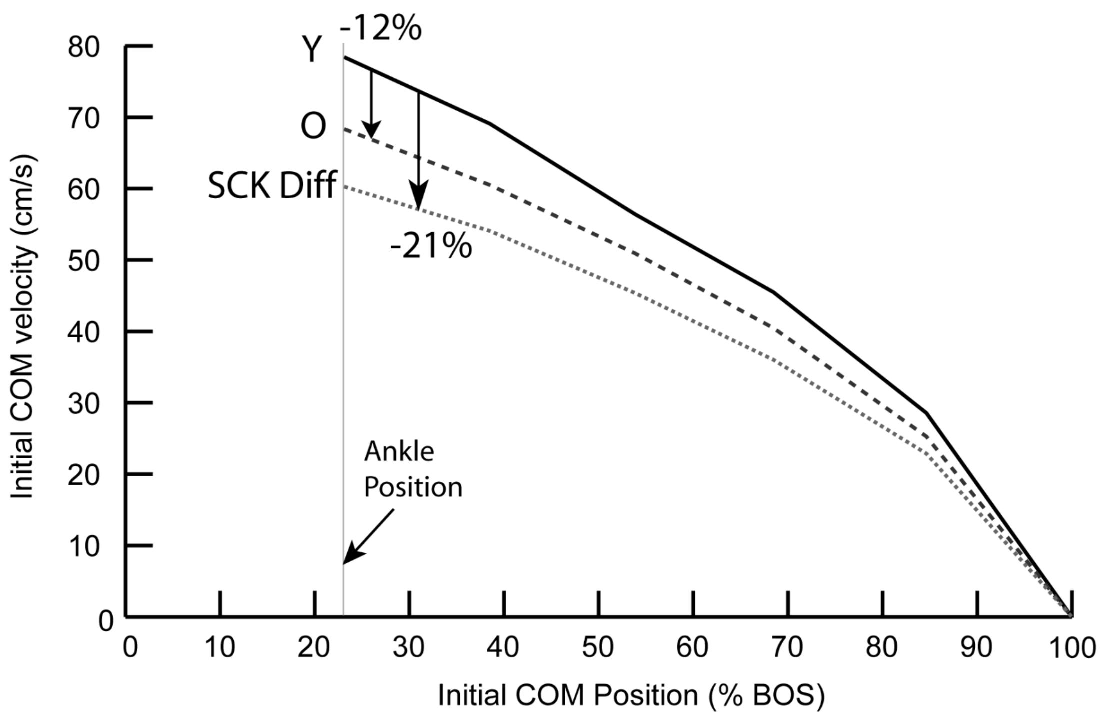

Figure 6.3: Effects of age and SCK difficulty on maximal initial COM velocities. Data showed a 12% reduction in maximal, initial COM velocities for healthy older females and a 21% reduction in older women with SCK difficulty, when compared to healthy young women.

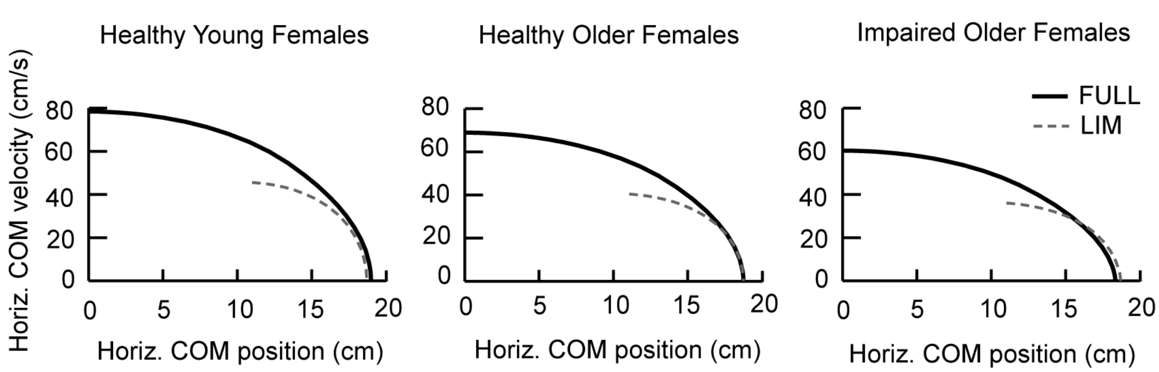

Figure 6.4: Effect of base of support length (FULL vs LIMITED[LIM]) on simulated COM phase plot of healthy young and older females and impaired older females with SCK difficulty.

6.4.2. Sensitivity Analysis

Considering fixed musculoskeletal parameters (i.e., ankle & hip peak torques of 300 Nm, and ankle & hip rates of torque development of 300 Nm/s), we evaluated the sensitivity of maximal initial COM velocity values to changes in musculoskeletal variables. As seen in **Figure 6.5**, changes in ankle and hip rate of torque development were found to have a greater effect on the maximal initial COM velocity than changes in the maximal torque capacity at the ankle or hip. Quantitatively, a 30% reduction in ankle and hip rate of torque development's nominal values, led to a 17% decrease in maximal initial COM velocity. Further reductions in ankle rate of torque development from 300 Nm/s to 100 Nm/s demonstrate a reduction in COM velocity sensitivity at smaller magnitudes of ankle rate of torque development (**Figure 6.6**).

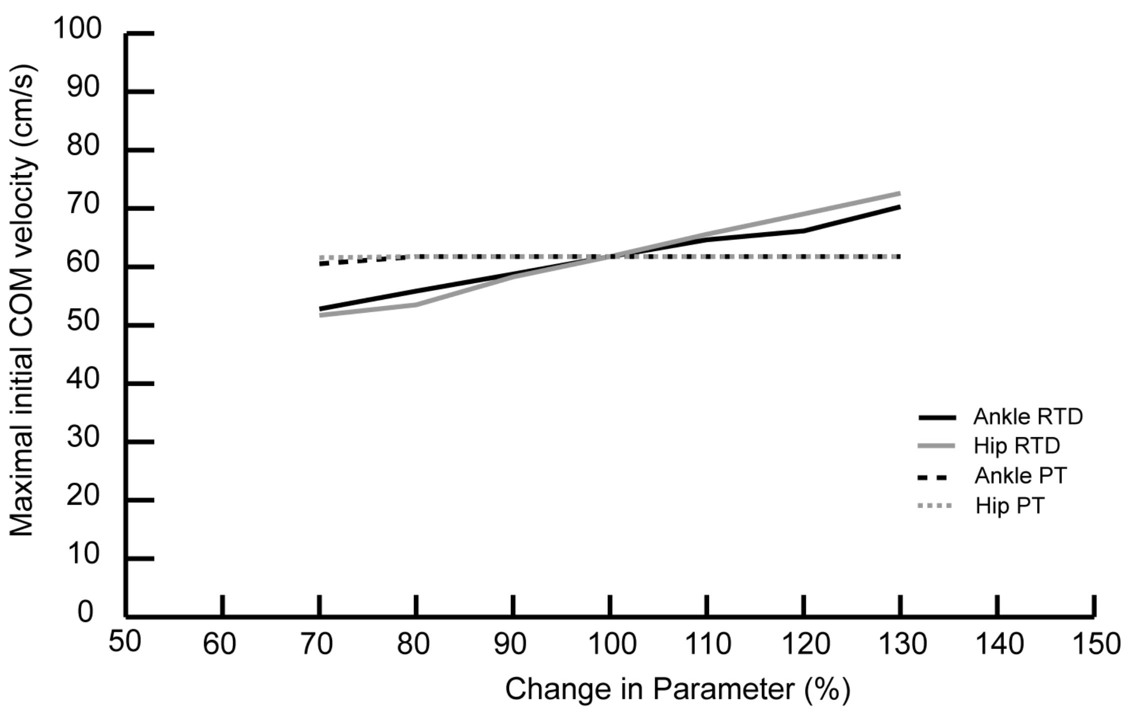

Figure 6.5: Sensitivity analysis demonstrates a greater impact of ankle and hip rate of torque development (RTD) on the maximal initial COM velocity when values are varied ± 30% from values of 300 N & 300 Nm, respectively, for peak torque and rate of torque development.

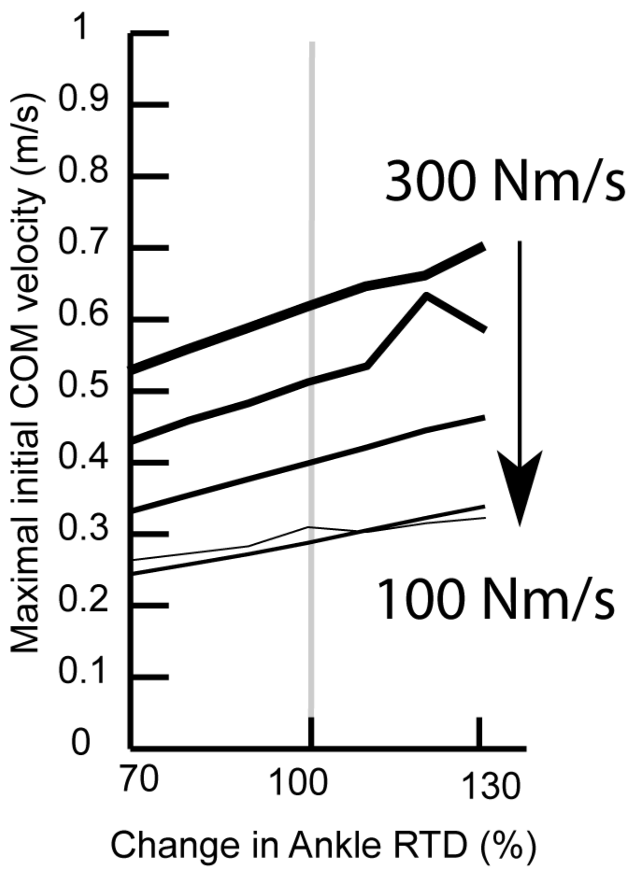

Figure 6.6: Effects of decreased ankle (RTD) rate of torque development on maximal initial COM velocities.

6.5. Discussion

This mathematical study demonstrated how a simple biomechanical model can be used to compare differences in dynamic balance capacities due to age related or impairment related changes. Based on data available in the current literature, deficits in maximal torque and rate of torque development due to healthy aging or SCK difficulty were found to lead to considerable differences in dynamic balance capacities. Similarly to previous studies (Patton et. al., 1999, Iqbal & Pai, 2000), all women can successfully arrest greater horizontal linear momentum, the farther they are from the anterior boundary of the BOS.

The importance of the rate of torque development on the ability to maintain balance has been demonstrated in simulations and experiments of forward reaching and leaning movements in older adults (Kozak, 1999, Mackey & Robinovitch, 2006, Robinovitch et al., 2002). In additional whole body movements such as a sit-to-stand, the required torques to rise from a chair were found to be well within the physiological limits of joint torque generation (Schultz et. al., 1992). Given that the maximal torque capacity is not required to perform many common functional tasks, it is apt for the rate of torque development to be a more significant factor, as a rapid response to a self-induced perturbation would be critical in maintaining balance.

The use of a feedforward controller for this simulation is supported by postural control studies in quiet stance, where negative feedback has been found to be insufficient to maintain balance (Fitzpatrick et al., 1996, Loram & Lakie, 2002). Firstly, ankle torque changes with leaning angle appear to be modulated predictively (Gatev et al., 1999, Loram et al., 2001). Secondly, paradoxical muscle movements observed in the control of an inverted pendulum, where the actuator and the pendulum move in opposite directions, suggest that the central nervous system requires a priori knowledge of postural sway (Lakie et al., 2003, Loram et al., 2009), consistent with feedforward control.

6.5.1. Distal versus proximal control

These results suggest that no significant difference occurs between distal and proximal musculature during the momentum arrest phase of a downward reach. Overall, similar characteristics were observed in the effects of either ankle or hip maximal torque or rate of torque development on the maximal initial COM velocity. This finding might be due to the limited scope of the controller, which did not optimize a control strategy for each individual joint, where changes in nominal modeling parameters might have yielded different results.

6.5.2. Limitations

The biomechanical model utilized in this study is limited by its simplicity, as only two joints (i.e., ankle and hip joints) are considered and movement is constrained to the sagittal plane. Normalized knee extensor strength has been found to be a significant predictor of SCK difficulty (Hernandez et al., 2008), but the lack of a knee joint did not allow us to evaluate the impact of the knee joint in downward reach and pick-up movements. The model utilized a simple feed-forward controller assuming that to arrest the initial momentum, the torque at each joint would need to utilize their maximal strength as rapid as possible. However, patterns of postural adjustments during induced body sway have shown that muscle activation begins in ankle and proceeds proximally in healthy adults (Badke & Duncan, 1983). Thus, further optimization on the coordination of joint torque generation might have led to additional differences between simulations of young and older women. Torque strength assumed no dependency on joint angle or angular velocity and assumed a fixed pin joint at the ankle and hip. No effects due to neuromuscular dynamics or sensory dynamics were considered, which would be expected to affect the maximal initial COM velocity. In future studies, the coordination of torque generation at each joint, the inclusion of a knee joint, limitations due to range of motion capacities, and neuromuscular noise should be explored, as well as validation in the downward reaching movements of older adults with and without SCK difficulty.

6.6. Conclusion

The use of a simple biomechanical model to compare the effects of healthy aging and SCK difficulty demonstrates how significant changes in peak torque, and particularly rate of torque development impact the maximal linear momentum that can be arrested during a downward reaching movement. This study suggests that older women, and particularly those with SCK difficulty, may benefit from going beyond traditional strength training and focusing on power generation. Training regimens focused on power generation would improve the participant's rate of torque development, and thus improve their dynamic balance capacity.

6.7. Significance

The simulations of downward reaching movements in this study provide general insight into the dynamic control of balance, and the significance of musculoskeletal parameters. The importance of the rate of torque development in arresting downward reaching movements provides additional significance to the previous studies in Chapters 3 & 4 on center of pressure control. As rapid center of pressure control would be crucial in generating the rapid torques needed to overcome self-induced or external perturbations in whole body movements.

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CHAPTER 7

GENERAL DISCUSSION

Through the study of the biomechanics of leaning and downward reaching tasks, we identified salient characteristics of older women with stooping, crouching, or kneeling (SCK) difficulty, and confirmed the importance of center of pressure (COP) submovements in the control of whole body movements. We first showed that older adults with self-reported SCK difficulty had greater impairments in lower extremity musculoskeletal function (Chapter 2). Given the significant role of lower extremity strength in predicting SCK difficulty and functional performance, we further explored the characteristics of COP control in whole body movements. Leaning movements were first examined as they represent the initial phase of all SCK movements. In sub-maximal leaning tasks resembling the initial phase of a typical forward reach, older adults exhibited slower, less efficient COP movements and a greater number of submovements, particularly when moving posteriorly (Chapter 3). We also examined small and large amplitude posterior leaning movements originating near the anterior limit of the functional base of support. In comparison to young women, older women were found to demonstrate a disproportionate decrease in the speed and accuracy of their COP primary submovements as movement amplitude increased (Chapter 4). In the final task, we investigated age-related changes in postural control when performing downward reach and pick-up movements with a limited base of support (BOS). Older women were found to require more support, sway more, reach less far, and lose their balance more often than young women (Chapter 5). A simple biomechanical model was also created that demonstrated significant differences in the capacity to arrest horizontal linear momentum during a downward reach movement due to age-related and SCK difficulty-related changes (Chapter 6). Experimental results support the hypothesis that while performing leaning and downward reaching movements towards a target, older women, compared to younger women, often exhibit slower but more frequent COP submovements in order to accomplish the task and regain the upright posture.

7.1. Factors Underlying the Control of SCK Movements

SCK movements require significant coordination, muscle control, and balance (Burgess-Limerick et al., 1995, de Looze et al., 1993, Toussaint at al., 1992, Toussaint at al., 1995). Bending down from an upright stance into crouching and kneeling involves significant ankle, knee, and hip range of motion, whereas stooping movements are characterized by a reduction of knee movement (Burgess-Limerick et al., 2001). Depending on the downward reaching strategy, constraints due to the body's configuration may compromise muscle or balance function. When kneeling, trunk strength is decreased due to a reduced capability to rotate the pelvis backwards and changes in the configuration of the leg (Gallagher, 1997). Adequate ankle dorsiflexor and plantar-flexor strength may be required to generate corrective torques about the ankle to maintain equilibrium by moving the center of mass forward or backward during stooping movements, due to the limited range of motion at the knee and hip. Consistent with limitations in trunk control, ankle plantarflexors and dorsiflexors co-activate when crouching down (Dionisio et al., 2008, Kuo et al., 2011). Furthermore, knee extensor strength would be expected to play a significant role in the recovery to an upright stance after crouching or kneeling, given the significant activity seen in thigh muscles during the upward recovery phase (Gallagher et al., 2011). Thus, recent experimental findings further validate the significance of lower extremity strength in predicting SCK difficulty and functional performance

This thesis presents evidence that decreased lower extremity strength is significantly associated with stooping, crouching, or kneeling difficulty (Chapter 2, Appendix A), consistent with previous findings of limited function in daily activities with reduced knee extension strength (Buchner et al., 1996). Thus, decreases in strength due to increased age (Harbo et al., 2011, Hurley et al, 1998) would be expected to contribute to SCK difficulty. Older adults require a greater relative amount of effort to perform functional tasks, in comparison to young adults (Hortobagyi et al., 2003, Kuo et al., 2011, Madhavan et al., 2009). Decreased segmental trunk angular kinematics may contribute to increased displacement kinematics and place the elderly at increased risk of injury and falling during loaded lifting tasks (Burgess et al., 2009). Muscle activity for the majority of thigh muscles increases as the base of support (BOS) is decreased when crouching (Gallagher et al., 2011). However, older adults have been observed to crouch in a shallow and heel-off posture during forward reaching tasks, in comparison to young adults (Kuo et al., 2011). This means that they decrease their base of support and thereby partly explains their observed increases in muscle activity (Kuo et al., 2011).

Functional impairment is expected to lead to significant changes in self-selected downward reach and pick-up strategies. Crouching produces higher loading conditions than kneeling, as indicated by higher varus and resultant moments (Pollard et al., 2011). Based on biomechanical models, spinal compression forces increase when kneeling, while shear forces increase when stooping (Gallagher et al., 1994). In adults with knee joint pain, relative to healthy controls, decreases in knee flexion and increases in ankle torque generation have been found when starting in a crouching configuration (Dionisio et al., 2011). Consistent with these findings, older adults with SCK difficulty tended to stoop more than those without SCK difficulty when lifting a box from the floor (Appendix C). Thus, older adults with SCK difficulty are likely to exhibit changes in their downward reaching strategies that best compensate for their limitations.

7.2. Speed-Accuracy Tradeoffs in COP Movements and Submovements

We found that age, movement direction, and movement amplitude significantly affect discrete COP submovement speed and accuracy. In order to achieve comparable levels of accuracy, healthy older women relied on slower volitional COP movements than young women in a wide range of targeted movements (Chapter 3 & 4). Consistent with previous findings in accuracy-constrained planar arm movements (Goggin and Meeuwsen, 1992, Ketcham et al., 2002), older women used a greater number of compensatory submovements than young women in moving their COP under conditions of similar movement amplitude (Chapter 3). Considering the stochastic optimized submovement model (Bullock and Grossberg, 1988, Crossman and Goodeve, 1983, Meyer et al., 1988), age-related changes in neuromuscular noise, motor-unit variability, or antagonist muscle coactivation (Fishbach et al., 2005, Galganski et al., 1993, Roos et al., 1997, Tracy and Enoka, 2002) would be expected to lead to slower volitional movements, an increased number of secondary submovements, and an increased ratio of peak-to-average velocity (Novak et al., 2000, Schmidt et al., 1979, Smits-Engelsman et al., 2002). Given the decreased movement speeds observed in healthy older women, in comparison to healthy young, we would expect older women to be less accurate than young women if they moved at similar speeds. This limitation poses a risk to older women as their COP movements might not be fast enough or accurate enough to successfully react and maintain balance in response to perturbations (Schulz et al., 2005, Tucker et al., 2008).

The disproportionate decreases in the speed and accuracy of COP primary submovements as movement amplitude increased seen in older women (Chapter 4) are consistent with findings older women being more cautious than young when approaching their maximum attainable reach (Feldman & Robinovitch, 2005). As proposed by Loram & Lakie, balance is achieved by the constant repetition of a neurally generated ballistic-like biphasic pattern of torque which can control both position and sway (Loram & Lakie, 2002). Thus, when older women faced greater risks of a loss of balance, they relied on more conservative strategies than young women, opting to move slower and arresting movement of their primary submovement well before the desired target that, in turn, led to a greater number of submovements (Chapter 4).

The observed decreases in volitional COP speed and accuracy in posterior movement, when compared to anterior movement is novel (Chapters 3 & 4). In contrast to previous studies of volitional COP movements (Danion et al., 1999, Duarte et al., 2005, Tucker et al., 2008, Tucker et al., 2009), our research focused primarily on discrete rather than continuous movements, which may help explain the novel findings. Discrete movements stress an adult's postural control capacity more than continuous movements, given their increased reliance on sensory feedback and decreased efficiency (Nelson, 1983, Smits-Engelsman et al., 2002). The disproportionate increase in the number of secondary corrective submovements used when older adults moved posteriorly, in contrast to young adults (Chapter 3), may have been due to decreases in foot plantar sensation and optic flow threshold sensitivity due to age (Inglis et al., 2002, McKeon and Hertel, 2007, Wade et al., 1995). The use of more frequent and variable COP movements by older adults when maintaining balance, particularly posteriorly, is undesirable, as it would present a greater risk for a loss of balance. Furthermore, findings from this thesis suggest that discrete volitional COP movements, such as those during gait initiation and termination, turning, or reaching may be a greater challenge to the postural control of older adults, when compared to continuous movements. However, part of the explanation for the observed changes in movement direction might be due to foot position, as a previous study (Appendix B) found that in healthy young adults, slower speeds are needed to maintain accurate COP movements when standing on the rearfoot, rather than when standing on the forefoot. Thus, further work remains in assessing the effect of movement direction on discrete COP movements within the limits of the FBOS.

7.3. Biomechanical Modeling

Based on experimental findings, a simple biomechanical model considering musculoskeletal parameters demonstrates significant changes in the maximal initial center of mass (COM) velocity that can be arrested without losing balance, when comparing young women and older women with and without SCK difficulty (Chapter 6). This model is consistent with findings of older people falling more often than young people after tripping due to slower development of ankle plantarflexor moments (Pijnappels et al., 2005).

Given the similarities found in the movement times of young and older women in our downward reach and pick-up study (Chapter 5), the increased incidence rate of losses of balance may be partly explained by our biomechanical model, as musculoskeletal changes due to age resulted in a decrease in the whole-body COM velocity that could safely be arrested within the limits of the BOS (Chapter 6). Thus, it would be expected that older women would need to slow down or risk falling.

Given the significant latencies observed in reflex pathways (Nashner and Berthoz, 1978), feedback control may be ineffective in prompting an adequate balance control strategy before a fall is imminent. The use of a feedforward controller for postural control is further supported by studies in quiet stance, where negative feedback has been found to be insufficient to maintain balance (Fitzpatrick et al., 1996, Loram & Lakie, 2002). Furthermore, while ankle torque changes with leaning angle appear to be modulated predictively (Gatev et al., 1999, Loram et al., 2001), paradoxical muscle movements observed in the control of an inverted pendulum, where the actuator and the pendulum move in opposite directions, suggest that the central nervous system requires a priori knowledge of postural sway (Lakie et al., 2003, Loram et al., 2009). Thus, current evidence in the literature supports the use of a feedforward controller in the modeling of postural control.

7.4. Clinical Implications

Falling while performing daily tasks such as rising from a chair or bed, stooping, or bending is a common cause of injury among community-dwelling older adults (Nevitt et al., 1991). The high correlation between self-reported SCK difficulty and clinical balance tests such as the unipedal stance test, maximum step length test, and timed up and go are evidence of the predictive power of this one item with traditional assessments of fall risk (Appendix A). Based on the specificity and sensitivity of picking up a slipper from the floor in discriminating between fallers from non-fallers, a downward reaching and pick-up movement is a complex, challenging task that provides an assessment of overall postural control (Chiu et al., 2003).

Findings from this thesis have implications for refining future clinical interventions for older adults with functional impairments. Older adults with SCK difficulty may benefit by learning task-specific strategies such as using support surfaces for assistance in SCK movements or customized whole-body movement strategies, or by using traditional rehabilitation treatments to address leg joint limitations. Practice of task-specific strategies not only leads to rapid adaptation (Grabiner et al., 2008), but is also an important component of improving self-efficacy (Bandura, 1986). When successfully performing downward reach and pick-up movements, utilizing a combination of stooping and crouching strategies, COP control in older is similar to that in young adults as long as balance is maintained (Chapter 5). Thus, improvements in the underlying contributors to SCK difficulty may lead to similar improvements in COP control capacity. Given the shared characteristics between older adults with SCK difficulty and older adults with a history of falls (Hurley et al., 1998, Tinetti et al., 1986, Tinetti et al., 2000), reducing SCK difficulty may also reduce fall risk.

7.5. Strengths

Chapter 2 provides the first in-depth investigation of mechanisms independently associated with increased SCK difficulty, going beyond previous studies focused on limited measures of muscle or back function. Chapter 3 is the first published study to describe discrete, rapid, and targeted COP movements in young and older adults. Chapter 3 provides a clinically relevant examination of the age-related changes of rapid volitional COP movements and its interactions with movement direction. To our knowledge, Chapter 4 is the first study to examine the effects of age and movement amplitude on discrete and large accuracy-constrained COP movements when leaning from an upright stance. In both Chapter 3 & 4, we present the first studies to allow for the long-term training of volitional COP movements, as three separate laboratory sessions were provided for participants to become comfortable with the accuracy-constrained leaning tasks. Chapter 5 is one of a handful of studies that has demonstrated increased losses of balance in healthy older adults, when compared to young in a functional task. The increased postural sway seen in older women when standing on a limited BOS (Chapter 5), suggests that older adults have more difficulty than the young lifting their heels off the floor for fear they might fall. Chapter 6 provides a simple biomechanical model in the effects of healthy aging and SCK difficulty demonstrate how significant changes in peak torque, and particularly rate of torque development impact the maximal linear momentum that can be arrested during a downward reaching movement. Furthermore, Appendix A is the first study to examine the differences in trunk and lower extremity muscle strength in older adults with SCK difficulty. To our knowledge, Appendix B is the first study to examine the effects of foot position and target placement on accuracy-constrained volitional COP movements. Lastly, Appendix C is one of the only studies of stooped lifting and reaching in older adults with SCK difficulty.

7.6. Limitations

This dissertation was primarily focused on a cohort of healthy young and older women (Chapter 3, 4, & 5). This limits the generalizability of our findings, to men, who might differ from women in their strategies for approaching the limits of their balance and reaching capacities.

In studies of accuracy-constrained COP movements (Chapter 3 & 4, Appendix B) it was our intent to have participants place an equal weight on accuracy and speed demands, but participants may have prioritized accuracy over speed or vice versa. Findings of decreased speed might have been assumed to be due to a change in prioritization, had we not seen similar levels of COP endpoint accuracy in both young and older adults in maximal tasks (Chapter 4). This suggests that for a similar level of accuracy, there are limits on how fast an older adult can move their COP in comparison to young adults.

All the accuracy-constrained movements were carried out in an upright position and not a crouched position (Chapter 3 & 4, Appendix B). Thus, there are limitations in extending our findings in COP control of an upright stance to downward reaching and pick-up movements, due to the changes in body configuration and available balance control strategies. However, in both tasks distal postural control should significantly contribute to balance, as the torso and upper extremity will be focused on target acquisition, while anke, knee, and hip corrective movements would be necessary to maintain balance. Given the decrease in available postural control strategies in a crouching position (as opposed to upright stance), we would expect volitional COP speeds in upright stance to provide an upper boundary for maximal capacity.

The lack of significant age-related changes in COP control strategies during downward reach and pick-up movements may have been due to the limited sample size in our study (Chapter 5). Moreover, in contrast to planar reaching studies, there were fewer trials used in the present study. Time limitations restricted the number of downward reaching movements that could be performed in each testing session.

The use of a single force platform limited our findings to a composite evaluation of postural control under both feet (Chapter 3, 4, & 5). Older adults may have significant body weight distribution asymmetry as a compensation strategy for decreased postural control in upright stance (Blaszczyk et al., 2000).

The lack of kinematic and electromyography (EMG) data is a limitation of this study, as they could have provided further measures to evaluate the strategies used by older adults to control leaning and downward reaching movements (Chapter 3, 4, & 5). As recent studies have shown, older adults crouch in a shallow and heel-off posture and utilize greater muscle activity to perform crouch-to-reach movements than younger adults (Kuo et al., 2011). Hence, balancing on the forefoot alone is not uncommon in the downward reaching task, and this makes the findings of Chapter 5 more meaningful, even if balancing on the forefoot appears at first to be unusual and artificial. Increases in the frequency of heel-toe rocking movements, an idealized task simulating activities such as reaching or bending, have been shown to not alter the minimum or maximum COP excursion, but can lead to decreases in COM excursion (Murnaghan et al., 2009).

The small number of degrees of freedom in the biomechanical study was a limitation of this thesis (Chapter 6). Normalized knee extensor strength has been found to be even more significant than normalized ankle plantarflexor or dorsiflexor strength (Hernandez et al., 2010), but the lack of a knee joint in our model did not allow us to evaluate the role of knee joint strength in downward reach and pick-up movements. Furthermore, patterns of postural adjustments during induced body sway have shown that muscle activation begins in ankle and proceeds proximally in healthy adults (Badke and Duncan, 1983). Thus, further optimization on the coordination of joint torque generation might have led to additional differences between simulations of young and older women.

The biomechanical model considered only one method to terminate movement: increased deceleration of the whole body COM through an extensor synergy in all limbs (Chapter 6). However, based on gait simulations, we could have considered deceleration due to each limb and energy or momentum transfer to dissipate remaining momentum (Oates et al., 2005). In experimental studies, increased deviations and maximal absolute range of COP movement can indicate an inability of participants to control their COP (Oates et al., 2005), yet limited contributions from COP control were investigated in our biomechanical model.

7.7. References

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CHAPTER 8

CONCLUSIONS

1. Older adults who self-report leg joint limitations, decreased balance confidence, or decreased knee extension strength are more likely to report stooping, crouching, or kneeling (SCK) difficulty (Chapter 2).
2. Despite being 27% slower, older women rely on nearly twice as frequent submovements to maintain center of pressure accuracy, particularly when moving posteriorly, thereby providing evidence of a compensatory strategy that may be used for preventing backward falls (Chapter 3).
3. Undershooting primary submovements and increased secondary submovements are indicative of an increasingly conservative strategy used by older adults near the limits of the functional base of support that may explain their slower speeds during whole body movements to maintain upright balance (Chapter 4).
4. Even healthy older women, when reliant upon their forefoot for balance in a downward reach task, demonstrate poorer performance than young women by requiring a longer base of support, swaying more, losing their balance more often and having a decreased reaching distance (Chapter 5).
5. Findings from a simple planar model suggest that a simple forward dynamic model that accounts for changes in musculoskeletal factors may distinguish

between healthy young and healthy older women with and without SCK difficulty (Chapter 6).

6. The significant correlations found between SCK difficulty and functional balance tests such as unipedal stance time, timed-up and go, and maximum step length extend the results of other investigators who found an increased risk of falls in older adults who reported trouble bending down to the floor (Appendix A).

7. In healthy young adults, slower speeds are needed to maintain accurate COP movements when standing on the rearfoot, rather than when standing on the forefoot, thereby suggesting a possible mechanism for balance deficits during stair descents (Appendix B).

CHAPTER 9

RECOMMENDATIONS FOR FUTURE WORK

1. Expand our understanding of stooping, crouching, or kneeling (SCK) impairment by examining center of pressure (COP) control changes in older women with SCK difficulty.
2. Investigate the relationship between self-reported and actual performance in SCK movements in older men.
3. Further increase the trunk lean task challenge by decreasing the COP target size so as to identify the smallest step change in volitional COP movement that older adults can reliably achieve.
4. Evaluate the effect of enhanced feedback modalities such as tactile or auditory cues in accuracy-constrained whole body SCK movements.
5. Explore age-related changes in volitional COP speed-accuracy tradeoffs by not only controlling for the accuracy of a task, but also by controlling for the speed of the task and accessing its effect on accuracy.
6. Examine the effect of age on kinematic and center of mass control differences in the first phase of SCK movements, the downward reach, versus the second phase, the subsequent recovery to an upright stance.

7. Examine whether changes to neuromuscular parameters such as sensor and motor noise or time delays significantly affect dynamic balance capacity in leaning or SCK tasks.
8. Evaluate the kinematics and muscle activation patterns of extensor and flexor musculature in the lower extremity to evaluate their contributions to the arrest of momentum in downward reach and pick-up movements.
9. Expand the biomechanical modeling to include the knee, so as to further evaluate the role of an additional degree-of-freedom in the control of downward reaching movements. Reevaluate the potential limiting biomechanical factors for older adults with SCK difficulty.

APPENDICES

APPENDIX A

DECREASED MUSCLE STRENGTH RELATES TO SELF-REPORTED STOOPING, CROUCHING, OR KNEELING DIFFICULTY IN OLDER ADULTS

A.1. Abstract

Bending down and kneeling are fundamental tasks of daily living, yet nearly a quarter of older adults report having difficulty performing or being unable to perform these movements. Older adults with stooping, crouching, or kneeling (SCK) difficulty have demonstrated an increased fall risk. Strength (force-generating capacity) measures may be useful for determining both SCK difficulty and fall risk. The purposes of this study were: (1) to examine muscle strength differences in older adults with and without SCK difficulty and (2) to examine the relative contributions of trunk and leg muscle strength to SCK difficulty. Community-dwelling older adults (age [X±SD], 75.5±6.0 years) with SCK difficulty (n=27) or without SCK difficulty (n=21) were tested for leg and trunk strength and functional mobility. Isometric strength at the trunk, hip, knee, and ankle was also normalized by body weight and height. The design of this research was a cross-sectional observational study. Compared to older adults with no SCK difficulty, those with SCK difficulty had significant decreases in normalized trunk extensor, knee

extensor, and ankle dorsiflexor and plantar-flexor strength. In 2 separate multivariate analyses, raw ankle plantar-flexor strength (odds ratio [OR] = 0.97, 95% confidence interval [CI], 0.95-0.99) and normalized knee extensor strength (OR = 0.61, 95% CI, 0.44-0.82) were significantly associated with SCK difficulty. Stooping, crouching, or kneeling difficulty also correlated with measures of functional balance and falls. Although muscle groups that were key to rising from SCK were examined, there are other muscle groups that may contribute to safe SCK performance. Decreased muscle strength, particularly when normalized for body size, predicts SCK difficulty. These data emphasize the importance of strength measurement at multiple levels in predicting self-reported functional impairment.

A.2. Introduction

Stooping, crouching (i.e., bending down), and kneeling movements are an integral component of many common activities, including gardening, shopping, and cleaning. Limitations in stooping, crouching, or kneeling (SCK) are associated with an increased likelihood of limitations in other lower- body functional tasks such as lifting and prolonged standing (Long and Pavalko, 2004) and also are associated with fall risk (O'Loughlin et al., 1993). As with standing up from a chair, SCK movements require coordination of the whole-body center of mass over a wide range of postures in order to prevent a loss of balance or fall. Moving from stance into crouching and kneeling involves significant ankle, knee, and hip range of motion, whereas stooping movements are characterized by a reduction of knee movement (Burgess-Limerick et al., 2001). In addition to mobility requirements, SCK movements can constitute a significant challenge to the balance and strength (force- generating capacity) capacities of older adults, which may explain why “bending down” tasks are included in clinical assessments such as the Physical Performance Test (Reuben and Siu, 1990) and Berg Balance Scale (Berg et al., 1992) that are thought to predict fall risk. However, even though SCK difficulty may be a significant indicator of mobility and independence in older adults, few data exist regarding the determinants of SCK difficulty and its association with fall risk.

The prevention of falls in older adults is the focus of much research effort. A key component of avoiding falls is the motor system’s ability to produce joint torques to counteract perturbations that would lead to losses of balance (Hall et al., 1999). The neuromuscular system invokes a series of response strategies involving extremity and trunk musculature (Horak and Nashner, 1986; Alexander et al., 1992; Grabiner et al., 1993; Marigold et al., 2003) to produce joint torques at various body segments in order to maintain the center of mass over the base of support or return the center of mass rapidly to within the base of support when external perturbations have altered its position. Muscle weakness contributes to both falls and self- reported SCK difficulty in older adults, but the relative contribution of different muscle groups, such as proximal (trunk and hip) versus distal (ankle) muscle strength, is not entirely clear (Moreland et al., 2004; Hernandez et al., 2008).

The association of reduced lower extremity muscle forces, particularly ankle strength, with falls is well established (Whipple et al., 1987; Studenski et al., 1991; Daubney and Culham, 1999). Leg strength also has been associated with functional performance such as gait speed and chair rises (Brown et al., 1995; Bean et al., 2002). Ankle dorsiflexor and ankle plantar-flexor strength measures predict performance on some (Timed “Up & Go” Test [TUG] and Berg Balance Scale) but not all (unipedal stance time [UST]) common clinical balance tests that are thought to predict falls in older adults (Daubney and Culham, 1999; Ringsberg et al., 1999). Although knee extensor strength is associated with static and dynamic capabilities as well as functional ability in older adults (Carter et al., 2002; Avlund et al., 1994), the role of more proximal trunk and hip muscles in determining functional performance has not been well characterized. Given that control of the flexing trunk is critical to avoiding falls when losses of balance occur (Grabiner et al., 1993), it is important to determine the relative contributions of both trunk and lower-extremity strength to functional tasks such as SCK.

The purposes of this study were: (1) to examine trunk and lower-extremity muscle strength differences in older adults with and without SCK difficulty and (2) to examine the relative contributions of trunk and leg muscle strength to SCK difficulty that may predict falls in older adults with a range of balance impairments. Of particular interest are the relative contributions of muscle strengths that have not been well studied in older adults, including strength at the trunk and hip. These data will advance the understanding of the contribution of strength measures, particularly of the proximal lower extremity and trunk, to self- reported functional performance and may allow a more thoughtful approach to the use of strength training in improving balance and thereby reducing falls. Given the previous data on the importance of ankle function in balance performance and fall risk, we hypothesized that lower- extremity strength would be significantly decreased in older adults with SCK difficulty and that distal strength measures would be the main predictors of SCK difficulty.

A.3. Method

A.3.1. Participants

Functionally independent, community-dwelling older adults were recruited largely from a database maintained by the University of Michigan Older Americans Independence Center Human Subjects and Assessment Core. Inclusion criteria for the study were age over 65 years, ability to speak and understand English, and ability to stand for 5 minutes without an assistive device. A nurse practitioner performed a screening medical history and physical examination and excluded those whose medical conditions precluded the ability to complete the test battery, such as those individuals who: (1) were medically unstable (e.g., chest pain upon exertion, dyspnea, acute infection), (2) reported severe and frequent back or lower-extremity pain, or (3) reported severe neurologically induced impairments that might affect balance (e.g., history of cerebrovascular accident, Parkinson disease). Out of 50 recruited individuals, 2 were excluded due to severe osteoporosis and severe and frequent back pain. During screening, participants rated their ability to stoop, crouch, or kneel, according to a 5-point difficulty scale, based on a single question on the Established Populations for Epidemiologic Studies of the Elderly (EPESE) questionnaire (Smith et al., 1990): no difficulty (n=21), a little (n=13), some (n=9), a lot (n=2), or unable to do (n=3). Participants were categorized into 1 of 2 groups: a no SCK difficulty group (n=21), if they reported no difficulty, or an SCK difficulty group (n=27) if they reported any SCK difficulty. Participants signed a written informed consent form approved by the University of Michigan Medical School Institutional Review Board. Based on previous work (Hernandez et al., 2008), power analysis (Faul et al., 2007) revealed that a sample size of 21 for each group was required to achieve a power of 0.80 with an alpha level of .05 in detecting strength differences between older adults with and without SCK difficulty.

A.3.2. Instrumentation and Measures

Self-report health measures. The nurse practitioner used self-report (interview-based) health measures to obtain patient data at the medical screening. The total number of chronic medical conditions was ascertained by asking participants if they had a previous history of osteoarthritis, rheumatoid arthritis, osteoporosis, myocardial infarction, stroke, joint re- placement, Parkinson disease, or peripheral neuropathy. Dizziness was determined by asking participants if they had a current episode of light-headedness or vertigo. Self-reported leg joint limitations were determined by the report of joint range-of- motion limitations due to pain or stiffness in the hip, knee, or ankle.

Clinical balance measures. The 3 clinical balance measures that were examined were the UST, the TUG, and the maximum step length (MSL). The UST is a commonly used clinical balance measure. Deficits in UST, defined as the inability to stand unsupported on one leg for more than 5 seconds, is a strong predictor of injurious falls (Vellas et al., 1997). Participants stood on their preferred leg while their arms were folded. The foot of the remaining leg was lifted and held approximately 5.08 cm (2 in) from the medial malleolus of the stance leg. Participants performed a practice trial followed by 2 experimental trials. The best time (maximum 30 seconds) was recorded as the UST. The UST has excellent interrater reliability (intraclass correlation coefficient [ICC]=.99) (Franchignoni et al., 1998).

The TUG is a reliable measure of functional mobility and dynamic balance in older adults (intrarater ICC=.99, interrater ICC=.99) (Podsiadlo and Richardson, 1991). Participants sat in a chair and, on command, stood up and walked 9.84 ft (3 m) at their "comfortable and safe pace" before turning around and returning to the seated position. The time to complete this task is the person's TUG score. Scores exceeding 14 seconds have been associated with increased fall risk in older adults (Shumway-Cook et al., 2000). Participants performed a practice trial, followed by 3 experimental trials. The TUG was recorded as the mean time (in seconds) of the 3 trials.

The MSL test is a reliable test of stepping ability, which correlates with standard balance measures (e.g., unipedal stance, tandem stance, tandem walk), functional mobility measures, (e.g., TUG, Performance-Oriented Mobility Assessment, Six-Minute Walk Test) and fall history (intrarater ICC=.91) (Medell and Alexander, 2000; Cho et al., 2004; Schulz et al., 2007). Standing with arms folded and feet together, participants stepped forward maximally with their dominant leg and returned to the original starting position. Leg dominance was ascertained using a simple screening question, namely the preferred foot used to kick a soccer ball. Participants performed a practice trial followed by 5 experimental trials, and MSL was recorded as the mean distance stepped over the 5 trials. To account for individual anthropometric differences, the MSL was normalized as a percentage of body height (% height).

Strength measures. Isometric peak torque of the trunk, hip, and knee extensors, as well as of the ankle plantar-flexors and dorsiflexors of the dominant lower extremity, was evaluated using the Biodex multijoint isokinetic dynamometer (Biodex Medical Systems Inc., Shirley, NY). After a practice trial, participants exerted a maximal contraction for 2 trials of 3 seconds each. Participants received approximately 45 seconds of rest between trials, the best of which was recorded as the peak torque. If the 2 torque measurements differed by more than 15%, an additional trial was performed to achieve consistency.

Trunk extensor isometric strength was evaluated using a back attachment (Biodex Medical Systems Inc., Shirley, NY) affixed to the multijoint isokinetic dynamometer. Participants were supported and stabilized in the seated position via chest, pelvic, and thigh straps, with the arms folded and the trunk and lower extremities configured at approximately 90 degrees to each other. Participants exerted a maximal trunk extensor contraction against a pad located at the interscapular region.

Hip extensor isometric strength was evaluated in the functional standing position as previously described (Dean et al., 2004). Briefly, this measure required the participants to stand supported and stabilized upright in a standing frame with the lower extremity straight and the hip flexed 15 degrees in the sagittal plane. Participants exerted a maximal hip extensor contraction by pushing against a pad located immediately distal to the popliteal fossa.

Knee extensor isometric strength was evaluated in the seated position with the knee flexed to 90 degrees and chest and thigh straps providing support and stabilization. With the arms folded, participants exerted a maximal knee extension contraction against a pad located at the anterior distal tibia.

Isometric strength of the ankle plantar-flexors and dorsiflexors was evaluated with the participants in a semi- reclined position with the tibia parallel to the floor, the foot and tibia at 90 degrees to each other, and the knee and hip at approximately 30 and 60 degrees of flexion, respectively. The lower leg was supported with a pad and straps, and with arms folded, participants either pushed maximally into (plantar flexors) or pulled away from (dorsiflexors) a footplate in which the foot was tightly secured with straps. Peak torque values are expressed as raw strength (in Nm) or as normalized strength, as a percentage of the product of body weight (BW) (newtons) and body height (BH) (meters). as done in previous studies (% BW x BH) (Hernandez et al., 2008; Schulz et al., 2007).

Reliability of isometric strength testing using the Biodex multijoint dynamometer system has been demonstrated previously in measurements of knee strength for older men (intrarater ICC=.90) (Symmons et al., 2005).

Fall-related measures. Data for fall-related measures were obtained during the medical screening by asking participants whether they had fallen within the past year. In the event that a fall was reported, the number of falls incurred and whether medical treatment was sought for their injuries (i.e., injurious falls) were ascertained. For analysis, the number of falls and number of injurious falls within the past year were dichotomized (i.e., falls ≥ 2 and injurious falls ≥ 1).

A.3.3. Data Analysis

All statistical analyses were carried out in SPSS 16.0 for Windows (SPSS Inc., Chicago, IL). To test for normality in continuous variables, the Kolmogorov-Smirnov test was used. Group comparisons for continuous participant description, self-reported health, clinical balance, and strength data in older adults with and without SCK difficulty were evaluated using an independent-samples t test when data were normally distributed and a Mann-Whitney U test when data were not normally distributed (e.g., number of chronic medical conditions, UST, trunk extensor strength). To determine group differences in dichotomous variables, either the Pearson chi-square test (e.g., sex, dizziness, self-reported leg joint limitations) or the Fisher exact test (e.g., fall-related measures) was performed. Relationships among strength, clinical balance measures, and SCK difficulty were evaluated using Spearman correlation coefficients, using the full 5-point scale of SCK difficulty. A forward stepwise binary logistic regression analysis included all the strength variables with a significant correlation ($P < .05$) to determine which measurements were important predictors of self- reported SCK difficulty. A log transformation resulted in a normal distribution for UST and trunk strength measures. Statistical analyses were carried out with and without the log transformation to the UST and trunk strength measurements, but yielded similar results. Thus, for ease of interpretation, the untransformed variables are presented in this article. Two separate logistic regression models, one for normalized strength and one for raw strength, are presented. We performed an analysis of multicollinearity among the variables. Even with all strength measures entered into a linear regression model, we found no evidence of collinearity, as the variance inflation factor was not greater than 10 for any raw or normalized strength measure (Myers, 1990). A significance level of .05 was used for all correlational and regression analyses, and a Bonferroni correction was used for multiple SCK group comparisons of strength (.05/5).

A.4. Results

A.4.1. Participant Characteristics and Strength Measures

Forty-eight older adults (62.5% women, 37.5% men) participated in the study. Participant characteristics (demographic, self-reported health, clinical balance, strength, and fall-related data) are presented for the entire sample in **Table A.1**. Compared to older adults without SCK difficulty, older adults with SCK difficulty were older and had increased self-reported leg joint limitations, decreased UST, increased TUG scores, and decreased MSL. The incidence of two or more falls within the previous year (P=.096) tended to be higher in older adults with SCK difficulty.

Compared to older adults without SCK difficulty, mean (SD) raw ankle plantar-flexor strength was significantly lower in those with SCK difficulty (76.4 ± 25.6 Nm vs. 53.9 ± 29.8 Nm, respectively) (**Table A.1**). Raw ankle dorsiflexor and knee extensor strength were decreased in older adults with SCK difficulty versus those without SCK difficulty, but when the Bonferroni correction was applied, the difference was not statistically significant.

Table A.1: Characteristics of older adults with and without stooping, crouching, or kneeling (SCK) difficulty.

Characteristic[a]	No SCK Difficulty (n=21) X (SD) or %	SCK Difficulty (n=27) X (SD) or %	P-value
Participant description			
Age (years)	73.5 (5.4)	77.1 (6.1)	0.037
Height (cm)	165.3 (9.6)	166.5 (10.7)	0.675
Weight (kg)	70.6 (11.2)	77.3 (18.0)	0.121
Body mass index (kilograms/meters^2)	25.8 (3.7)	27.6 (4.4)	0.145
Gender, female (%)	62	63	0.94
Self-reported health			
Chronic medical conditions (n)	0.6 (0.6)	0.9 (0.7)	0.141
Dizziness (%)	19	26	0.733
Self-reported leg joint limitation (%)	5	41	0.004
Clinical balance measures			
UST (seconds)	22.0 (8.8)	9.5 (9.9)	<.001
TUG (seconds)	9.6 (1.5)	11.5 (2.5)	0.004
MSL (% height)	48.0 (8.6)	40.1 (8.6)	0.003
Raw strength measures			
Ankle dorsiflexor strength (Nm)	27.7 (11.9)	20.1 (8.4)	0.013
Ankle plantarflexor strength (Nm)	76.4 (25.6)	53.9 (29.8)	0.008
Knee extensor strength (Nm)	128.0 (37.9)	95.9 (46.3)	0.013
Hip extensor strength (Nm)	81.6 (26.1)	72.8 (29.4)	0.284
Trunk extensor strength (Nm)	72.4 (45.2)	53.1 (33.8)	0.053
Fall-related			
≥2 falls (%)	5	22	0.096
≥1 injurious fall (%)	10	22	0.22

Notes: [a] Means and standard deviations (SD) are presented for continuous variables, and proportions are presented for categorical variables. For determining group differences, continuous measures were evaluated using either an independent samples t-test or a Mann-Whitney U test, and dichotomous measures were evaluated using a Pearson chi-square test or Fisher's Exact Test. SCK = stooping, crouching, or kneeling; UST = unipedal stance time; TUG = Timed "Up and Go" Test; MSL = maximum step length.

After normalization of strength data to account for body size (e.g., body height and weight), mean normalized strength was higher in older adults without SCK difficulty compared to those with SCK difficulty in ankle dorsiflexors (2.4 ± 0.9 % BW x BH vs. 1.6 ± 0.6 % BW x BH, respectively), ankle plantar-flexors (6.8 ± 2.4 % BW x BH vs. 4.3 ± 2.3 % BW x BH, respectively), knee extensors (11.1 ± 2.2 % BW x BH vs. 7.6 ± 3.0 % BW x BH, respectively), and trunk extensors (6.3 ± 3.7 % BW x BH vs. 4.3 ± 2.3 % BW x BH, respectively) (P<.01, **Figure A.1**). Mean normalized isometric peak torque (% Body Weight x Body Height) for all muscle groups tested in older adults with and without SCK difficulty. Error bars show one standard deviation (* indicates p < .01). ADF = Ankle Dorsiflexors; APF = Ankle Plantarflexors; KE = Knee Extensors; HE = Hip Extensors; TE = Trunk Extensors.

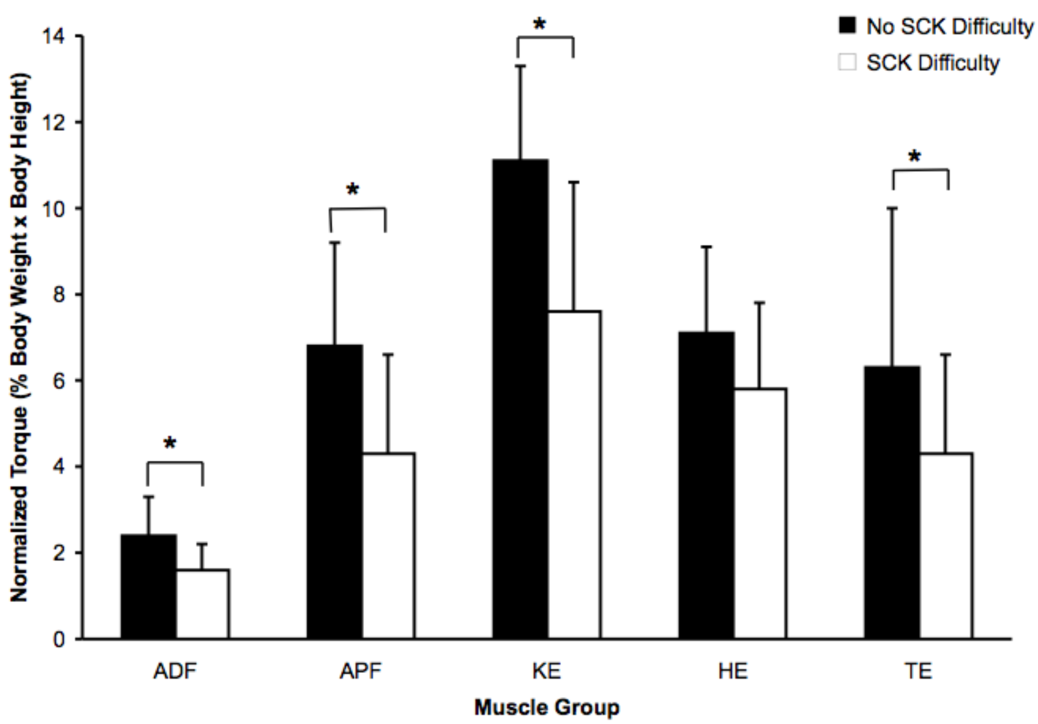

Figure A.1: Mean normalized isometric peak torque (% Body Weight x Body Height) for all muscle groups tested in older adults with and without SCK difficulty. Error bars show one standard deviation (* indicates P < .01). ADF = Ankle Dorsiflexors; APF = Ankle Plantarflexors; KE = Knee Extensors; HE = Hip Extensors; TE = Trunk Extensors.

A.4.2. Correlations with SCK Difficulty

Considering the full 5-point ordinal scale of SCK difficulty, SCK difficulty was associated with raw knee extensor (r=-.43) and ankle plantar-flexor (r=-.39) strength (P<.01), and to a lesser degree with ankle dorsiflexor strength (r=-.31, P<.05). However, no significant correlations between SCK difficulty and raw hip or trunk extensors were observed. Normalization of strength variables led to strong associations between all strength measures and SCK difficulty, such that decreased strength was associated with SCK difficulty (r=-.37 to -.58, P<.01, **Table A.2**). Stooping, crouching, or kneeling difficulty was also strongly correlated with functional mobility in all 3 clinical balance tests (P<.005), particularly with UST (r=-.62).

Table A.2: Correlations between SCK Difficulty and both Normalized Strength and Clinical Balance Measures[a].

	SCK Difficulty
Ankle dorsiflexor strength	-0.44[c]
Ankle plantarflexor strength	-0.48[c]
Knee extensor strength	-0.58[c]
Hip extensor strength	-0.38[b]
Trunk extensor strength	-0.37[b]
UST	-0.62[c]
TUG	0.47[c]
MSL	-0.46[c]

Notes: [a] Values are Spearman's Correlation Coefficient (r). SCK difficulty was scored on a 5-point ordinal scale. Strength measures were normalized for body size (height x weight) and MSL was normalized for body height prior to computation of Spearman's r. UST = Unipedal Stance Time; TUG = Timed Up and Go; MSL = Maximum Step Length.
[b] Significant at P < .01.
[c] Significant at P < .005.

A.4.3. Strength as a Predictor of SCK Difficulty

Based on findings from the primary analyses, raw ankle and knee strength were entered in a forward stepwise binary logistic regression. The model showed that ankle plantar-flexor strength (odds ratio [OR] = 0.97; 95% confidence interval [CI], 0.95-0.99) was significantly associated with SCK difficulty (P=.014), but explained only 11% of the variance. Considering the primary analyses of normalized strength, all five strength measures were entered into a forward stepwise binary logistic regression model. The model showed that the most significant predictor for SCK difficulty was normalized knee extensor strength (OR = 0.61; 95% CI, 0.44-0.82 P=.001), which explained 26% of the variance in SCK difficulty (**Table A.3**).

Table A.3: Results of Logistic Regression Analyses of Muscle Strength Measurements as Predictors of SCK Difficulty[a].

	Variable	P	Odds Ratio	95% Confidence Interval
Model 1	Ankle Plantarflexor[b]	0.014	0.97	0.95-0.99
Model 2	Normalized Knee Extensor[c]	0.001	0.61	0.44-0.82

Notes: [a] SCK Difficulty was scored as a dichotomous measure.
[b] R2 = .11 (Hosmer & Lemeshow). Raw Muscle Strength Model $\chi2(1) = 7.23$.
[c] R2 = .26 (Hosmer & Lemeshow). Normalized Muscle Strength Model $\chi2(1) = 17.22$.

A.5. Discussion

The results of this study highlight the significant association between strength and self-reported SCK difficulty. We hypothesized that lower- extremity strength would be significantly decreased in older adults with SCK difficulty and that distal strength measures would be the main predictors in SCK difficulty. Results from the primary analyses of raw and normalized strength are mostly in agreement with our hypothesis, as leg strength, particularly after normalization, was significantly decreased in older adults with SCK difficulty. Two novel findings of this study were: (1) the significant correlation between SCK difficulty and both normalized strength and functional balance measures, and (2) that among a broad sample of leg and trunk muscle strengths, those having the greatest association with SCK difficulty were raw ankle plantar-flexor and normalized knee extensor strengths. Overall, these findings suggest that in older adults, the major strength determinant of SCK difficulty is strength of the distal leg musculature, thereby providing a common link with functional tests of balance.

This is the first study to examine the differences in trunk and lower extremity muscle strength in older adults with SCK difficulty. Although ankle plantar-flexor strength was found to be a significant predictor of SCK difficulty, normalization of strength measures to account for differences in body size demonstrated that normalized knee extensor strength is a more significant predictor of SCK difficulty as evaluated by the percentage of variance explained (26% for the knee versus 11% for the ankle). As in the present study, decreased lower- extremity strength has been found to be associated with SCK limitations (Hernandez et al., 2008; Janssen et al., 2002) and limited functional performance in daily activities such as rising from a chair, walking, or stair ascent or descent (Moxley Scarborough et al., 1999; Buchner et al., 1996; Hurley et al., 1998). Similarly, frequent fallers have reported decreased lower extremity strength (Tinetti et al., 1986), and older adults with a high fall risk have been best identified by maximum isometric push-off force in a leg press apparatus (Pijnappels et al., 2008).

The significant correlations found between SCK difficulty and functional balance tests such as UST, TUG, and MSL extend the results of other investigators who found an increased risk of falls in older adults who reported trouble bending down to the floor (O'Loughlin et al., 1993). Distal strength consistently had stronger correlations to SCK difficulty than proximal strength measures and may provide a common link between SCK difficulty and functional balance tests. Adequate ankle dorsiflexor and plantar-flexor strength may be required to generate corrective torques about the ankle to maintain equilibrium by moving the center of mass forward or backward during stooping movements, due to the limited range of motion at the knee and hip. Knee extensor strength would be expected to play a significant role in the recovery to an upright stance after crouching or kneeling. Similarly to SCK movements, lower extremity strength is of primary importance to the performance of functional tests such as the UST, TUG, and MSL (Brown et al., 1995; Bean et l., 2002; Ringsberg et al., 1999; Riemann et al., 2003). Even though fall-related measures such as the incidence of more than 2 falls or an injurious fall were not found to be significantly different between older adults with and without SCK difficulty, this might have been due to the overall health of this study cohort or limitations in sample size.

After normalization of strength measures, hip extensor and trunk extensor isometric strength were found to correlate with SCK difficulty. These muscle groups would be expected to play a role in performing stooping tasks and recovery of the trunk after bending down to the ground, but appear to be of less significance to distal musculature as seen in the multivariate analysis. A possible explanation for the lack of significance of hip and trunk extensors is that additional muscle groups, such as lateral hip muscles (e.g., hip abductors), may be playing a role in the stabilization of the torso during the large motions undertaken while stooping, crouching, or kneeling.

The results of this study have important implications for clinicians working to reduce fall risk in older adults. Rehabilitation or intervention programs aimed at addressing deficits in self-reported SCK performance should focus on improving distal strength. Although reduced strength is a significant contributor to SCK difficulty, older adults with SCK difficulty also may benefit from more comprehensive programs that address balance confidence, coordination, leg joint limitations such as stiffness and pain, and sensory capacities.

A limitation of this study is that although we included participants with a wide range of SCK difficulty and balance capabilities, some of whom may be at risk for a fall, the participants were all community-dwelling volunteers and generally active. Therefore, it is undetermined whether similar results would be seen in a frailer cohort of older adults. Future studies should include older adults who are regular fallers to understand the strength determinants of SCK difficulty and clinical measures of balance and trunk control in those having the highest risk for falls. Finally, a further limitation of this study is that although it focused on muscle groups that were key to rising from SCK, there are other muscle groups, such as hip abductors, that may contribute to safe SCK performance, particularly as the older adult lowers himself or herself.

A.6. Conclusions

Decreased muscle strength, particularly when normalized for body size, predicts SCK difficulty. These data emphasize the importance of strength measurement at multiple levels in predicting self-reported functional impairment. Further studies are needed to determine whether rehabilitation programs with a focus on training specific muscle groups are effective in improving self-reported functional performance and whether improvements in self-reported functional performance are associated with fewer falls in older adults.

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APPENDIX B

CHANGES IN DISTAL POSTURAL CONTROL ACCURACY NEAR THE LIMITS OF THE BASE OF SUPPORT

B.1. Abstract

Balance is an integral component of daily living; requiring rapid control of the center of pressure (COP) to maintain the body's center of mass within a finite base of support while standing. However, little is known about changes in volitional COP movement accuracy with changes in the location of foot support. This study examined the effect of foot support and target position on volitional COP movements. Eleven healthy young adults performed accuracy-constrained COP movements of the trunk, using visual-feedback, to targets placed at either the limits or towards the center of the functional base of support (FBOS), while standing on either their forefoot or rearfoot. COP movements when standing on the rearfoot were 17% slower and had a 23% higher ratio of peak over average COP velocity ($P<.005$) than those when standing on the forefoot. In comparison to targets placed towards the center of the FBOS, targets placed near the limits of the FBOS led to 16% faster and 67% more accurate COP movements ($P<.005$). Intraclass correlation coefficients demonstrated good to excellent reliability

across all dependent measures, varying from 0.72 to 0.98. These findings demonstrate that changes in the desired volitional COP movement endpoint can lead to significant changes in speed and accuracy. We conclude that in healthy young adults, slower speeds are needed to maintain accurate COP movements when standing on the rearfoot, rather than when standing on the forefoot, thereby suggesting a possible mechanism for balance deficits during stair descents.

B.2. Introduction

Balance is an integral part of daily living; requiring rapid control of the center of pressure (COP) to maintain the body's center of mass (COM) within a finite base of support during bipedal stance. As the COP represents the cumulative input of both the postural control system and the force of gravity, movements of the COP while standing are subject to increased variability due to postural sway, particularly under leaning conditions (Duarte & Zatsiorsky, 2002, Latash et al., 2003). Furthermore, balance can be affected by the position and orientation of the foot on the floor (Kirby et al., 1987, Mezzarane & Kohn, 2007). The inherent variability of the postural control system may be one of the limiting factors of fine and accurate volitional movements of the COP (Duarte & Freitas, 2005, Danion et al., 2006). However, even though foot position can significantly affect balance, the relationship between volitional COP movements and location of foot support remains largely unexplored.

Accuracy-constrained movements have demonstrated consistent tradeoffs in their velocities across a wide number of tasks (Duarte & Freitas, 2005, Danion et al., 1999, Fitts, 1954, Meyer et al., 1982, Duarte & Latash, 2007). Studies of volitional COP movements have evaluated the effect of accuracy constraints on movement time and speed (Duarte and Freitas, 2005, Danion et al., 1999), but have not yet reported the reliability of these measures. Accuracy-constrained movements using visual feedback from a computer display are not the typical task that the postural control system performs. Thus, it is our impetus to not only evaluate changes in distal postural control near the limits of the base of support, but to also verify the reliability of fundamental measures of volitional COP movements.

B.3. Methods

B.3.1. Subjects

Eleven healthy young adults, 6 male and 5 female, participated in this study. Their age ranged from 18 to 28. The mean (±SD) subject height, mass, and foot length were 1.68 ± 0.26 m, 68.4 ± 15.6 kg, and 0.30 ± 0.01 m, respectively. All subjects were healthy with no prior physical or mental disorders. All participants gave informed consent according to the procedures approved by the Institutional Review Board of the University of Michigan.

B.3.2. Protocol

The functional base of support (FBOS) is a quasi-static measure of functional stability (King et al., 1994). The FBOS was calculated by instructing participants to fold their arms across their chest and to lean as far forward and backward as possible without losing balance, while strapped to a ceiling harness. While standing in a parallel stance, subjects viewed a computer monitor with visual feedback of both their COP and anteroposterior targets (**Figure B.1**). During experimental trials, subjects were asked to move their COP ‘as fast and as accurate as possible’ between the targets for a duration of

15 seconds, while standing on either their forefoot or rearfoot (foot position effect). Movement amplitude and target size were fixed at 25% and 12.5% of the FBOS, respectively. Targets were positioned towards the limit or center of the FBOS (target placement effect). Subjects performed four repeated measures (blocks) of a set of randomized trials, consisting of a 60-second training period that preceded each experimental trial.

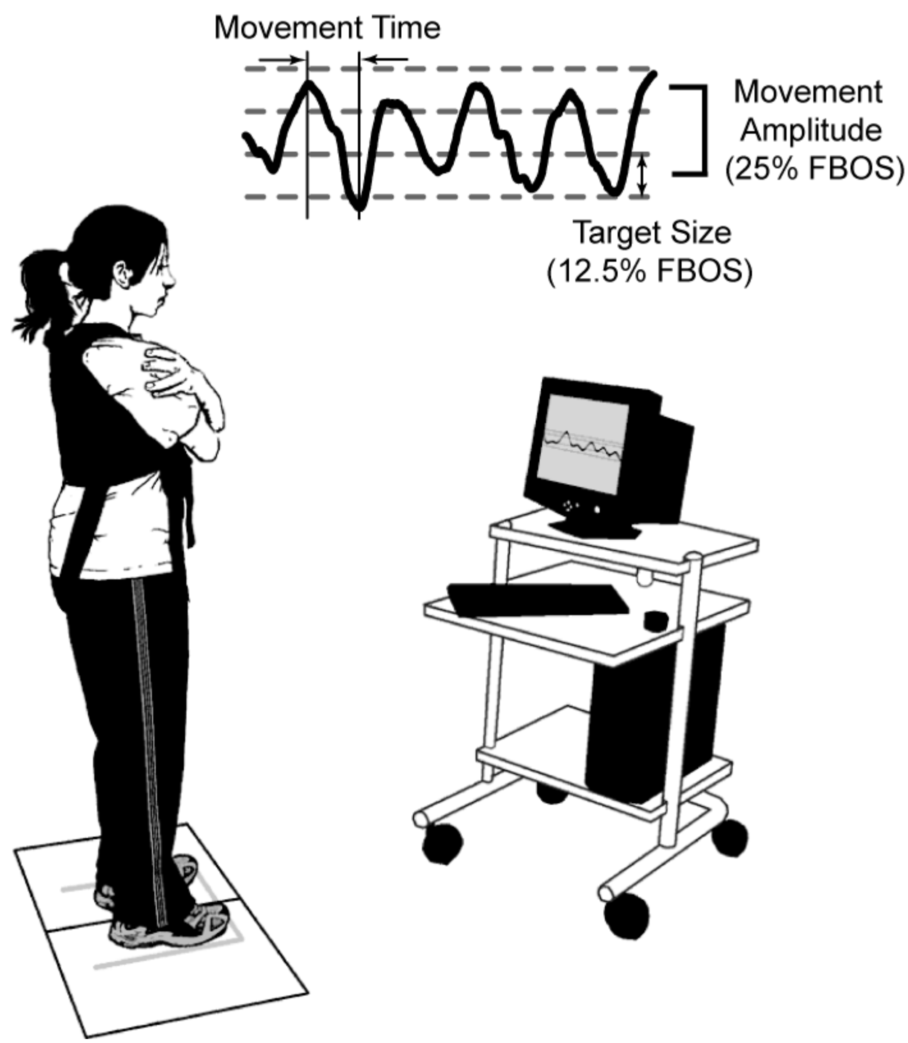

Figure B.1: Top) Exemplary COP movement illustrating the definition of movement time, movement amplitude, and target size. Bottom) Illustration of experimental setup, including force plates and real-time biofeedback.

B.3.3. Data Collection and Processing

Real-time COP calculations were performed using NDI Toolbench software (Northern Digital, Inc., Waterloo, Canada) from force plate data sampled at 100 Hz (OR6-7-1000, AMTI, Watertown, MA). Custom Matlab (v7.4, Natick, MA) software was written for data analysis. Raw force plate data was processed using a 4^{th} order, zero-lag, low-pass Butterworth filter with a 30 Hz cutoff frequency. COP velocity was calculated using a five-point finite difference derivative algorithm. Using an automated procedure, individual COP movements were extracted from each trial and characterized by the amplitude and elapsed time from anterior to posterior maxima of the COP trajectory. Movement speed was calculated by averaging movement amplitude over movement time. The ratio of peak over mean COP velocity (RCOPV) was calculated dividing each movement's mean by their peak COP velocity. The accuracy rate was determined by calculating the percentage of movements in a trial whose endpoints were within target boundaries. To test for the effect of foot position and target placement, the mean outcome measures were averaged over the four experimental trials, whereas all experimental conditions were aggregated for each individual block to assess test-retest reliability.

B.3.4. Statistical Analysis

To test the effect of foot position and target placement on mean COP speed, RCOPV and accuracy rate, a two-way repeated-measures ANOVA was performed using SPSS 16.0 for Windows (SPSS Inc., Chicago, IL). A full factorial analysis was utilized for each dependent measure. Hochberg's step-up method was used to control for multiple comparisons, with statistical significance set at $P < .05$. Post hoc, paired sample t-tests were performed to identify differences between individual foot position and target placement conditions. Test-retest reliability was assessed using both a repeated-measures ANOVA and intraclass correlation coefficients (ICC).

B.4. Results

COP movements while standing at the rearfoot were 17% slower and had a 23% higher ratio of peak over average COP velocity (RCOPV) than those while standing at the forefoot. Targets placed near the limits of the FBOS led to 16% faster and 67% more accurate COP movements than targets placed towards the center of the FBOS (**Table B.1**). After controlling for multiple comparisons, no significant differences were seen in the accuracy rate between foot positions. Trends towards an increased RCOPV in movements towards the center of the FBOS were not significant. No significant interactions were found between target placement and foot position for all dependent measurements. Post-hoc paired sample t-tests were used to illustrate significant differences between target placement toward the anterior and posterior limits of the FBOS (LIMIT) or towards the center of the FBOS (CENTER), and between the forefoot and rearfoot foot positions in **Figure B.2**.

Significant increases in the mean COP speed and decreases of the RCOPV occurred over the four blocks of repeated measures (**Figure B.2, Table B.2**). Trends for increased accuracy over time were found to be insignificant after correcting for multiple comparisons. Across all four blocks, the intraclass correlation coefficients (ICCs) demonstrated adequate to excellent reliability. The ICCs for accuracy rate, RCOPV and COP speed were 0.72, 0.79 and 0.98, respectively. Post-hoc paired sample t-tests were used to illustrate significant differences between consecutive blocks in **Figure B.2**.

Results demonstrated slow learning effects in the overall COP movement speed, and rapid learning in the RCOPV.

Table B.1: Effects of foot position and target placement on outcome measures

Effect	df	F	p-value
COP speed			
Foot Position	1,10	29.53	<0.001*
Target Placement	1,10	20.91	<0.001*
Foot Position x Target Placement	1,10	0.1	0.761
Ratio of Peak over Average COP velocity (RCOPV)			
Foot Position	1,10	21.38	0.001*
Target Placement	1,10	8.47	0.016
Foot Position x Target Placement	1,10	0.32	0.587
Accuracy Rate			
Foot Position	1,10	6.41	0.03
Target Placement	1,10	229.27	<0.001*
Foot Position x Target Placement	1,10	0	0.955

*Denotes a significant difference, as determined by Hochman's step-up test.

Table B.2: Effect of repeated measures on dependent variables

Measure	df	F	p-value
COP speed	3,30	14.24	<0.001*
Ratio of Peak over Average COP velocity (RCOPV)	1.6,16.0	15.87	<0.001*
Accuracy Rate	3,30	3.29	0.034

*Denotes a significant difference, as determined by Hochman's step-up test.

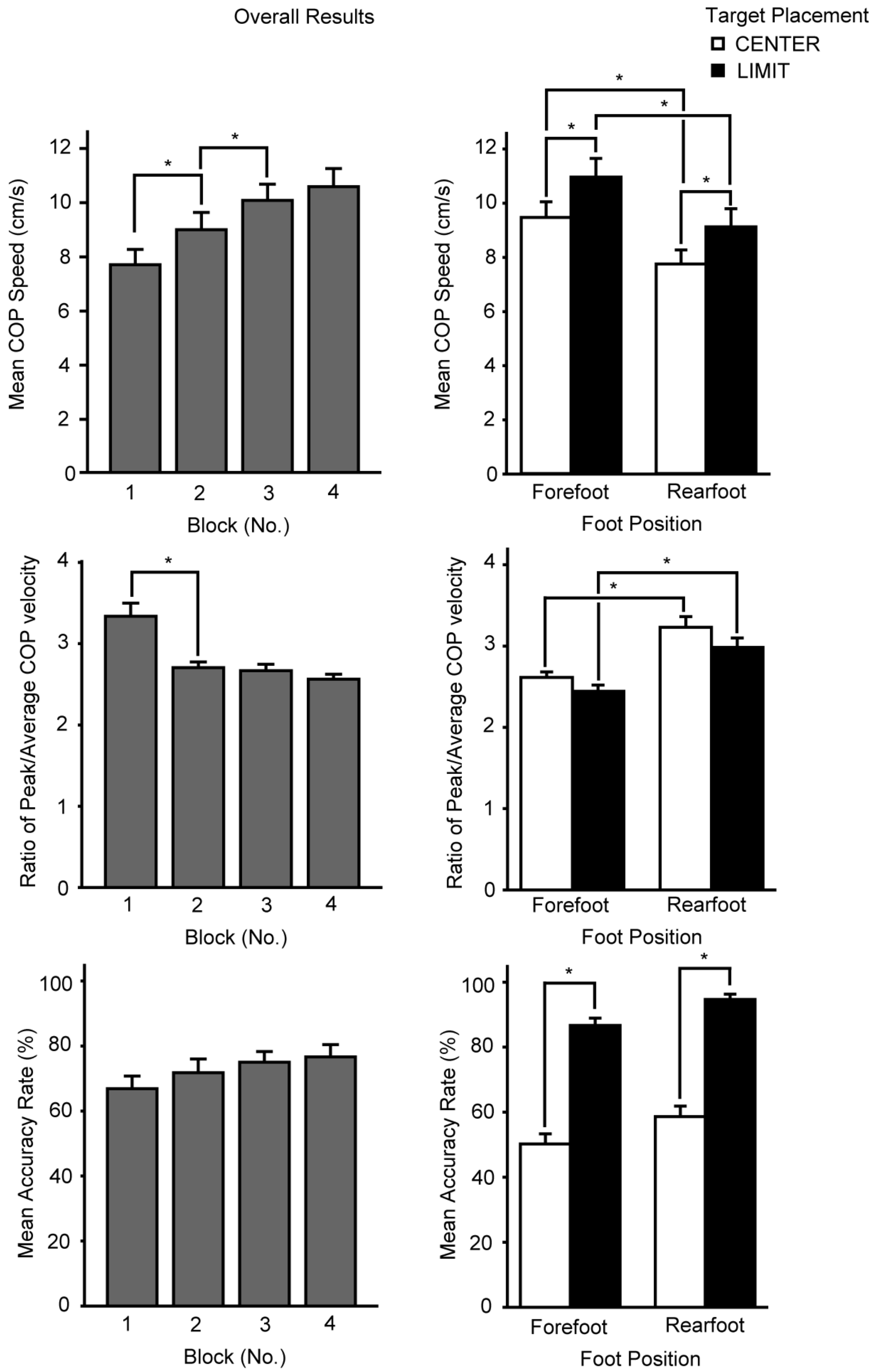

Figure B.2: Left) Reliability of dependent variables over four blocks of repeated measures. Right) Effect of target placement and foot position on the mean (SE) COP speed, the ratio of peak over average COP velocity (RCOPV), and accuracy rate. * Denotes a statistical significance of P < .05.

B.5. Discussion

To our knowledge, this is the first study to examine the effects of foot position and target placement on accuracy-constrained volitional COP movements. Consistent with previous studies of balance under toes-up and toes-down foot positions, distal postural control was affected (Mezzarane and Kohn, 2007). All dependent measures were found to be reliable, particularly speed, which presents another novel aspect of this study.

Changes in foot position in the sagittal plane can lead to the use of different cutaneous receptors and changes in muscle spindles, which may affect COP control (Mezzarane and Kohn, 2007; Inglis et al., 2002). In particular, anteroposterior postural control of upright stance is significantly affected by diminished plantar cutaneous sensation (McKeon and Hertel, 2007). Consequently, the changes seen in COP speed and RCOPV in this study may be associated with changes in distal postural control capacity between the toe and heel.

Positioning targets near the limits of the FBOS may pose a significant risk of loss of balance to subjects, which explains the increase in movement accuracy when compared to targets towards the center of the FBOS. Trends of a decreased RCOPV in movements towards the limits, rather than the center of the FBOS, suggest an increased reliance on feedforward control (Smits-Engelsman et al., 2002). Thus, providing an explanation for the increased speed of COP movements seen at the anterior and posterior limits.

Accuracy-constrained movements in this study were undertaken near the limits of each subject's FBOS, which provided a challenging postural configuration. Frequent practice was used to promote mastery, or at the very least, familiarity, in experimental trials. A limitation of this study was its one-day duration, as test-retest reliability would be susceptible to short-term learning and fatigue effects. Accuracy was not strictly regulated in this study, which might have led to an underestimation of target position effects on speed. Only healthy young adults were examined in this study. Thus, it would be of interest to extend the study of accuracy constrained movements to other populations of interest, such as older adults, which may be at risk of losses of balance, and their subsequent falls, due to age-related changes in COP control.

In the accuracy-constrained task introduced in this study, slow or negligible learning effects were observed in the accuracy and speed of volitional COP movements in healthy young adults. The rapid learning that occurred in RCOPV is indicative of a rapid change in efficiency at controlling volitional COP movements in this novel task.

The significant effect of foot position and target placement on the characteristics of accuracy-constrained COP movements is clinically relevant. As slower speeds are required to preserve accuracy when standing on the rearfoot, and would require the appropriate changes in strategies to compensate for the constraints in COP speed. We conclude that in healthy young adults, slower speeds are needed to maintain accurate COP movements when standing on the rearfoot, rather than when standing on the forefoot, thereby suggesting a possible mechanism for balance deficits during stair descents.

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APPENDIX C

DOWNWARD REACH AND PICK-UP STRATEGIES IN OLDER FEMALES WITH SELF-REPORTED DIFFICULTY

C.1. Abstract

The purpose of this study was to examine the differences between older women with and without self-reported stooping, crouching, or kneeling (SCK) difficulty in self-selected base of support configuration and lifting strategy. Sixteen older women were recruited in southeastern Michigan to undergo a cross-sectional study, consisting of one-handed and two-handed downward reach pick-up tasks. Video analysis was used to measure the incidence of stance asymmetry, width of the base of support, distance from the base of support to the object, and postural index at the time of pick-up. Older women with SCK difficulty stooped more in one-handed tasks ($P<.05$) and used a wider base of support in two-handed tasks ($P=.10$) than those without SCK difficulty. Single-handed tasks discriminated better between the groups than two-handed tasks, further supporting the use of single-handed tasks in clinical fall risk batteries, and those with self-reported SCK difficulty stooped more and user a wider stance to pick up an object from the floor.

C.2. Introduction

Stooping, crouching, or kneeling (SCK) are common activities of daily life, being necessitated during downward reach and pick-up tasks such as gardening, doing laundry, or putting groceries away in low cabinets. A lot of difficulty or being unable to stoop, crouch, or kneel is prevalent in adults over the age of 65, as evidenced by a 25% incidence in the baseline data of the established populations for epidemiologic studies of the elderly (Taylor et al., 1993). However, lifting strategies in affected older adults have not been well characterized (Puniello et al., 2000).

The importance of stooping, crouching or kneeling, is evidenced by the use of downward reach-pick-up tasks in several fall risk batteries. In the physical performance test, the penny pick-up was introduced, which consisted of a penny being placed a foot in front of the participant, and the measurement of the time required to carry out the task with no restrictions on bending strategies (Reuben and Siu, 1990). In a questionnaire of activities-specific balance confidence (Powell and Myers, 1995), participants are questioned about their confidence to not lose their balance or become unsteady while picking up a slipper from the floor. Furthermore, the Berg Balance Scale (BBS) considers a task where a slipper is placed in front of the person, and the ability to independently perform the slipper pick-up is evaluated (Berg et al., 1992).

In community-dwelling adults, difficulty bending down to pick up an object from the floor has been found to be associated with increased fall risk (O'Loughlin et al., 1993). Additionally, in a case-control study of older adults, a comparison of four functional tests of fall risk, the BBS, Tinetti Mobility Score, Elderly Mobility Scale and Timed Up and Go (TUG) Test, found the BBS items of picking up a slipper from the floor and standing on one leg to be the most significant tasks in the discrimination of fallers from non-fallers (Chiu et al., 2003). Thus, difficulty in downward reach-pick-up (DRPU) tasks may be indicative of an increased risk of falling amongst older adults.

Increased self-reported stooping, crouching, or kneeling difficulty is independently associated with decreased activities specific balance confidence, self-reported leg joint limitations, and decreased isometric knee extension strength (Hernandez et al., 2008). Older adults with low balance confidence (i.e., confidence in the ability to maintain balance and avoid a fall) would be expected to have a greater difficulty in performing a DRPU, as this task would require significant coordination of the whole body over a wide range of postures to maintain balance, and balance confidence is associated with fall risk (Carpenter et al., 2006). In adults with pain or stiffness-induced joint limitations, SCK movements may be limited (Edmon and Felson, 2003); thereby compensatory DRPU strategies are expected to arise as a means to cope with day-to-day limitations. Furthermore, older adults with weak hip and knee extensors rely primarily on a back dominant lifting strategy, typically referred to as stooping (Puniello et al., 2001). Thus, stooping, crouching, or kneeling difficulty is expected to lead to compensatory lifting strategies, as a means to cope with limitations with balance, knee strength, and leg pain or stiffness.

This study tested the hypothesis that compared to women with no self-reported stooping, crouching, or kneeling difficulty, those with self-reported stooping, crouching, or kneeling difficulty use (a) an asymmetric stance, (b) a wider base of support, (c) a closer stance to the object, and (d) more stooping when picking up an object from the floor.

C.3. Methods

C.3.1. Participants

In this study, participants consisted of women aged 65 years or older recruited, primarily, from a congregate housing facility in southeastern Michigan (n=16). Participants underwent a brief screening to exclude individuals that had medically unstable conditions (e.g., chest pain upon exertion, dyspnea, recent fracture, acute infection) or reported mobility limitations (e.g., dependent upon physical assistance or wheelchair). Participants were recruited into two groups based on their response to a single question on the Established Populations for Epidemiologic Studies of the Elderly (EPESE) questionnaire (Smith et al., 1990). Participants rated their ability to stoop, crouch, or kneel according to a five point difficulty scale. Those who reported having "a lot of difficulty" or being "unable to do" stooping, crouching, or kneeling tasks were categorized as having Self-Reported Difficulty (n=8) and those who reported no, a little, or some stooping, crouching, and kneeling difficulty were categorized as having No Self-Reported Difficulty (n=8).

C.3.2. Data Collection

Self-report health: Self-report (interview-based) measures were taken during the medical screening. Self-reported balance confidence was ascertained with the use of the Activities-specific Balance Confidence Scale (ABC Scale; range 0-100) (Powell and Myers, 1995). Self-reported leg joint limitations were determined by the report of pain or stiffness in the hip, knee, or ankle.

Unipedal Stance Time: The unipedal stance time was used as a measure of static balance (Vellas, 1997). Participants stood on their dominant leg, and lifted their other leg a few inches above the ground, with their arms folded across their chest. A practice trial preceded two measured trials, and the best trial constituted the UST.

Downward Reach Pick-Up: Participants stood in front of a cardboard box that was placed 1.8 m away. The box had a mass of 0.6 kg and dimensions of 25.4 cm in height, 30.5 cm in width, and 38.1 cm in depth with three handles, two on the sides for bilateral tasks and one on the top for unilateral tasks. Participants were instructed to pick up the box in front of them, using one or two hands, without any restrictions on the use of a lifting strategy. Participants performed one trial of a native unilateral or bilateral downward reach pick-up task. Two digital camcorders were used for video recording of downward reach pick-up tasks in the frontal and sagittal planes for use in lifting strategy analyses.

C.3.3. Data Analysis

To quantify the lifting characteristics of older women with and without SCK difficulty, we examined both the stance and lifting strategy employed at the initiation of the lifting movement. To quantify the type of base of support stance, the incidence rate of an asymmetric stance; the base of support width, in relation to the participant's shoulder width, and the distance from the edge of the BOS to the object, as a percentage of foot length, were measured (**Figure C.1**). To analyze lifting strategies, the postural index, a measurement introduced by Burgess-Limerick and Abernethy in 1997, is considered. The postural index is defined as the difference in knee flexion angle from normal standing at the moment of lift, divided by the sum of ankle, hip, and lumbar vertebral flexion from normal standing. As seen in **Figure C.2**, lower postural indices suggest more of a stoop, or back dominant strategy, whereas postural indices closer to one suggest more of a crouch, or leg dominant strategy.

Digital film footage was converted into a series of still-frame photos at the time of pick-up. Two undergraduate students were trained to rate video footage for the desired dependent variables. Statistical analysis consisted of t-test and Fisher's exact test, for testing group differences in quantitative and qualitative dependent variables, respectively.

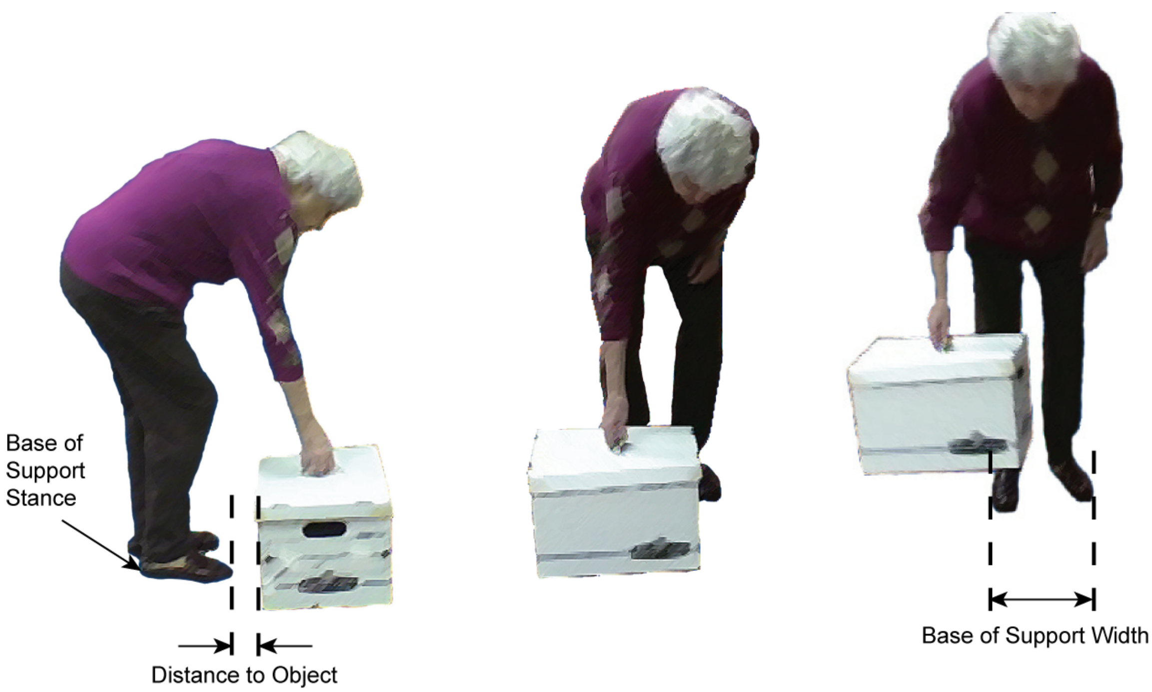

Figure C.1: Illustration of the outcome measures used to quantify the type of base of support stance: the base of support width and the distance from the edge of the BOS to the object.

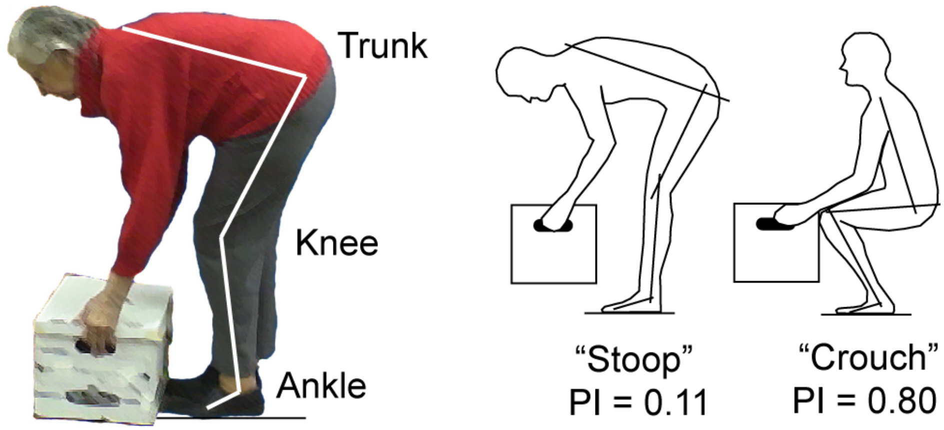

Figure C.2: (Left) Modified measurement of the postural index (Burgess-Limerick and Abernethy, 1997) considering ankle, knee, and trunk (i.e., hip and lumbar vertebral) angles. (Right) Illustrations of a stoop or crouch, and their respective postural indices.

C.4. Results

Characteristics of participants with self-reported difficulty and no self-reported difficulty are shown in **Table C.1**. On average, women who reported stooping, crouching, or kneeling difficulty, demonstrated trends toward a lower activities-specific balance confidence and a higher incidence of self-reported leg joint limitations due to pain and stiffness. No significant differences were observed in age, body mass index, or best unipedal stance time.

Table C.1: Characteristics of older women with and without self-reported stooping, crouching, or kneeling difficulty.

Characteristics	No Self-Reported Difficulty (N=8)	Self Reported Difficulty (N=8)	p-value
Age (yrs)	76 (9)	78 (10)	0.64
Mean BMI (kg/m2)	28 (4)	30 (6)	0.415
ABC (0-100 scale)	80 (13)	65 (21)	0.102
Best Unipedal Stance Time (s)	14 (11)	7 (9)	0.236
Leg Joint Limitations (%)	25%	63%	0.157

Lifting Strategy: As seen in **Figure C.3**, more stooping, as suggested by the increased postural index, is seen in older women with self-reported SCK difficulty in single-handed tasks, when compared to women without self-reported SCK difficulty (P<.05). Single-handed tasks demonstrated trends towards an increased use of stooping lifting strategies when compared to two-handed tasks (P<.10).

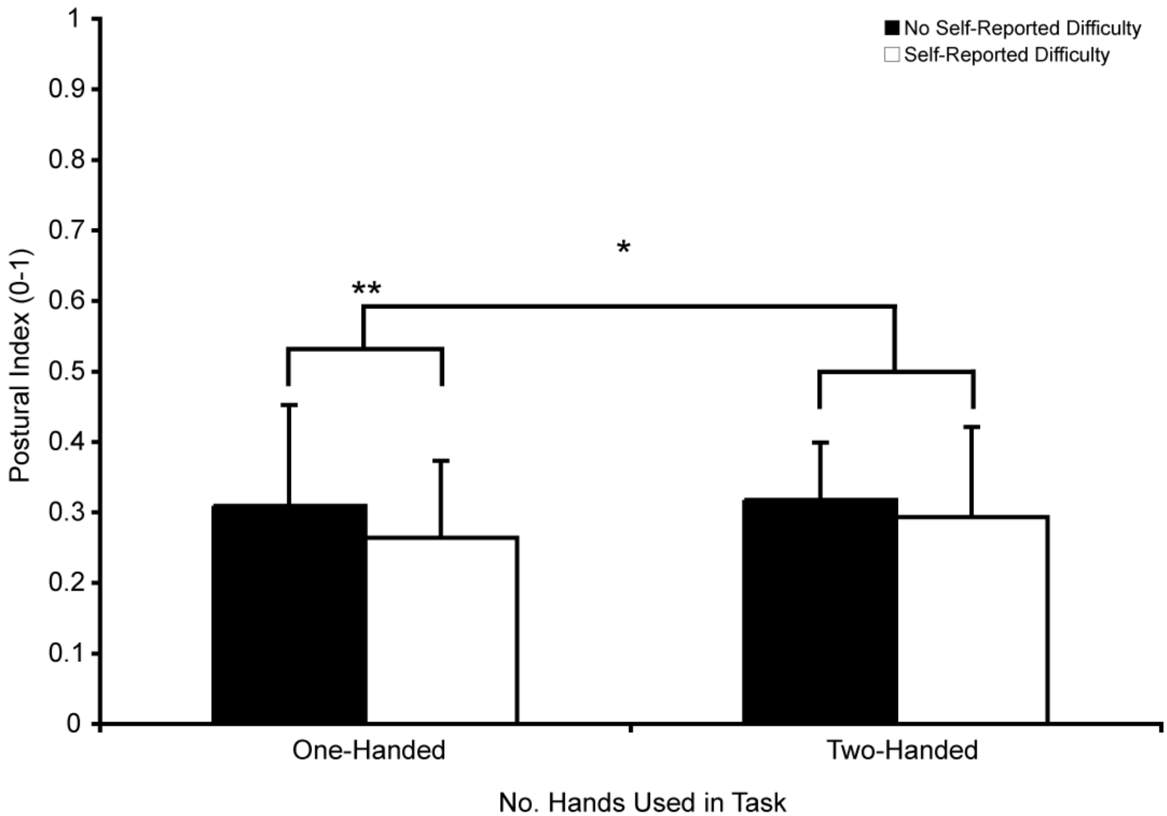

Figure C.3: Mean (SD) postural indices for older women with (black bars) or without (white bars) self-reported stooping, crouching, or kneeling difficulty. Comparisons between tasks and groups were achieved using t-tests (*P ≤ .10, and **P ≤ .05).

Base of Support Stance: Older women with self-reported SCK difficulty demonstrated an increased distance to the object in single-handed trials (.42 to .21), and a reversal in trends in two-handed trials (.18 to .27) (**Figure C.4**). Between tasks, no consistent trends are observed. In single-handed tasks, older women with self-reported SCK difficulty demonstrated an increased use of a wide base of support and no difference in the use of an asymmetric stance at pick-up of the box (**Table C.2**). However, no statistically significant differences were seen. In two-handed tasks, older women with self-reported SCK difficulty had trends toward an increase in the use of a wide base of support ($p = .1$), but no significant differences in the incidence rate of asymmetric stances. However, between tasks, trends of a decreased use of an asymmetric stance in a two-handed tasks was seen, going from an average of 50% to 31% in the use of an asymmetric stance (.50 to .31, $p = .133$).

Table C.2: Lifting strategy characteristics in (A) single-handed and (B) two-handed downward reach and pick-up trials.

(A) Single-handed reach results

Lifting Strategy Characteristics	No Self-Reported Difficulty (N=8)	Self Reported Difficulty (N=8)	p-value
Base of Support Width (% > shoulder width)	0%	25%	0.233
Base of Support Stance (% assymetric stance)	50%	50%	0.69

(B) Two-handed reach results

Lifting Strategy Characteristics	No Self-Reported Difficulty (N=8)	Self Reported Difficulty (N=8)	p-value
Base of Support Width (% > shoulder width)	0%	38%	0.1
Base of Support Stance (% assymetric stance)	38%	25%	0.5

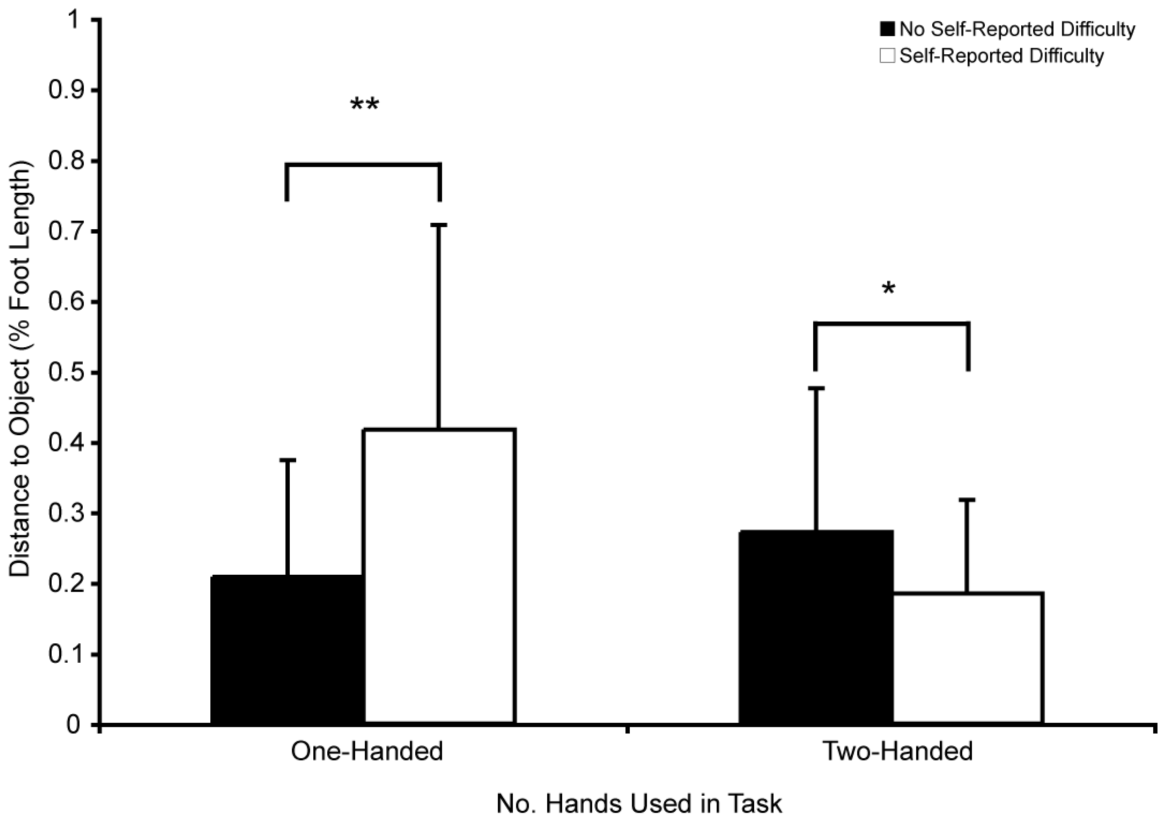

Figure C.4: Comparison of distance to object, from edge of base of support, in older women with and without self-reported stooping, crouching, or kneeling difficulty. Means and standard deviations (SD) of distance to object as a percentage of foot length. Comparisons between tasks and groups were achieved using t-tests (*P ≤ 0.1, **P ≤ 0.05).

C.5. Discussion

In summary, when testing the hypothesis that compared to women with no self-reported stooping, crouching, or kneeling difficulty, those with self-reported SCK difficulty use an asymmetric stance, wide base of support, closer stance to the object, and more stooping; we found no evidence for group differences in use of an asymmetric stance, or closer stance to the object, but found supporting evidence of the use of a wide base of support and more stooping when picking up an object from the floor.

Differences between women with and without SCK difficulty in their respective lifting strategies may be primarily attributed to decreased knee flexion and the increased incidence of leg joint limitations. Complimentary to previous studies, the self-reported leg joint limitations could explain the increased reliance of a back-dominant strategy (Puniello et al., 2001).

Among the strengths of this pilot study, we considered that this preliminary work is one of the only studies of stooped lifting and reaching in an affected population; its use of semi-quantitative measurements provides additional information in comparison to existing qualitative assessments of time and ability, and the use of blinded raters helps control for possible bias in the measurement of dependent variables. Limitations of this pilot study include its small sample size, and low difficulty of tasks, which include the large size of the target and the relatively small demand on range of motion to achieve the task.

C.6. Conclusions

In conclusion, single-handed tasks discriminated better between the groups than two-handed tasks, further supporting the use of one-handed tasks in clinical fall risk batteries, and those with self-reported SCK difficulty stooped more and user a wider stance to pick up an object from the floor.

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