THE CONTRIBUTION OF ARM MOTION DURING THE FEET-IN-PLACE FALL-RECOVERY RESPONSE

by

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ABSTRACT

Falls are the leading cause of non-fatal, unintentional injury in the U.S. In at-risk groups, such as those with chronic stroke, more than a third of falls are due to externally applied perturbations. In order to maintain stability after such a perturbation, an individual can use different biomechanical mechanisms to regain stability. In this series of studies, we are interested in the mechanism of rotating segments about the center of mass, as commonly done by rapidly moving the arms. We are specifically interested in the ability to recover from a posterior disturbance without taking a step, as this ability has been prospectively related to falls in older adults and can be quantified reliably using single-stepping thresholds. When responding to a posterior disturbance, elevating the arms through shoulder flexion shifts the center of mass anteriorly and applies a reactive moment to the rest of the body that resists rotation of the fall. We do not know, however, if arm motion meaningfully influences the posterior single-stepping threshold measure. For those with chronic stroke, impaired arm function is associated with a greater risk of falls. It is unknown, however, if chronic stroke alters arm rotation during posterior fall recovery. The aims of this project are to specifically 1) quantify how constraints in arm motion alter the response to a posterior postural disturbance, and then 2) determine if individuals with chronic stroke have a diminished, asymmetrical arm response.
To address the Aim 1 hypothesis, ten young, adults with no impairment were recruited for this study. Participants attempted to prevent steps in response to a progressive series of rapid, precise treadmill belt accelerations delivered by a computer-controlled treadmill. The posterior single-stepping thresholds, as represented by the disturbance magnitudes that consistently elicited a step in each direction, were determined for each participant. Kinematics were recorded to quantify the resulting dynamic stability as a means to explain a potential underlying mechanism of hypothesized group differences. When the arms were constrained, lower posterior single-stepping thresholds were observed (p=0.03 Cohen’s $d=0.86$). When evaluating dynamic stability at the highest disturbance levels, the $MoS_{min}$ was not different between unconstrained conditions arms–constrained conditions ($p = 0.55$, Cohen’s $d = 0.20$). Considering that the unconstrained responses involved larger disturbances, this lack of differences in the minimum margin of stability suggests that the active response to the perturbation were impaired in the arms constrained condition.

To address the Aim 2 hypothesis, ten individuals with chronic completed the same single-stepping threshold assessment, and we compared results to the unconstrained condition of Aim 1. Peak shoulder flexion velocities as well as a symmetry index were compared between groups. Those with chronic stroke displayed more asymmetry than the unimpaired participants ($p = 0.005$, Cohen’s $d=1.34$). We did not, however, detect meaningful between-group differences in shoulder motion between unimpaired and impaired individuals ($p > 0.20$, Glass's $\Delta < 0.49$).
In conclusion, these results demonstrate that arm motion does play a role in posterior fall performance, as characterized by single-stepping thresholds. When applying this concept to those with chronic stroke, a population that has a high fall risk, it was apparent that the arm response was asymmetrical, perhaps altering its contribution to fall recovery. Moving forward, this information justifies consideration of the arm response as a potentially modifiable aspect of the fall recovery response.
Chapter 1

INTRODUCTION

1.1 Overview

At 9.4 million injuries per year, falls are the leading cause of non-fatal, unintentional injury in the U.S. [1]. At-risk groups include older adults and individuals with chronic stroke, populations with annual fall incidences that exceed one in four people [1], [2]. In older adults, one out of every five falls is associated with a serious injury such as a bone fracture or head trauma [3], [4]. In those individuals with stroke, up to 55% of falls require medical attention, and nearly one out of every five is associated with fracture, joint dislocation, or cerebral hemorrhage [5]–[7]. Within these groups, more than a third of falls are due to external disturbances [5], [8], [9], such as that caused by a trip or a slip. Thus, the ability to recover from external disturbances is likely an underlying factor of the high fall incidence and risk of fall-related injury in these groups [1], [10].

In laboratory studies, the ability to recover from an external disturbance is impaired with older age or chronic stroke [11], [12]. Although fall-recovery performance is reduced, the underlying biomechanical factors that influence these deficits are not fully understood. In order to maintain stability after a disturbance, an individual can use three mechanisms to regain stability [13]. The first mechanism involves moving the center of pressure (CoP) location under the base of support, an action that can be achieved by generating moments about the ankle [14] or by taking a step [15]. Second, a person can rotate segments about the center of mass (CoM), as commonly done through flexion or extension of the hips [14] or by rapidly moving the
arms [16]–[20]. Third, external forces can be applied, as evident when individuals use a handrail or safety harness.

We are specifically interested in the ability to recover from a posterior disturbance without taking a step, as this ability has been prospectively related to falls in older adults [21] and can be quantified using single-stepping thresholds, or the disturbance magnitudes that consistently evoke a step [22]. With this task of a non-stepping response, the limited base of support relative to the ankle joint center limits the efficacy of moving the CoP to regain stability (i.e. Mechanism I), in turn increasing reliance on the second mechanism of segment rotation. When responding to a posterior disturbance, elevating the arms through shoulder flexion shifts the center of mass anteriorly [16] and applies a reactive moment to the rest of the body [23] that resists the posterior rotation of the fall. Alternatively, an opposite, protective strategy is often used by older adults as a likely attempt to reduce injury severity from fall impacts, taking the initial fall impact with the extremities as opposed to the pelvis, trunk, and/or head [24]. Another effect of age can be observed during the posterior stepping response of slip recovery, where young adults use upper extremity motion that benefits trunk rotation to a greater extent than that of older adults [19]. We do not know, however, if arm motion meaningfully contributes to the posterior stepping threshold measure.

Balance reactions of those with chronic stroke have been characterized by altered lower-extremity muscle activation, smaller stepping thresholds, and less-effective step kinematics [25]–[36]. Given that fall-recovery is often characterized as a whole-body response [37], it is surprising that so little has been done in evaluating how stroke alters arm motion during fall-recovery, especially given that in unimpaired
individuals, constraining the arms limits the ability recover balance with a feet-in-place response (See chapter 2) [20]. The response of older adults, another at-risk population, often demonstrate a protective strategy of shoulder flexion of the arms in the response to an anteroposterior fall [24]. For those with chronic stroke, impaired arm function is associated with a greater risk of falls [38]. This association could be a result of an inability to use arm motion to arrest the fall. It has not been investigated, however, if chronic stroke alters arm rotation during fall recovery. By addressing these unknowns, we can determine if arm motion may serve as a target for improving fall-recovery and, subsequently, reducing the risk of falls.

1.2 Significance

The contribution of this study will be to specifically quantify how constraints in arm motion alter the response to a posterior postural disturbance, and then determine if individuals with chronic stroke have a diminished, asymmetrical arm response. This contribution is significant because it will improve our general understanding of how adults maintain standing balance after a perturbation and it will identify a potential target for interventions to improve balance in individuals with chronic stroke. If constraining the arms significantly reduces the ability to recover from a posterior fall, then we have justification for focusing on how age, neuromuscular impairment, or interventions alter arm motion during posterior fall recovery. If individuals with chronic stroke have reduced arm motion after a posterior perturbation, then we can begin to determine if that limited arm motion can serve as a modifiable target for improving the fall recovery response. This information will inform task-specific, perturbation based interventions [39], [40] to improve fall recovery and, potentially, reduce fall rates.
1.3 Innovation

Evidence suggests that coordinated arm motion can influence fall-recovery. This role of arm motion has been observed for the stepping [16–18], [41] and feet-in-place responses to anterior falls [18], [20], [24], as well as the stepping response to a posterior fall [16], [19], [42], [43]. What is not known is how the arms contribute to posterior feet-in-place response, a skill that has clinical relevance given its documented, prospective relationship with falls in older adults [21]. Other studies have assessed the role of arm motion with a limited scale of disturbance magnitudes (e.g. three magnitudes) [20], [41], an approach with restricted sensitivity in detecting the benefits of arm motion. Finally, the role of arm motion has been determined by experimentally constraining the arms [20], [41] or by evaluating the arm response of older adults [44], [45]. We do not know the effects of chronic stroke on arm motion, despite reduced arm function being reported by individuals with stroke that experienced multiple falls [38]. The proposed work is innovative because, unlike previous studies, it addresses how the arms contribute to posterior feet-in-place fall-recovery, an outcome that has been shown to have clinical relevance. It is also innovative because we are evaluating a component of the balance reaction in those with chronic stroke that has not been studied before, despite an initial indication that the arms are correlated to fall risk. Through the evaluation of how limited arm motion affects single-stepping thresholds, we can objectively quantify how arm motion alters fall-recovery ability within a promising assessment of fall risk. These aims represent translational work by applying this measure in experimentally manipulated (i.e. basic) and clinically relevant (i.e. applied) contexts of altered arm motion. This assessment, when combined with motion analysis, also allows us to evaluate the role of arm
motion on dynamic stability maintenance. Such analysis may provide a biomechanical explanation of how constraints in arm motion alter fall-recovery ability.
Chapter 2

CONSTRAINING THE ARMS REDUCES POSTERIOR, BUT NOT ANTERIOR SINGLE-STEPPING_THRESHOLDS IN YOUNG, UNIMPAIRED ADULTS

2.1 Introduction

Falls are the leading cause of non-fatal, unintentional injury in the United States, exceeding the next-leading cause (being struck by or against an object) by 91% [46]. It is estimated that, with inflation, direct medical costs for fall injuries will be $50 billion per year [47]. In addition to injury, a fall can lead to limited activity due to a fear of falling [48], [49], which subsequently results in a loss of physical capabilities and an increased risk of falling [50]–[52].

Falls are often caused by external disturbances, defined here as perturbations that are caused by an interaction with the surrounding environment [5], [8], [9], [53]–[55]. These disturbances include slips and trips associated with poor stairway design, inadequate lighting, clutter, slippery or uneven floors, or unsecured mats and rugs [56]. Conversely, causes of internal disturbances include syncope, dizziness, pain, or lower extremity instability. Although it may be possible to reduce exposure to external disturbances through behavioral or environmental modifications, complete avoidance of risk factors is not practical, and attempts to do so may lead to the aforementioned, negative consequences of physical activity avoidance. Alternatively, interventions may focus on the specific skill of recovering from an external disturbance should it occur. Successful recovery from a fall, however, requires rapid coordination of the whole-body response [37]. Given this complex task, we must first identify which aspects of this response contribute to fall-recovery success. These aspects, then, can
serve as candidate targets for interventions to improve balance reactions and, subsequently, reduce the risk of falls.

There are three biomechanical mechanisms by which an individual can maintain balance after a postural disturbance [57]. These mechanisms, as determined by a theoretical inverted pendulum model, include (M1) moving the center of pressure within the base of support, (M2) counter-rotating segments around the whole-body center of mass, and (M3) applying an external force [13]. In the absence of an external force (M3; e.g. that from a safety harness or handrail), a standing individual must rely on the first two mechanisms to regain stability. After an anterior or posterior perturbation, the first mechanism of moving the center of pressure (M1) can be achieved by generating moments about the ankle joint (i.e. “ankle strategy”) [14] or, should the disturbance magnitude be large, extending the base of support with a step [58]. The second strategy of counter-rotating segments about the center of mass (M2) often supplements this first strategy. In order to prevent a step, a “hip strategy” of hip flexion or extension can aid recovery in response to anteroposterior disturbances [14]. Previous studies have focused on the role of ankle strategies, hip strategies, and step kinematics during fall recovery [14], [59], [60]. An aspect given less attention is how reactive arm motion, another form of counter-rotation (M2), contributes to stabilization. Given that the arms represent 10% of the total body mass [61] that can be accelerated rapidly, a focus on how arm reactions contribute to fall recovery is warranted.

Previous studies suggest that rapid arm motion can indeed contribute to balance reactions. Arm motion during stepping responses help to limit whole-body rotation and COM translation during the recovery response [16]–[19], [41], [43].
When a step is needed to maintain balance during an anterior fall, young adults demonstrate a relative posterior motion of the arms (i.e. shoulder extension) that counters the induced, forward angular momentum of the rest of the body [17], [18]. Older adults, a group at high risk of falls and fall-related injury[3], [4], [62]–[64], employ an opposite, protective strategy of shoulder flexion likely in an attempt to reduce injury severity from fall impacts [18]. In work that experimentally constrained arm motion during an anterior stepping response, prohibiting arm motion was associated with a faster rate of falling, shortening the available preparation period for stepping [41]. A limited ability to recover from an anterior disturbance was also observed during a feet-in-place recovery, a task constraint that should increase reliance on counter-rotation mechanisms. When arm motion was constrained, steps (i.e. a failed response) were more prevalent and the rate of stabilization after the perturbation was slower [20].

Older age also appears to have an influence on arm motion in response to a posterior fall. During the stepping response, young and older adults elevate the arms in an anterior (shoulder flexion) and outward (circumduction) motion [16], [24], [43]. This elevation strategy shifts the center of mass anteriorly [16], [65] and counters angular momentum of the fall, as evident by a reduction in trunk extension velocity [60]. Although such beneficial effects on trunk motion were observed in both young and older adults, a larger reduction in trunk extension velocity was evident in the younger group [60]. Other work confirms the effect of age on the functional use of the arms in posterior fall recovery, as older adults displayed a delayed shoulder flexion moment during a stepping response [42]. During feet-in-place recovery to posterior perturbations, young and middle-aged groups demonstrated initial arm movements
that were in the same direction as trunk motion as a means to regain stability, while older adults displayed arm motion in the direction of the fall [24]. As with anterior-fall recovery, this age-related difference in arm motion may reflect a shift from a compensatory strategy of balance recovery to a protective strategy of modifying fall impact [18], [24], [45]. Regardless of the response intent, previous work confirms that this muscle activity generated by the upper extremities is functionally-driven, and not the result of a startling response [16], [17], [66].

Although there is an apparent effect of age on arm motion during the feet-in-place-response to posterior falls, the degree to which such motion benefits fall-recovery performance in this context is not known. Because the arms play a meaningful role in anterior feet-in-place responses [20], as well as posterior stepping responses [19], [42], [43], it is likely that the arms will also contribute to posterior feet-in-place responses. The specific posterior fall-recovery response is worthy of consideration because 1) compared to that of anterior fall recovery, anatomical constraints likely limit the effectiveness of the ankle and hip strategies during posterior fall recovery [67], and 2) the ability to recover from a posterior disturbance without a step has been prospectively related to falls in older adults in our previous studies (In Review) and other published work [21]. Conversely, in both our work and that of others, the ability to prevent a step in response to an anterior fall, as well as the ability to limit the posterior response to a single step, were not prospectively related to falls. Posterior single-stepping thresholds, or the disturbance magnitudes that consistently evoke a backwards step, are a reliable means of quantifying the ability to recover with a feet-in-place response [22]. It is not known, however, if arm motion plays a meaningful role in this evaluation.
The purpose of this study was to determine the effects of arm-motion constraints on posterior single-stepping thresholds in young, non-impaired adults. We hypothesized that arm-motion constraints would reduce such thresholds. In order to investigate possible underlying mechanisms of our hypothesis, we evaluated the resulting dynamic stability of the fall-recovery response, as measured by the margin of stability [68]. Briefly, this measure accounts for the center of mass position and velocity relative to the base of support boundary. We anticipated that, at a common disturbance magnitude, constraining arm motion would be characterized by more instability (i.e. a more negative margin of stability).

2.2 Methods

Ten adults with no self-reported neuromuscular impairment or previous injury that limits motion were recruited for this study (mean ± s.d. [range] age = 21.7±1.6 [18 – 23] years; BMI = 22.53 ± 2.96 [19.17-29.62] kg/m²). From preliminary data, between-condition effects of six participants per group (Cohen’s d = 1.16), ten participants per group provided substantial power (1-β = 0.90) for this study. In the process of evaluating posterior stepping thresholds, anterior single-stepping thresholds are also determined [22]. Although not the primary focus of this study, we evaluated the effects of arm-constraints on anterior thresholds in order to confirm the aforementioned effects of arm-constrains on anterior feet-in-place fall-recovery [20], [41]. Of note, we observed small preliminary effects in the anterior direction (Cohen’s $d = 0.27$). So, we did not anticipate nor were we powered to detect significant between-group differences in anterior thresholds. This study was approved by the University of Delaware institutional review board and all participants provided informed consent prior to participation.
Participants were outfitted with a safety harness attached to an overhead rail. They stood on a computer-controlled treadmill (ActiveStep®, Simbex, Lebanon, NH, Figure 1), and the height of the harness was minimally elevated so that the participant’s knees and hands could not come into contact with the treadmill. Participants were instructed to “try to prevent a step” in response to rapid, 400 ms belt translations [22]. Initial belt accelerations began at 0.5 m/s², lasting a period of 200 ms followed by a 200 ms deceleration phase, in total resulting in a 2 cm total displacement. For subsequent trials, the initial accelerations were increased or not changed by 0.5 m/s² depending on the response success or failure, respectively. A failed attempt was defined as when the participant took a step or when the force transducer (Dillon, Fairmont, MN) between the harness and overhead rail recorded more than 20% of the participant’s body weight. This force threshold is low or comparable to that of other perturbation studies [69]. The disturbance direction and timing of disturbances were pseudorandomized so that, at most, three consecutive disturbances were delivered in the same direction. Each disturbance was preceded by a 3 to 10 second delay [22]. This variability in direction and timing was intended to discourage a pre-planned response or initial postural orientation that favored certain disturbance directions and timing. Here, anterior and posterior refer to the direction of the fall, not the direction of the treadmill belt translation (i.e. “anterior disturbance” is a posterior belt translation that elicits an anterior fall).

Stepping thresholds, defined as the disturbance magnitude that elicited four failed responses in a given direction, were determined in the anterior and posterior directions [22]. Using a simple inverted pendulum model, thresholds were expressed as the destabilizing torque (τ) at the base of an inverted pendulum [22], [44]:

\[ \tau = |m \cdot a \cdot l| \]
where $m$ is body mass, $a$ is the initial belt acceleration, and $l$ is the inverted pendulum length ($0.586 \times$ height) [61]. The progression to identify anterior and posterior stepping thresholds was conducted under two conditions: first unconstrained (Figure 1A, mean ± S.D. number of trials= 24 ± 5, range = 16-33) and, second, with the arms constrained by having participants hold a strap behind their body (Figure 1B, number of trials= 22 ± 5, range 16-32).

Figure 1. (A) A participant successfully responds to a posterior disturbance (a=2.5 m/s²) in the unconstrained condition. (B) The same participant successfully responds to a disturbance of the same magnitude with the arms constrained by holding a rope behind the pelvis.
All trials were recorded with a 12-camera motion capture system operating at 120 or 240 Hz (Motion Analysis®, Santa Rosa, CA, replaced mid-study with Qualisys®, Göteborg, Sweden). Thirty-five passive-reflective markers facilitated the definition of 13 body segments: head/neck, trunk, pelvis, upper arms, forearms, thighs, shanks and feet. Marker trajectories were filtered via a fourth-order Butterworth filter with a low pass 6 Hz cutoff.

Dynamic stability was evaluated from the center of mass position and velocity relative to the base of support, quantified as the margin of stability (MoS) [68] as follows:

$$\text{MoS} = d + \frac{v}{\sqrt{g/l}} - \text{BoS}$$

where $d = x_{\text{COM}} - x_{\text{heel}}$ and $v = v_{\text{COM}} - v_{\text{heel}}$ (i.e. the positions and velocities of the CoM relative to edge of the base of support). The term $g$ represents gravity ($9.81 \text{ m/s}^2$), $l$ is the instantaneous distance between the mean ankle joint center and whole-body center of mass [61], representing an inverted pendulum length, and $\text{BoS}$ represents the position of the relevant edge (anterior or posterior) of the base of support. The MoS was calculated as a continuous measure for all perturbation responses, and the minimum value (i.e. most negative value) was identified for each response (Figure 2). All calculations of dynamic stability were done using custom software (LabVIEW, National Instruments®, Austin, TX).

In order to determine if lower stepping thresholds and dynamic stability measures were lower in the arms-constrained condition as compared to the unconstrained condition, an analysis using dependent t-tests (SPSS v24, IBM, Armonk, NY; $\alpha=0.05$) and effects sizes (Cohen’s $d$ for repeated measures were used
Two planned comparisons of $MoS_{min}$ were conducted across conditions. First, the $MoS_{min}$ was evaluated for the successful response to the largest disturbance within each condition (i.e. one level below stepping threshold values). This comparison provided insight on the maximum instability from which participants could recover. Second, the $MoS_{min}$ was evaluated for the successful response to the largest disturbance magnitude common to both conditions. This comparison provided insight on the resulting stability from a theoretically challenging, common disturbance magnitude.

Figure 2. Depiction of posterior margin of stability for a participant successfully recovering from an anterior perturbation (accel. = 2 m/s$^2$ with arms unconstrained). Above zero indicates the participant is dynamically “stable”. Below zero represents dynamic instability, or a situation in which a fall cannot be prevented by only altering the center of pressure location alone.

2.3 Results

Posterior Fall Recovery Responses

Single-stepping thresholds when the arms were unconstrained (250 ±73 N·m) were significantly larger than when the arms were constrained (change in thresholds = 23 ± 28 N·m, $p = 0.03$, Cohen’s $d = 0.86$, Figure 3A), supporting our hypothesis that
arm constraints would reduce feet-in-place fall recovery performance. With the arms constrained, thresholds were reduced by 9.6 ± 14.0%. At the highest levels from which the participant was able to successfully recover (i.e. near-threshold levels), the $MoS_{min}$ was not different in unconstrained conditions ($MoS_{min} = -8.2 \pm 5.1$ cm) compared to that with the arms constrained (change in $MoS_{min} = -0.9 \pm 3.0$ cm, $p = 0.38$, Cohen’s $d = 0.31$, Figure 3B). At the highest common disturbance magnitude, the unconstrained ($MoS_{min} = -8.2 \pm 5.1$ cm) and constrained conditions (change in $MoS_{min} = -0.8 \pm 3.7$ cm), did not exhibit different minimum margins of stability ($p = 0.36$, Cohen’s $d = 0.31$, Figure 3C).

Figure 3. (A) Posterior single-stepping thresholds in the unconstrained condition and arms-constrained conditions. (B) Minimum margin of stability values ($MoS_{min}$) in response to the highest disturbance magnitude (i.e. typically larger disturbances in the unconstrained condition) within the unconstrained and arms-constrained conditions. (C) Minimum margin of stability values ($MoS_{min}$) in response to the highest common disturbance. Individual participants are represented by unique colors/symbols.
Anterior Fall Recovery Responses

Single-stepping thresholds when the arms were unconstrained (311 ± 75 N·m) were not significantly different than when the arms were constrained (change in thresholds = 5 ± 35 N·m, p=0.69, Cohen’s d = 0.14, Figure 4A). At the highest levels from which the participant was able to successfully recover, the MoS$_{min}$ was not different in unconstrained conditions (-20.4 ± 10.0 cm) compared to that with the arms constrained (change in MoS$_{min}$ = -0.8 ± 3.7 cm, p = 0.55, Cohen’s d = 0.20, Figure 4B). At the highest common disturbance magnitude between the unconstrained (-19.4 ± 7.7 cm) and unconstrained conditions (change in MoS$_{min}$ = -1.5 ± 4.8 cm), the MoS$_{min}$ was not significantly different between conditions (p = 0.35, Cohen’s d =0.40, Figure 4C).

Figure 4. (A) Anterior single-stepping thresholds in the unconstrained condition and arms-constrained conditions. (B) Minimum margin of stability values (MoS$_{min}$) in response to the highest disturbance magnitude (i.e. typically larger disturbances in the unconstrained condition) within the unconstrained and arms-constrained conditions. (C) Minimum margin of stability values (MoS$_{min}$) in response to the highest common disturbance. Individual participants are represented by the same unique colors/symbols, as in Figure 3.
Kinematics

A comparison of average peak shoulder flexion/extension velocities across conditions confirmed that arm motion was indeed reduced during anterior (unconstrained peak shoulder extension velocity: 109 ± 76 deg/s, constrained: 47 ± 64, p=0.0009, Cohen’s $d=1.60$) and posterior (unconstrained peak shoulder extension: 118 ± 83 deg/s, constrained: 32 ± 9 deg/s, p=0.004, Cohen’s $d=1.00$) balance reactions.

2.4 Discussion

The purpose of this study was to determine the effects of arm-motion constraints on posterior single-stepping thresholds in young, non-impaired adults. We hypothesized that arm-motion constraints would reduce feet-in-place fall recovery performance, indicated by stepping thresholds. This hypothesis was supported, as thresholds were reduced with arm constraints (Figure 3A). Such constraints, however, did not alter anterior stepping thresholds (Figure 4A), contrary to what previous literature had reported on the anterior feet-in-place fall-recovery response.

Posterior stepping thresholds were reduced by 10%, on average, when constraining the arms (Figure 3A). Although shoulder flexion has been associated with improved fall recovery kinematics [16], [65],[60], this is the first study, to our knowledge, to quantify how a reliable, objective metric of posterior fall-recovery ability is reduced with arm motion constraints. To evaluate potential underlying factors of this observation, dynamic stability was analyzed at near-threshold trials under both conditions. At these often disparate levels (i.e. often a larger disturbance in the unconstrained condition), participants did not have meaningfully different levels of instability. One interpretation of this result is that, when participants had their arms constrained, smaller disturbance magnitudes resulted in a level of instability.
characteristic of near-threshold responses. In other words, the active response to the destabilizing effects of the perturbation were impaired with arms constrained. When evaluating MoS\textsubscript{min} at common disturbance magnitudes, no between-condition effects were observed for the MoS\textsubscript{min} (Figures 3C & 4C). We had anticipated that use of the arms would result in less instability. Perhaps the disturbance magnitudes were not challenging enough to elicit a beneficial arm response in unconstrained conditions. Much like how a “hip strategy” is incorporated with the “ankle strategy” in response to larger disturbance magnitudes [14], it could be that an “arm strategy” is then incorporated in response to even larger disturbances.

Unlike the effects on posterior single-stepping thresholds, anterior thresholds were not altered by arm constraints (Figure 4A). Similarly, such arm constraints did not alter the resulting dynamic stability of the disturbance (Figures 4B-4C). This observed lack of effect in forward fall recovery is in partial disagreement with observations from using a forward lean-release perturbation [20]. When using a forward lean-release perturbation, the arms played a significant role in recovering from moderate lean angles, but that role did not persist in assisting recovery from the most difficult angles. Therefore, the stepping threshold measure, a reflection of performance at the most challenging disturbance levels, may not be best suited for revealing the beneficial effects of arm motion. To investigate if the arms were beneficial at moderate levels, we evaluated dynamic stability (MoS\textsubscript{min}) for the successful response to a disturbance two “levels” (-1.0 m/s\textsuperscript{2}) below the largest common disturbance magnitude across conditions. At these moderate levels, arm constraints did not have a large effect on MoS\textsubscript{min} (p= 0.17, Cohen’s d=0.64). It may be that other factors, such as between-study differences in the type of disturbance (i.e.
lean release vs surface translation), led to these inconsistent between-study results. Previous studies have determined that the type and waveform of the perturbation can alter the effects of age [71]. Additionally, surface translations are characterized by a deceleration phase that provides a stabilizing ground reaction force [72], perhaps limiting the need for arm motion to contribute to stabilization. The disturbance-delivery method also alters the initial positions prior to the disturbance. During a forward lean release, the participant does not begin in a neutral, upright position. The initial leaning orientation before a lean-release positions the center of pressure close to the anterior edge of the base of support. This initial position limits modulation of the CoP, likely requiring participants to rely more on the counter-rotation mechanism of stability maintenance.

Recently, using the same participants and protocol of the present study, we validated an “arm-contribution” measure to feet-in-place balance reactions [73]. This measure quantifies the relationship between the change in angular momentum of the arms to that of the body represented as an inverted pendulum. When the arms were not constrained, the arm-contribution measure held acceptable convergent validity, as evident by significant correlations with peak shoulder flexion or extension velocity. In a comparison of similar-magnitude disturbances, constraining the arms reduced this arm-contribution measure for posterior, but not anterior fall-recovery. Our present analysis of how arm-constraints alter stepping thresholds provides more evidence that, in the context of our perturbations, the arms affect posterior, but not anterior balance reactions.

Our method for constraining arms was different from that of previous studies in which participants crossed their arms in front of their chest [20]. Our method of
constraint, with participants holding a rope behind their back, allowed the arms to be in similar starting positions between unconstrained and constrained conditions. This consistency across conditions likely minimized variation in the initial center of mass position of the head-arms-trunk segment. In response to posterior falls, participants may have been able to pull on the rope to assist in hip extension as a counter-rotation strategy. Additional post hoc analyses, however, revealed no between-condition differences in peak hip extension velocities (unconstrained peak hip extension: 48.81±47.06 deg/s, constrained: 44.61+37.50 deg/s, p=0.40, Cohen’s $d = 0.35$).

From this study and our previous validation work on the same cohort [73], it is clear that arm motion can contribute to the feet-in-place response to a posterior fall. It is not yet known, however, if arm constraints also alter the efficacy of the posterior stepping response, as measured by an object, reliable assessment. Given the benefits of arm motion on trunk extension velocity [60], we hypothesize that arm constraints would also reduce posterior, multiple-stepping thresholds [22]. Additionally, we do not know if the contribution of arm motion can be improved with an intervention, especially in at-risk populations such as older adults. Task-specific fall-recovery training, which involves repeatedly practicing fall-recovery after a perturbation, is a promising means to improve fall-recovery and reduce falls in the free-living environment [39], [40]. This approach, modified with specific feedback about arm motion, may lead to a more substantial arm contribution to balance reactions.
Chapter 3

CHRONIC STROKE REDUCES ARM MOTION SYMMETRY, BUT NOT PEAK SHOULDER FLEXION VELOCITIES IN RESPONSE TO POSTERIOR POSTURAL DISTURBANCES

3.1 Introduction

Individuals with chronic stroke have a fall incidence rate as high as three in four people per year [62]–[64]. In this population, approximately 72% of falls are injurious, with more than half of fall-related injuries requiring medical attention [5], [74]. Those with chronic stroke are seven times more likely to experience a fracture, and four times more prone to hip fracture than age-matched groups [75]. This fracture risk increases as the time progresses post stroke [10]. Falls experienced post stroke are associated with readmission to hospitals [76], decreased social activity [77], and limited functional recovery through rehabilitation [38]. Given this evidence, addressing falls is a priority for reducing injury and extending the health span of this at-risk population.

More than one in three falls in those with chronic stroke are caused by external disturbances [5], [8], [9], [53]–[55]. When given such a disturbance in a laboratory setting, individuals with chronic stroke exhibit impaired balance reactions. Specifically, when support-surface translations are delivered to standing participants, those with a past stroke sway more than unimpaired controls [28], [78], [79]. These individuals also take steps in response to smaller forces while standing, and they demonstrate lower multiple-stepping thresholds in all directions compared to that of unimpaired controls [25], [26]. During a feet-in-place recovery response, the paretic limb muscle response is delayed and smaller [27], [29]–[36], [80], [81], and the non-paretic limb muscles compensate with earlier, larger activations [29], [30], [33], [36], [82]–[85].
This response asymmetry is also apparent in the resulting ground reaction forces [78], [79], [82]. When a step is required for fall recovery, those with chronic stroke typically step with their non-paretic limb [83], [84]. Compared to controls, participants with chronic stroke have demonstrated slip-recovery steps that were shorter and closer to the whole-body CoM, resulting in a 71% failure rate compared to 0% for controls [85]. From this evidence, it is apparent that a previous stroke is associated with asymmetrically impaired balance reactions.

Although balance reactions of those with chronic stroke have been characterized by altered lower-extremity muscle activation, smaller stepping thresholds, and less-effective step kinematics, little has been done in evaluating how stroke alters arm motion during fall-recovery. In unimpaired individuals, constraining the arms limits the ability recover balance with a feet-in-place response (See chapter 2) [20]. Older adults, another at-risk population, often demonstrate a protective strategy of shoulder flexion in the response to an anteroposterior fall [24]. This shoulder flexion is opposite to the extension motion of young adults that assists in stabilizing the body. When a step is taken in response to posterior disturbances, younger adults demonstrate faster shoulder flexion than that of older adults [24], [42], in turn reducing trunk extension to a greater extent [19].

Evidence suggests that limited arm function in those post stroke is related to an increased risk of experiencing multiple falls. In this population, repeat fallers have reported reduced arm function, as measured by a clinical test [38]. It is unknown if this relationship between arm function and falls is due to an impaired grasping ability and/or the use of arm motion to arrest the fall. Determining how chronic stroke affects the upper-extremity response to a balance disturbance represents a first step to
understanding the specific means by which upper-extremity function influences fall risk. In turn, we may reveal a specific target for interventions to reduce falls and fall-related injury.

The purpose of this study was to determine the extent to which arm motion during posterior, feet-in-place fall-recovery is reduced for individuals with chronic stroke compared to those with no neuromuscular impairment. We hypothesized that peak shoulder flexion velocities during posterior fall-recovery would be lower in individuals with chronic stroke and that these individuals would display more asymmetry than that of unimpaired individuals. The specific focus on posterior responses is informed by our previous work demonstrating the beneficial role of arm motion during posterior, but not anterior feet-in-place responses of unimpaired adults (See Chapter 2) [73].

3.2 Methods

Data Collection

This study represents secondary analyses of data collected in the context of two other studies. In one of these studies, we aimed to determine if constraining the arms reduced balance reaction capabilities in unimpaired individuals (See Chapter 2). For the present analysis, we included data from the “unconstrained” condition in which arms were allowed to move freely. This condition was first for all participants. The ten unimpaired adults (mean ± s.d. age: 21.7 ± 1.62 years, range 18-23; BMI: 22.5 ± 2.9 kg/m², range 19.17-29.62) had no current or recent self-reported neuromuscular conditions or injuries that would affect balance or extremity motion. Data for ten participants with chronic stroke (age: 64.0 ± 8.8 years, range 53-77; years post stroke: 3.4±1.6 years, range 1-5; BMI: 30.8 ± 4.6 kg/m², range 22.0-37.3) were recorded in
the context of a fall-recovery training study (In Review). For the present analysis, we included data from baseline measurements. The individuals with chronic stroke were recruited from the University of Delaware Stroke Studies Registry. Exclusion criteria included other neurologic disorders, musculoskeletal surgeries within the past year, recent cardiovascular events, or other conditions that precluded safe participation. Participants had a self-reported ability to walk a city block without a gait aid and demonstrated moderate-to-high scores on the Berg Balance Scale (48.9 ± 7.6 out of 56, range 36-56). Those who were 50 years of age or older underwent a Dual-energy X-ray absorptiometry (DXA) screening to ensure that they were not osteoporotic (total hip or femoral neck bone mineral density t-score < -2.5) [86]. This screening criteria was in place to reduce the risk of fractures from the impact of fall recovery steps or falls into the safety harness.

The specific protocol discussed here was identical across studies. All participants were outfitted with a safety harness attached to an overhead rail, adjusted so that the knees and hands could not come into contact with the treadmill (Figure 5). Participants stood on a computer-controlled treadmill (ActiveStep®, Simbex, Lebanon, NH, Figure 1) and were instructed to “try to prevent a step” in response to rapid, 400 ms belt translations [22]. Initial belt accelerations began at 0.5 m/s², lasting a period of 200 ms followed by a 200 ms deceleration phase, in total resulting in a 2 cm total displacement. For subsequent trials, the initial accelerations were increased or decreased by 0.5 m/s² depending on the response success or failure, respectively. A failed attempt was defined as when the participant took a step or when the force transducer (Dillon, Fairmont, MN) between the harness and overhead rail records more than 20% of the participant’s body weight. This criterion was low or comparable
to that of other perturbation studies [69]. The disturbance direction and timing of disturbances were pseudorandomized so that at most, 3 consecutive disturbances were delivered in the same direction, ranging from a 3 to 10 second delay [22]. This variability in direction and timing was intended to discourage a pre-planned response and initial postural orientation that favored certain disturbance directions and timing. Here and going forward, anterior and posterior refer to the direction of the fall, not the direction of the treadmill belt translation (i.e. posterior belt translations elicit an anterior fall). Stepping thresholds, defined as the disturbance magnitude that elicited four failed responses in a given direction [22], were determined in the anterior and posterior directions.

Trials were recorded with a 12-camera motion capture system operating at 120 or 240 Hz (Motion Analysis®, Santa Rosa, CA, replaced mid-study with Qualisys®, Göteborg, Sweden). Thirty-five passive-reflective markers facilitated the definition of 13 body segments: head/neck, trunk, pelvis, upper arms, forearm, thigh, shank, and foot. Marker trajectories were filtered via a fourth-order Butterworth filter with a 6 Hz

Figure 5. (A) Participant with stroke (left) successfully recovers from a posterior postural disturbance (initial acceleration = 2 m/s²), displaying an asymmetrical arm response. (B) Participant with no impairment (right) successfully recovers from a posterior postural disturbance (initial acceleration = 3 m/s²), displaying a symmetrical arm response.
cutoff. In order to characterize arm responses at the most challenging levels for participants, arm kinematics were analyzed at the near-threshold level disturbances (i.e. a successful response to the highest disturbance magnitude). Peak shoulder-flexion velocities of this posterior, feet-in-place recovery responses were determined using custom software (LabVIEW, National Instruments®, Austin, TX).

Given that upper extremity impairment is often asymmetrical in those with chronic stroke [87]–[89], we anticipated that the arm-response would also be asymmetrical. To capture this asymmetry specific to the fall-recovery response, the extremities within each individual were labelled as the faster (i.e. higher peak shoulder velocity) and slower limbs. The symmetry index (S.I.) of these peak velocities was calculated as: $S.I. = \frac{v_{\text{fast arm}} - v_{\text{slow arm}}}{v_{\text{fast arm}}} [90].$

**Data Analysis**

To characterize if performance levels were lower in individuals with stroke than unimpaired individuals, a Mann-Whitney U test was used to compare the disturbance magnitude (i.e. initial acceleration) between GROUPS (stroke vs unimpaired). To determine if peak shoulder flexion velocities during posterior fall-recovery were lower in individuals with chronic stroke than that of unimpaired individuals, a mixed factorial ANOVA was used to see the effects of GROUP (stroke vs unimpaired) and LIMB (fast vs slow), where post-hoc evaluations included a Sidak correction for multiple comparisons. Lastly, in order to determine if the individuals with stroke demonstrated more asymmetry than that of unimpaired individuals, the symmetry index was compared between GROUPS (stroke vs unimpaired) using an independent t-test. Between-group effects were also expressed as Glass’s $\Delta$, an effect size in which mean differences are expressed relative to the standard deviation of the
unimpaired group. All statistical analyses were conducted using SPSS (SPSS v24, IBM, Armonk, NY). Preliminary results from six unimpaired participants and six participants with chronic stroke suggested a medium-to-large effect of the GROUP x LIMB interaction ($f = 0.30$). Given a conservatively estimated correlation between within-subject measures (LIMB, $r = 0.60$ compared to the observed $r = 0.84$ from our preliminary data), ten participants per group were needed to have adequate power ($1-\beta = 0.80$) to detect this interaction as significant.

3.3 Results

Posterior Thresholds

As expected, posterior thresholds were found to be significantly higher for the unimpaired group ($3.15 \pm 0.58 \text{ m/s}^2$) compared to the stroke group $1.8 \pm 0.48 \text{ m/s}^2$, $p<0.001$, Mann-Whitney $U = 4.5$). Thus, the disturbances analyzed in this study were higher for the unimpaired group compared to that of the stroke group.

Shoulder Flexion Velocities
As anticipated, an ANOVA analysis revealed a significant ARM*GROUP interaction on peak shoulder flexion velocity (p=0.017), with trends suggesting a greater asymmetry in the stroke group. Post hoc evaluations determined that "fast" limbs were indeed faster than the “slow” limbs in both the stroke (p<0.001, Cohen's d = 2.53) and non-stroke (p = 0.04, Cohen's d = 0.92) groups. However, there were not meaningfully significant or large differences between the unimpaired and impaired groups in the “fast” arm peak shoulder velocity (p = 0.82, Glass's Δ= -0.10) or “slow” arm peak shoulder velocity (p = 0.20, Glass's Δ= 0.48). From the symmetry-index measure, it was confirmed with a Mann-Whitney U test that those with chronic stroke did have more asymmetry in their arm response (Figure 6C, p=0.005, Glass's Δ = 1.34).

Figure 6 (A) Posterior single-stepping thresholds in unimpaired and impaired participants. (B) Peak shoulder flexion velocities in both Unimpaired and Impaired groups. Arms were categorized as either the faster or slower arm during the response. "Fast" limbs were faster than the “slow” limbs in both the stroke (p<0.001, Cohen's d = 2.53) and non-stroke (p = 0.04, Cohen's d = 0.92) groups. No significant or large between-group differences were observed in the “fast” arm peak shoulder velocity (p = 0.82, Glass's Δ= -0.10) or “slow” arm peak shoulder velocity (p = 0.20, Glass's Δ= 0.48). (C) Symmetry Index of the peak shoulder flexion velocities of Unimpaired and Impaired groups in response to the highest disturbance magnitude.
3.4 Discussion

The purpose of this study was to determine the extent to which arm motion during posterior fall-recovery is reduced for individuals with chronic stroke compared to those with no neuromuscular impairment. We hypothesized that peak shoulder flexion velocities during posterior fall-recovery would be lower in individuals with chronic stroke. We also hypothesized that individuals with stroke would demonstrate an asymmetrical upper extremity response, similar to what previous observation had found[87]–[89]. Although individuals with stroke confirmed more asymmetry than our unimpaired participants, we did not detect large between-group differences in that motion.

From analyses of peak shoulder flexion velocity alone, it does not appear that upper extremity motion in response to a posterior fall is meaningfully altered for those with chronic stroke. Instead, deficiencies in the posterior fall-recovery response could be primarily due to alterations in lower-extremity muscle activation, trunk rotation, and step placement [85],[91]. Perhaps, however, shoulder flexion velocity alone does not best characterize how stroke affects the use of the upper extremities during this task. Recently (i.e. after the current study was proposed), and using the same unimpaired control participants as this study, we validated an “arm-contribution” measure of feet-in-place balance reactions [73]. Briefly, this contribution measure quantifies the relationship between the change in angular momentum of the arms to the change in angular momentum of the body represented as an inverted pendulum. So, if the change in arm momentum closely corresponds with that of the inverted pendulum, a higher “arm-contribution” will result. Unlike peak velocity measures, this contribution measure is influenced by the relative timing of the arm motion, as well as the contribution of other stabilizing strategies. Of note, when we instead used this
measure to compare the net-effect of arm motion on stabilization (Figure 7), we detected a remarkably large difference between groups (p=0.002, Glass’s Δ=1.10). Given the results of the asymmetry index (Figure 6C) and this more sophisticated biomechanical measure, we believe that there are indeed group-differences in how individuals used arm motion to restore stability. We do not know, however, if this is due to poor execution of the arm strategy or more reliance on other mechanisms (e.g. moving the center of pressure, using a “hip strategy”) to restore stability.

Similar to previous studies of unimpaired participants[20], there was notable between-participant variability in arm motion for the unimpaired individuals. This between-participant variability in arm motion was more profound than that observed in the stroke group (Figure 6B). Five of ten unimpaired participants displayed peak shoulder velocities less than 90 degrees/s, which corresponded with “arm contributions” of 5% or less. This result, as well as the observed asymmetry in some unimpaired adults (Figure 6C), suggests that unimpaired individuals do not inherently use their arms to retain stability during a fall. Such between-participant variability in
the use of the arms could underlie our lack of between-group differences in shoulder motion.

A limitation of our study is that we do not know if the between-group effects were due to stroke impairment or the influence of age, as the two groups had significantly different ages (p < 0.001, Cohen’s d = 6.66). Previous studies investigating the influence of old age on the slipping response found a moderate effect size of age on peak shoulder flexion angles (p<0.1, Cohen’s d= 0.47) when comparing younger and older groups [42]. Our observed, between-group effect on the contribution of arm motion was much larger (p=0.002, Glass’s Δ=1.10). This between-study discrepancy in effect sizes could reflect the much larger effect of chronic stroke and age compared to that of age alone. In a subsequent ANCOVA analyses accounting for age as a covariate, the effects of group on arm contribution were no longer significant (p=0.210), although the effects of group on asymmetry persisted (p=0.048). In order to remove the confounding variable of age on our variables, future studies should match participants by age.

Given that arm contributions were different between groups, the arm response may serve as a target for interventions to improve balance reactions. Previously, the reactive arm response of grasping a rail was improved with practice in older adults [92]. We do not know if the arm motions studied here are modifiable with practice, especially for those with chronic stroke. Balance reactions in general are modifiable in this population, as evident by those who were able to recover from larger surface perturbations with a feet-in-place response, improving weight-bearing symmetry [80]. Even between first and second exposures of simulated slips, individuals with chronic stroke have been able to modify the stepping response of their paretic limb [93]. Given
this evidence of improvement with task-specific practice, we hypothesize that the arm contributions of those with chronic stroke could also be improved. We do not know, however, if such improvement would be through compensation of the less-affected arm or through improvements in the response of both limbs. In general, intensive exercise can improve arm function in this population, although the optimal forms of exercise have not been determined [94].
Chapter 4

**CONCLUSION**

*Study 1: Constraining the arms reduces posterior, but not anterior single-stepping thresholds in young, unimpaired adults*

The purpose of this study was to determine the effects of arm-motion constraints on posterior single-stepping thresholds in young, non-impaired adults. We hypothesized that arm-motion constraints would reduce such thresholds and the resulting dynamic stability for the arms constrained condition. Constraining the arms impaired the posterior fall-recovery response based on significant reductions in posterior single-stepping thresholds. Our interpretation of our dynamic stability results is that, when participants had their arms constrained, smaller disturbance magnitudes resulted in a level of instability characteristic of near-threshold responses. Unlike the effects on posterior single-stepping thresholds, anterior thresholds were not altered by arm constraints. The method of perturbation may play a role in explaining the differences between our results and that of previous studies.

By providing evidence that the arms contribute to the posterior fall-recovery response, we have identified a potential target for interventions to improve fall recovery. Previous work has demonstrated that the arm response during fall recovery is modifiable in older adults, as seen by an improved ability in the grasp response after training[92]. So, it may also be feasible to improve the arm reaction that stabilizes the body through the counter-rotation mechanism.
**Study 2: Chronic stroke reduces arm motion symmetry, but not peak shoulder flexion velocities, in response to posterior postural disturbances**

The purpose of this study was to determine the extent to which arm motion during posterior, feet-in-place fall-recovery is reduced for individuals with chronic stroke compared to those with no neuromuscular impairment. We hypothesized that peak shoulder flexion velocities would be lower and more asymmetrical in individuals with chronic stroke. Stroke survivors demonstrated more asymmetry than unimpaired participants, although we did not detect large between-group differences in peak shoulder flexion velocity. When considering a recently-validated arm contribution measure, we detected large between-group differences. We do not know if these between-group effects were due to stroke impairment or the influence of age. Thus, we can hypothesize, but not confirm that upper extremity function was the underlying source of the asymmetry observed in the stroke group. Future studies are needed to address this hypothesis, as well as to determine if this compensatory arm response is modifiable with intervention.
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Appendix

APPENDIX

INSTITUTIONAL REVIEW BOARD APPROVAL LETTER

DATE: February 7, 2018

TO: Jeremy Chernshaw, PhD
FROM: University of Delaware IRB

STUDY TITLE: [635464-3] A study of how counter-rotation movements help maintain balance
SUBMISSION TYPE: Continuing Review/Progress Report
ACTION: APPROVED
APPROVAL DATE: February 7, 2018
EXPIRATION DATE: February 16, 2019
REVIEW TYPE: Expedited Review
REVIEW CATEGORY: Expedited review category # (4)

Thank you for your submission of Continuing Review/Progress Report materials for this research study. The University of Delaware IRB has APPROVED your submission. This approval is based on an appropriate risk/benefit ratio and a study design wherein the risks have been minimized. All research must be conducted in accordance with this approved submission.

This submission has received Expedited Review based on the applicable federal regulation.

Please remember that informed consent is a process beginning with a description of the study and insurance of participant understanding followed by a signed consent form. Informed consent must continue throughout the study via a dialogue between the researcher and research participant. Federal regulations require each participant receive a copy of the signed consent document.

Please note that any revision to previously approved materials must be approved by this office prior to initiation. Please use the appropriate revision forms for this procedure.

All SERIOUS and UNEXPECTED adverse events must be reported to this office. Please use the appropriate adverse event forms for this procedure. All sponsor reporting requirements should also be followed.

Please report all NON-COMPLIANCE issues or COMPLAINTS regarding this study to this office.

Please note that all research records must be retained for a minimum of three years.

Based on the risks, this project requires Continuing Review by this office on an annual basis. Please use the appropriate renewal forms for this procedure.
If you have any questions, please contact Nicole Farnese-McFarlane at (302) 831-1119 or nicolefm@udel.edu. Please include your study title and reference number in all correspondence with this office.