FALL-RECOVERY TRAINING APPLIED TO INDIVIDUALS WITH CHRONIC STROKE

by

Loren J Pigman II

A dissertation submitted to the Faculty of the University of Delaware in partial fulfillment of the requirements for the degree of Doctor of Philosophy in Biomechanics and Movement Science

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ABSTRACT

Individuals with chronic stroke have a fall risk that is twice that of age- and sex-matched peers. A third of falls in this population are from a trip or a slip. An impaired ability to recover from anterior and posterior falls likely contributes to this high risk of trip- and slip-related falls. Although neural factors underlie the impaired fall-recovery response of those with stroke, the subsequent effects of lower-extremity impairment on fall-recovery kinematics are not well understood. Such kinematics are important, as they are determinants of fall-recovery success. By addressing this unknown, we hope to identify specific targets for interventions to improve recovery from common fall causes.

Traditional exercise has not reduced falls in those with chronic stroke. Perturbation-based balance training, a specific form of exercise that targets fall-recovery, has reduced falls in other at-risk populations. The extent to which, and specific means by which fall-recovery training benefits those with chronic stroke, however, is not known. If the fall-recovery response of this population is improved with practice, then the efficacy of exercise in reducing falls may be improved.

The purpose of this study was to determine the effects of lower limb, stroke-related impairment on anterior and posterior fall-recovery performance, and then determine the benefits of exercise focused specifically on improving fall-recovery skill. We hypothesized that compensatory steps with the paretic limb would be associated with worse fall-recovery performance and kinematics. We also hypothesized that such aspects would improve with specific fall-recovery training.

Performance was quantified as the proportion of successful recoveries within a series of perturbations, as well as the highest perturbation magnitude within that series. "Worse" kinematic features included shorter and wider recovery steps relative to the CoM, as well as larger peak trunk rotation angles and angular velocities.

Fifteen relatively high-functioning individuals (29-77 years old, 2-15 years post stroke, 76-100 Activities Specific Balance Confidence Scale, 36-56 Berg Balance Scale) with chronic stroke performed up to six sessions of fall-recovery training. Each session consisted of four progressions of treadmill-induced perturbations. The four progressions focused on responding to anterior or posterior falls with the initial step being the paretic or non-paretic limb. Fall-recovery performance as well as step and trunk kinematics were compared between the initial stepping limbs in the first session. Limb-specific outcomes were also compared between the first and last training sessions.

There were no between-limb differences in *anterior* fall-recovery performance in the first session, however, there were between-limb differences in compensatory step placement. In response to an anterior fall, steps with the paretic limb were wider and shorter relative to the center of mass (p's < .056, d's > 0.61). With training, participants successfully recovered from a higher proportion of anterior falls (p's = .011, Cohen's d's > 0.73), progressed to larger perturbation magnitudes (p's < .065, d's > 0.54), initial paretic limb steps became longer (p = .034, d = 0.66), and trunk forward rotation was reduced when first stepping with the non-paretic limb (p = .029, d = 0.62).

There were between-limb difference in *posterior* fall-recovery performance in the first session. Initial posterior steps with the non-paretic limb were associated with a higher proportion of success than initial steps with the paretic limb (p = .015, Cohen's d = 1.26). In response to a posterior fall, steps with the paretic limb were wider relative to the center of mass (p = .011, d = 1.50). With training, participants successfully recovered from a higher proportion posterior falls (p's < .049, d's > 0.54), as well as progressed to larger perturbation magnitudes (p's < .042, d's > 0.61). There were no significant changes in kinematic variables with posterior fall-recovery training (p's > .071, d's < 0.51).

The initial stepping limb affects relevant step kinematics during anterior and posterior fall recovery of high-functioning individuals with chronic stroke. We demonstrated that, because we saw performance-based and kinematic adaptations to the fall-recovery response, our fall-recovery training is a potentially beneficial exercise intervention for those with chronic stroke. Anterior fall-recovery training improved performance and select kinematic outcomes. Although this study provides evidence that the skill of posterior stepping in response to a fall can be improved with practice in those with chronic stroke, we were not able to identify the underlying kinematic mechanisms behind this adaptation. Further study is required to determine if this form of training is effective at reducing trip- and slip-related falls in the free-living environment in survivors of stroke.

Chapter 1

INTRODUCTION

1.1 Overview

Chronic stroke is the leading cause of long-term disability in the United States [1, 2]. As of 2012 there are an estimated 6.6 million Americans $\geq$ 20 years of age that have had a stroke [3]. Projections suggest a 20.5% increase in stroke prevalence by 2030 [3]. On average, someone in the United States has a stroke every 40 seconds, with approximately 610,000 new individuals experiencing a stroke every year [4]. Improvements in post-stroke treatment have decreased stroke mortality [5], from this we can infer that there are more individuals living with chronic stroke for a longer period of time.

Individuals living with chronic stroke have an increased risk and rate of falling. This population is at an increased risk of falling during the acute, sub-acute, and chronic stages of stroke recovery [6]. Previous research suggests that 57% of community-dwelling individuals with stroke fall two or more times over a twelve month time period [7–11]. During hospitalization care, up to 65% of individuals experience at least one fall [9, 12–14]. Furthermore, as many as 73% of survivors of stroke experience a fall during their first six months after release from impatient care [7, 9, 15–17]. Those with a stroke are almost twice as likely to experience a fall at home (59%) compared to matched unimpaired peers (32%) [18]. The greater percentage of falls occurring in the home can be attributed to individuals spending more time in their home, being more cautious or reluctant to leave the house. The high risk of falling persists in the later stages of chronic stroke. The self-reported, annual fall incidence of individuals living with chronic stroke is 36%, compared to 24% in age- and sex-matched peers [19]. This risk persists after many years post stroke. Individuals that were, on average, ten years after their stroke had a risk of falling that was more than two times higher than that of age- and sex-matched peers (adjusted relative risk 2.2, 95% CI: 1.1 – 4.3), and their risk of frequent falls was also higher (adjusted relative risk 3.4, 95% CI: 1.0 – 11.7) [20].

1.2 Significance and Innovation

Falling represents a major threat of serious injury for those living with chronic stroke. Of the falls experienced by survivors of stroke, 72% resulted in injury, with 55% of the injuries requiring medical attention, including 43% of the injuries resulting in an emergency room visit or impatient hospitalization [21]. Individuals with chronic stroke have a risk of fracture that is seven times that of those who have not had a stroke [22]. Approximately 84% of fractures in this population are due to an accidental fall [23]. About 45% of these fracture are of the hip, and most post-stroke bone fractures occur on the paretic side [23]. Those who have had a stroke have an incidence of hip fracture that is four times higher than that of an aged-matched reference population, with the risk ratio being higher for the younger age group [23]. It is estimated that, at one-year post stroke, the risk of sustaining a fracture is 4%. After five years, the risk increases to 15%, and after ten years, it increases to 24%. Those with stroke can develop disuse osteoporosis in the paretic limb, increasing the risk of bone fracture [24, 25]. In addition, the five-year mortality for those who have had a stroke and experienced a hip fracture (80%) is significantly higher than that of hip fracture patients with no history of stroke (60%) [26], with significant between-group differences also observed at one month (9% vs 3%) and one year (29% vs 17%) post fracture.

Experiencing a fall reduces independence, and the overall quality of life in survivors of stroke. Post-stroke falls can lead to increased care-giver stress [17], decreased social activity [17], and readmission to the hospital [27]. Individuals with chronic stroke have an increased fear of falling and decreased fall's self-efficacy. Up to 80% of people with chronic stroke express a fear of falling [28]. Falls self-efficacy is a determinant of physical activity, engagement, and participation in the free-living environment after stroke [13, 29–32]. Activity level is an independent predictor of life satisfaction [33] and quality [34], and it has a substantial impact on post-stroke health [4, 35]. About 45% of those with mild stroke impairment report problems with community mobility, with a 20% to 42% loss of pre-stroke regular activities [34, 36]. Those with stroke likely limit their activity as a protective means to prevent falls and injury. Therefore, an intervention that reduces the risk of falling is likely an important component to improving the independence, quality of life, and the health of those living with chronic stroke.

Individuals living with chronic stroke have an impaired fall-recovery response. When compared to unimpaired controls, survivors of stroke demonstrate greater sway after support-surface translations [37–39], as well as a lower static force threshold that can be resisted while standing [40]. In response to a static force applied at the waist, those with chronic stroke take a recovery step in response to smaller forces compared to unimpaired controls [40]. Compared to that of unimpaired controls, individuals with chronic stroke demonstrate lower multiple-stepping thresholds in the anteroposterior and lateral directions [41]. Here multiple-stepping thresholds are the disturbance magnitudes that elicit more than one step. These lower thresholds are associated with a delayed and reduced muscle response of the paretic limb. When compared to unimpaired controls, individuals with chronic stroke have a reduced ability to arrest trunk rotation and peak trunk rotation velocity in response to treadmill-induced trips and slips [42]. When posterior falls are induced, individuals with chronic stroke exhibited recovery steps that were shorter and closer to the whole-body CoM, resulting in a 71% failure compared to 0% for the controls [43]. These aspects are relevant, as trunk rotation and step kinematics are determinants of fall-recovery success after simulated trips [42, 44, 45] and slips [42, 43].

Paretic limb dysfunction directly impairs the fall-recovery response. The impaired fall-recovery response is most apparent when the fall is toward the paretic limb [37, 39–41, 46, 47]. During a feet-in-place response to a postural disturbance, the paretic limb muscles produce a diminished and delayed response characterized by co-contraction [37, 48, 49]. In response to posterior falls, those who fell had delayed muscle activation of the paretic limb [50]. These deficiencies were associated with more trunk rotation, in turn placing the trunk further outside of the base of support. Individuals with chronic stroke most often prefer to take an initial recovery step with their non-paretic limb [45, 51, 52], and they sometimes perform a "hopping" strategy on their non-paretic limb in order to delay stepping with their paretic limb. In a previous study, the inability to take fall-recovery steps with the paretic limb was prospectively related to falls in the free-living environment [53].

Traditional post-stroke exercise interventions have not successfully reduced falls for individuals with chronic stroke. A meta-analysis of exercise interventions for those with chronic stroke found no significant effect on the rate of falls (rate ratio

0.75, 95% CI 0.41 – 1.38, 412 participants) or the risk of falling (risk ratio 1.02, 95% CI 0.83 – 1.24, 616 participants) [54]. This is further supported with a similar review in which exercise interventions were unable to reduce fall rates and fall risk (rate ratio 1.22, 95% CI 0.76 – 1.98; risk ratio 0.95, 95% CI 0.73 – 1.22) [55]. The interventions in these reviews included both supervised and unsupervised strength, gait, balance, flexibility, and endurance training. Although these types of exercises have been shown to improve general measures of gait and balance [56], they have failed to reduce falls in this at risk population. A limitation of these commonly used post-stroke exercise interventions is the lack of processing specificity [57]. In other words, the motor responses needed to prevent a fall are not elicited or effectively trained using traditional exercise modalities. One-third of falls by individuals with chronic stroke are caused by a trip or a slip [21]. Therefore, we suggest that by *specifically* practicing the rapid, coordinated skill of trip- and slip-recovery, the ability to prevent falls from common causes can be improved in those with chronic stroke.

Perturbation-based training has reduced falls in other populations. Based on a review of 404 older adults and individuals with Parkinson's diseases, external and internal perturbation-based training reduced the risk of falling by 30% (risk ratio 0.71, 95% CI 0.52 – 0.96) [58]. Perturbation-based training has also reduced falls in individuals in the sub-acute stages of stroke recovery [59]. External perturbations were delivered using a waist push or pull by a physical therapist. Internal perturbations consisted of rapid voluntary movements such as kicking a ball. The training was made more challenging as the participants improved. After training was performed participants reported falls for 6 months. The number of reported falls were compared to reported historical falls from individuals with sub-acute stroke that did not perform perturbation-based training. The group that performed the perturbation-based training reported 10 falls (0.84 falls per person per year) compared to the non-trained control group that reported 31 falls (2.0 falls per person per year). When these results were controlled for follow-up duration and motor function, fall rates were significantly lower in the trained group compared to the untrained group (risk ratio 0.36, 95% CI 0.15 – 0.79). Although, the application and effectiveness of perturbation-based training with survivors of chronic stroke is not well understood.

The fall-recovery response of individuals with chronic stroke appears to be modifiable. Those with stroke improved the feet-in-place response to platform surface translations with practice [60, 61]. Improvements included an increase in the maximum movement amplitude that was successfully recovered from without the use of a hand support [60, 61], and improved weight-bearing symmetry while standing [60]. Within a single session those with chronic stroke were able to modify their recovery response to a slip on the second attempt [62]. Improvements included a more stable position of the whole-body CoM at the point of toe-off of the recovery step and longer recovery steps. Those with chronic stroke were able to improve anterior fall-recovery after completing a ten day intensive physical therapy program focused on functional mobility, upright static and dynamic balance, and underlying impairments in strength and range of motion [63]. Adaptations from this training included, a reduction in the time needed to regain stability from a feet-in-place response to a platform surface translation, as well as improving weight-bearing symmetry. In a recent report, a six-week perturbation-based intervention for those with chronic stroke resulted in higher scores on the reactive balance subset of the mini-BESTest compared to a control group receiving traditional therapy [64]. Although, as seen in these examples, the fall-recovery response appears to be modifiable with practice for those with chronic stroke, we do not know how the initial stepping limb affects that adaptation. To address this gap in knowledge, we will determine how lower-limb impairment alters the stepping response of individuals with chronic stroke. We will then evaluate the extent to which, and specific means by which, fall-recovery can be improved using task-specific training.

To our knowledge, our study is the first to train those with chronic stroke using treadmill-induced simulated trips and slips. With a six-week perturbation-based intervention, those with chronic stroke demonstrated higher scores on the reactive balance subset of the mini-BESTest compared to a control group receiving traditional therapy [64]. This evidence suggests that such training can improve the stepping response. However, the benefit of perturbation-based training on subsequent falls was inconclusive, with no between-group differences in post-training fall rates. Given that perturbation-based balance training has reduced falls in older adults, individuals with Parkinson’s disease, and those with sub-acute stroke [58, 59], the benefits of such training on those with chronic stroke warrant investigation beyond a single study. In the aforementioned randomized controlled trial [64], balance training consisted of self-initiated tasks and therapist-delivered pulls and pushes, a feasible approach in many settings. Along with its magnitude, the method of applying a perturbation (e.g. surface translations, waist pulls, lean releases) alters the response to the perturbation and influences the degree to which responses reflect balance impairment [65]. It is reasonable to explore whether other methods of delivering perturbations, especially those that allow for large-magnitude disturbances, could elicit greater benefits in terms of reducing the risk of falls.

Our training differs from previous exercise interventions in that we *specifically* practiced the rapid, coordinated skill of trip- and slip-recovery using a computer-controlled treadmill. Computerized treadmills have been used to implement controlled, repeatable, and challenging perturbations [66–70]. Rapid treadmill belt accelerations, directed posteriorly, require a similar recovery response to that of trip-recovery [44, 71], while rapid, anteriorly-directed surface translations, require a similar recovery response to that of slip-recovery [72, 73]. Unlike training using therapist-induced perturbation [59, 64], our treadmill perturbations are more controlled, precise, and repeatable. This allows for the reapplication and analysis within and across training sessions. It has been shown that the method and waveform of the perturbation influences the destabilizing effect of the perturbation, with surface translations more effectively revealing age-related balance deficiencies compared to waist pulls [65]. In addition, allowing multiple steps in response to a forward fall aligns with the multistep responses of trip-recovery [74]. The ability to use a treadmill that closely emulates actual trip- and slip-induced falls in a safe environment may serve as a means to reduce falls in those with chronic stroke using exercise.

Our study is **significant**, because it addresses the high risk of falling in those with chronic stroke. In doing so, we are also addressing a notable source of injury and a barrier to physical activity. Our study is **innovative** because it represents a departure from the current, ineffective approach to fall prevention by focusing on the specific motor processes needed to successfully arrest a fall from a trip or slip. As a first step to this departure, we will evaluate how stroke-related lower-limb impairment influences the skill of fall-recovery and its adaptation with task-specific exercise. In doing so, we are evaluating and addressing biomechanical mechanisms that underlie the impaired fall-recovery response in this population.

1.3 Purpose

The purpose of this study is to determine the effects of lower limb, stroke-related impairment on anterior and posterior fall-recovery performance, and then determine the benefits of exercise focused specifically on improving fall-recovery skill.

1.4 Hypothesis

We hypothesize that compensatory steps with the paretic limb would be associated with worse fall-recovery performance and kinematics. We also hypothesize that such aspects will improve with specific fall-recovery training. We will quantify performance as the proportion of successful recoveries within a series of perturbations, as well as the highest perturbation magnitude within that series. “Worse” kinematic features include shorter and wider recovery steps relative to the CoM, as well as larger peak trunk rotation angles and angular velocities.

Chapter 2

ANTERIOR FALL-RECOVERY TRAINING APPLIED TO INDIVIDUALS WITH CHRONIC STROKE

2.1 Introduction

Up to 75% of those living with stroke fall each year [6, 17, 75], and individuals with chronic stroke have a fall risk that is twice that of age and sex matched peers [20]. Of the falls experienced by survivors of stroke, 72% result in injury, with 55% of injuries requiring medical attention. Approximately 84% of fractures in this population are due to accidental falls [23]. Up to 80% of chronic stroke survivors express a fear of falling [28]. Having reduced falls self-efficacy limits physical activity, engagement, and participation in the free-living environment after stroke [13, 29–32]. Therefore, an intervention that reduces the risk of falling is likely an important component to preventing injury and improving the independence, quality of life, and the health of those living with chronic stroke.

Survivors of stroke have an impaired fall-recovery response. Compared to unimpaired controls, they demonstrate more sway after support-surface translations [37–39]. Those with chronic stroke take a recovery step in response to smaller forces applied at the waist [40]. Similarly, they demonstrate lower anterior multiple-stepping thresholds, defined as the perturbation magnitude that elicits more than one recovery step [41]. This lower threshold is associated with a delayed and reduced muscle response of the paretic limb [50]. In response to larger treadmill-induced falls, individuals with chronic stroke have a reduced ability to limit trunk rotation [42]. When stepping to avoid a fall, those with chronic stroke prefer to step with their non-paretic limb [45, 51, 52] and avoid bearing weight on the paretic limb [45]. The inability to take a recovery step with the paretic limb has been prospectively related to falls in the free-living environment [53]. Despite this distinct and relevant influence of limb function asymmetry on fall-recovery after stroke, the effects of the stepping limb (i.e. paretic or non-paretic) on subsequent fall-recovery kinematics (e.g. trunk rotation, foot placement relative to the whole-body center of mass (CoM)) are not known. Such kinematics are important, as they are determinants of fall-recovery success [44, 45, 71, 76].

Trips and slips cause one-third of post-stroke falls [21]. Successful recovery from a trip-induced fall is often dependent upon step placement and the ability to limit trunk forward rotation [44, 45, 76]. Exercise interventions aimed at reducing falls in this population have focused on strength, gait, standing balance, flexibility, and endurance but do not seem to have an effect on rate of falls or risk of falling [54, 55]. A limitation of these previous interventions is the lack of *processing specificity*, whereby the motor responses needed to prevent a fall are not elicited or effectively improved using traditional exercise modalities [57]. The extent to which the compensatory stepping response of those with chronic stroke can be improved with such repetitive practice is not well understood. With a six-week intervention that included external perturbations (e.g. manual pushes from a therapist), those with chronic stroke demonstrated higher scores on the reactive balance subset of the mini-BESTest compared to a control group receiving traditional therapy [64]. This evidence suggests that such training can improve the stepping response, although a pre-test/post-test design is necessary to confirm this hypothesis.

The purpose of this study was to assess the effects of the initial stepping limb (i.e. paretic or non-paretic) on anterior fall-recovery performance and kinematics, as well as to determine the benefits of specific fall-recovery training on those outcomes for individuals with chronic stroke. We hypothesized that compensatory steps with the paretic limb would be associated with worse fall-recovery performance and kinematics. We also hypothesized that such aspects would improve with specific fall-recovery training. Performance was quantified as the proportion of successful recoveries within a series of perturbations, as well as the highest perturbation magnitude within that series. “Worse” kinematic features included shorter and wider recovery steps relative to the CoM, as well as larger peak trunk forward rotation angles and angular velocities [44, 45, 71, 76]. We assessed the first and second steps of each response.

In order to explore the neuromuscular mechanisms that may underlie observed between-limb or between-session differences in anterior fall-recovery, we evaluated EMG of the non-stepping, stance limb plantar and dorsiflexors when feasible to do so. We expected the paretic limb to be characterized by a delayed response with less plantarflexor activity and more co-contraction. We also expected that training would result in less delay, more plantarflexor activity, and less co-contraction.

2.2 Methods

2.2.1 Participants

We recruited eighteen participants with a unilateral chronic stroke from the University of Delaware’s Stroke Studies Registry. This study was approved by the University of Delaware’s Institutional Review Board, and all participants provided written informed consent prior to participation. Additionally, participants and research staff that appear in photographs gave written consent. Fifteen (out of 18) of these participants who completed at least five of the six training sessions were included in this analysis (12 males, 3 females) (Table 2.1). Exclusion criteria included other neurologic disorders, musculoskeletal surgeries within the past year, recent cardiovascular events (past three months), or other conditions that precluded safe participation. Participants had the self-reported ability to walk a city block without a gait aid such as a walker or cane. 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) [77]. This screening criterion, which has been used previously in studies of older adults [78], was conservatively in place to reduce the risk of fractures from the impact of fall-recovery steps or falls into the safety harness. No individuals were excluded from the study due to DXA screening. Of note, two participants wore articulating ankle foot orthosis during training that they typically wore on a day-to-day basis. We anticipated that removing the orthosis for training may have presented an unreasonable injury risk.

Table 2.1. Demographic and clinical assessment data.

Measure	**Mean (SD), Range**
Age (Years)	57 (12), 29 – 77
BMI (kg/m^2)	28.6 (3.6), 22.0 – 33.9
Years after stroke	5 (3.5), 2 – 15
Fugl-Meyer LE	24 (6), 8 – 32
Activities Specific Balance Confidence Scale (ABC)	91 (8), 76 – 100
Functional Gait Assessment (FGA)	17 (6), 9 – 29
Berg Balance Scale (BBS)	50 (7), 36 – 56

Note: Prior to starting fall-recovery training, descriptive measures of the Fugl-Meyer Lower Extremity assessment [115], Activities-Specific Balance Confidence (ABC) scale [116], Berg-Balance Scale [117], and the Functional Gait Assessment [118] were used to characterize our participants balance and mobility.

2.2.2 Training Protocol

Our training protocol was modified from previous interventions aimed at benefitting older adult women [66, 67] and individuals with lower-extremity amputations [68, 69]. The perturbations delivered within our training were designed to elicit the rapid, coordinated stepping response similar to that of trip recovery [71]. All participants attempted to complete six sessions of the training protocol. The sessions included two progressions of treadmill belt perturbations that induced anterior falls (ActiveStep®, Simbex, Lebanon, NH). Progressions within a training session focused on initial steps with the non-paretic limb (Figure 2.1) or paretic limb. These progressions were limited to either 15 minutes or 36 perturbations, whichever occurred first, with rest periods lasting approximately five minutes between each progression. In addition, two progressions that focused on posterior fall recovery were delivered within each session. The results from these latter progressions are reported in Chapter 3 of this dissertation. Progression durations were determined to reasonably limit fatigue and to keep training sessions within an hour. Six training sessions occurred over approximately three weeks.

Participants wore well-cushioned, closed-toe shoes with no elevated heels. They were outfitted with a full-body safety harness (Delta™, Capital Safety, Bloomington, MN) attached to a custom-built overhead rail system. The support straps were adjusted so that the participant's hands and knees could not come into contact with the treadmill. The harness was instrumented with a force transducer (Dillon, Fairmont, MN), the peak forces of which were recorded for each trial.

When awaiting a perturbation, participants stood self-supported on the treadmill without the use of handrails (Figure 2.1). They placed their feet at a comfortable width, with their toes evenly positioned in the anteroposterior direction.

The perturbation velocity waveforms consisted of an initial, 500 ms acceleration followed by a deceleration phase at 0.38 m/s^2. Participants were instructed to "try not to fall" in response to these perturbations, and to specifically step with the targeted limb. The first perturbation of each progression had an initial acceleration of 0.5 m/s^2, resulting in a displacement of 0.06 m. After a successful recovery, the subsequent perturbation had an initial acceleration 0.25 m/s^2 greater than the previous perturbation [69]. After a failed recovery, the subsequent trial acceleration was reduced by 0.25 m/s^2. Failures were defined as responses in which the force transducer attached to the harness recorded more than 20% body weight [79], as well as responses in which the participant stepped with the wrong limb. Each perturbation was preceded by a 1 - 5 s delay to limit pre-planned timing of the response. Additionally, small perturbations (0.3 ms duration, 0.03 m displacement) resulting in a posterior fall were introduced approximately once every six trials to limit anticipatory adjustments. Participants were asked to inform research staff if the training intensity became too much for them to tolerate (i.e. minor muscle soreness, general fatigue, or uneasiness being on the treadmill). In such cases, we attempted to continue training at the highest perturbation magnitude tolerated for the remainder of the session. This approach was intended to maintain compliance with study participation while promoting practice repetitions.

All trials were recorded with a 12 camera motion capture system operating at 120 Hz (Motion Analysis®, Santa Rosa, CA, replaced mid-study with Qualisys®, Göteborg, Sweden). The positions of 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.

Muscle activity was recorded using surface electromyography (Delsys, Natick, MA, 1200 Hz) of the bilateral ankle dorsi- and plantar-flexors [50, 80], where four wireless bipolar surface electrodes were placed bilaterally on the medial gastrocnemius (MG) and tibialis anterior (TA). Sensors were oriented in the direction of the muscles' fibers and at half the distance between the motor endpoint and the distal end of each muscle [81]. Unfiltered EMG signals were shifted to account for a 48 ms delay between EMG sensors and kinematic data (as per Delsys equipment documentation). The signals were then de-meaned, bandpass filtered (10 - 300 Hz), rectified, and lowpass filtered (8th order Butterworth) at 50 Hz for muscle onset latency and 4 Hz for peak activation and co-contraction ratios. A higher frequency cutoff was used for muscle onset latency due to the effect of filtering on time-based measures of muscle activity [82].

2.2.3 Analysis

Fall-recovery performance was quantified from the proportion of successful recoveries and the highest perturbation magnitude achieved within a session. To determine how lower-extremity impairment affected performance at baseline, we compared the separate progressions of stepping with the paretic or non-paretic limb within the first training session. To evaluate if these measures changed with training, we compared limb-specific outcomes on the first and last training sessions.

Custom LabView software (National Instruments, Austin, TX) was developed to calculate kinematic variables (Table 2.2, Table A.1). To determine if stroke-related lower-extremity impairment affected these variables, we compared paretic and non-paretic limb stepping responses from the first training session. In addition, to explore if the stance limb muscle response affected anterior fall-recovery we compared EMG of the paretic and non-paretic stance limb. As to remove the confounding effect of perturbation magnitude, we evaluated successful responses to the largest common perturbation magnitude across limbs. In addition, kinematic and EMG outcomes were compared on the first and the last sessions to evaluate a training effect. Within each initial stepping limb, successful responses to the highest common perturbation magnitude across sessions were evaluated. All comparisons of performance-based and kinematic measures were evaluated using paired t-tests and Cohen's *d* (SPSS 25, IBM, Armonk, NY, alpha = 0.05).

Table 2.2. Anterior fall-recovery kinematic and EMG variables.

Variables	**Definition**
Step length CoM	The anterior distance between the stepping-limb toe marker and the whole-body center of mass at step contact with the treadmill.
Step width CoM	The lateral distance between the stepping-limb toe marker and the whole-body center of mass at step contact with the treadmill.
Peak trunk forward rotation angle	The peak trunk forward rotation angle relative to the standing starting position. Positive values indicate forward trunk rotation relative to the starting position.
Peak trunk forward rotation angular velocity	The peak value of the first time derivative of the trunk forward rotation angle. Positive values indicate trunk forward rotation.
Co-contraction ratio	The integral of the concurrent activity between the stance-limb tibialis anterior and medial gastrocnemius muscles, scaled to pre-disturbance median activity, from disturbance onset until the first step contact.
Muscle onset latency	The time after disturbance onset at which muscle activity in the stance-limb medial gastrocnemius exceeded three standard deviations above the median activity 500 milliseconds before disturbance onset, and was sustained for at least 50 milliseconds.
Peak muscle activation	Peak muscle activation was calculated as the maximum amplitude achieved by the stance-limb medial gastrocnemius muscle within the first step contact, scaled to pre-disturbance median activity.

2.3 Results

Fourteen out of the eighteen participants successfully completed all six training sessions. There were no serious adverse events. Two participants voluntarily withdrew from the study on the first training session, stating that they were not comfortable continuing with the training. Of note, the perturbation magnitudes that these participants experienced did not elicit a step or cause a fall, and they reported no physical discomfort. A third participant performed three fall-recovery training sessions. This participant, however, withdrew from the study due to an unanticipated seizure that occurred outside of our study. A fourth participant only completed five sessions due to an acute illness prior to the sixth session, and scheduling conflicts prevented the rescheduling of the sixth session in a timely manner (greater than 30 days). There was one non-serious, reasonably anticipated event in which a participant delayed training after the second session due to minor muscle soreness in their non-paretic hip. After approximately three days of rest, the participant reported that the soreness had subsided. They resumed training and successfully completed all six sessions without further reports of soreness.

2.3.1 Performance-Based Outcomes

There were no between-limb differences in the proportion of successful responses, or for the highest disturbance magnitude achieved in the first training session (Table 2.3). Across sessions, participants successfully recovered from a higher proportion of falls, and progressed to significantly larger perturbation magnitudes when initially stepping with the paretic limb (all p's $< .065$). During the first session, stepping with the wrong limb caused 24% and 12% of failures for paretic and non-paretic limbs respectively; this was reduced to 10% and 3% in the last session. Five participants experienced falls, fully engaging the safety harness. Observationally, by the last session these five participants successfully recovered from the same perturbation magnitudes that originally caused them to fall (Figure 2.1). Three of these participants only experienced one fall, while the other two participants experienced multiple falls across training. The initial treadmill belt acceleration associated with these falls into the safety harness ranged from 1.5 m/s^2 to 4.5 m/s^2.

Table 2.3. Anterior fall-recovery performance-based outcomes. (n = 15)

Variable	**Initial Step Limb**	**First Session**	***p*-value (Cohen's d)**	**Change w/ training**	***p*-value (Cohen's d)**
% Successful Trials (%)	Non-Paretic	88 (19)	.197 (0.45)	+9 (12)	.011* (1.00)
	Paretic	76 (27)		+13 (17)	.012* (0.72)
Largest Perturbation (m/s^2)	Non-Paretic	3.5 (1.0)	.142 (0.48)	+0.4 (0.8)	.065 (0.54)
	Paretic	2.9 (1.4)		+0.4 (0.4)	.001* (1.00)

Note: Data from the first session, as well as the change observed on the last training session, are displayed as mean (SD). *****Significant ($p < 0.05$) between-session differences from the first session and the last sessions of training.

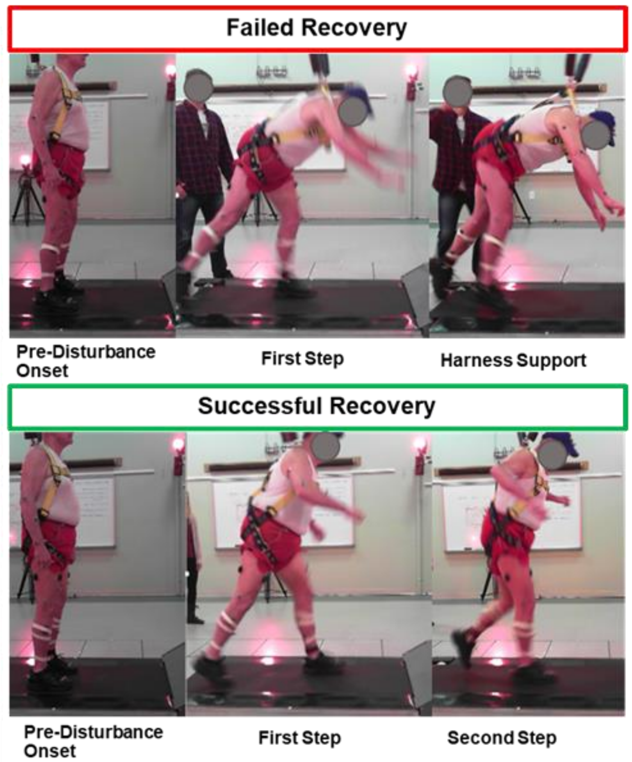

Figure 2.1. An individual with chronic stroke performs trip-recovery training. Treadmill-induced perturbations were applied to standing participants necessitating steps to prevent a fall into the safety harness. The top series of images is from the first training session and depict a failed trip recovery when initially stepping with the non-paretic limb at a perturbation magnitude of 4.25 m/s2. The bottom series of images is from the final training session and depict a successful trip recovery when initially stepping with the non-paretic limb at the same perturbation magnitude that previous caused a fall into the safety harness. Note the amount of forward trunk rotation in the failed recovery compared to the successful recovery.

2.3.2 Kinematic Variables

In the first training session, participants demonstrated significantly wider steps relative to the whole-body CoM when initially stepping with the paretic limb (Table 2.4, Figure A.1B and C). Although not significant, there was a trend of shorter initial step lengths relative to the whole-body CoM when stepping with the paretic limb (Table 2.4, Figure A.1A).

Compensatory step kinematics, as well as trunk kinematics improved from the first to the last sessions at a highest common perturbation magnitude. First step lengths relative to the whole-body CoM became longer when stepping with the paretic limb (Table 2.5, Figure A.3). Peak trunk forward rotation angles were reduced with training when initially stepping with the non-paretic limb (Table 2.5, Figure A.4).

2.3.3 EMG Variables

There were no observed between-limb (Table A.2) or between-session (Table A.4) differences in EMG variables.

Table 2.4. Between-limb anterior fall-recovery kinematic variables on the first training session. (n = 14)

	First Step Limb		
	Non-Paretic	**Paretic**	***p*-value (Cohen's d)**
Step length CoM (cm)	28.1 (5.4)	25.2 (6.2)	.056 (0.61)
Step width CoM (cm)	11.5 (3.2)	14.4 (3.4)	.011* (0.82)
	Second Step Limb		
	Paretic	**Non-Paretic**	***p*-value (Cohen's d)**
Step length CoM (cm)	24.5 (8.6)	26.4 (9.6)	.506 (0.19)
Step width CoM (cm)	15.7 (5.0)	12.3 (3.8)	.011* (0.72)
	Trunk Kinematics		
Peak trunk forward rotation angle (deg)	24.6 (9.3)	31.4 (19.7)	.092 (1.07)
Peak trunk forward rotation angular velocity (deg/s)	104.0 (38.8)	105.8 (30.6)	.823 (0.06)

Note: Non-paretic limb and paretic limb data are displayed as mean (SD). *****Significant ($p < 0.05$) between-limb differences on the first session of training at a common perturbation magnitude between-limbs.

Table 2.5. Between-session anterior fall-recovery kinematic variables on the first and last sessions.

	First Step: Non-Paretic Limb (n = 15)			First Step: Paretic Limb (n = 14)		
	First Session	**Change w/ Training**	***p*-value (Cohen's d)**	**First Session**	**Change w/ Training**	***p*-value (Cohen's d)**
Step length CoM (cm)	29.1 (6.9)	-0.8 (5.5)	.614 (0.14)	23.8 (6.1)	+4.3 (6.7)	.034* (0.66)
Step width CoM (cm)	11.7 (3.4)	-0.9 (3.2)	.315 (0.27)	13.9 (4.7)	-0.9 (4.3)	.423 (0.21)
	Second Step: Paretic Limb			**Second Step: Non-Paretic Limb**		
	First Session	**Change w/ Training**	***p*-value (Cohen's d)**	**First Session**	**Change w/ Training**	***p*-value (Cohen's d)**
Step length CoM (cm)	23.9 (9.2)	+2.7 (7.2)	.174 (0.35)	29.5 (6.8)	+1.0 (7.2)	.623 (0.18)
Step width CoM (cm)	13.7 (3.4)	-1.2 (3.1)	.161(0.43)	11.6 (3.7)	-0.1 (4.0)	.889 (0.34)
	Trunk Kinematics			**Trunk Kinematics**		
	First Session	**Change w/ Training**	***p*-value (Cohen's d)**	**First Session**	**Change w/ Training**	***p*-value (Cohen's d)**
Peak trunk forward rotation angle (deg)	25.2 (8.4)	-2.5 (4.0)	.029* (0.62)	29.0 (12.6)	-3.0 (8.3)	.197 (0.36)
Peak trunk forward rotation angular velocity (deg/s)	114.5 (40.5)	-15.2 (32.0)	.083 (0.46)	108.1 (36.6)	+0.4 (32.3)	.966 (0.02)

Note: First session, last session, and change with training data are displayed as mean (SD). *Significant ($p < 0.05$) between-session differences from the first session and the last sessions of training at a common perturbation magnitude within each limb.

2.4 Discussion

The purpose of this study was to assess the effect of the initial stepping limb (i.e. paretic or non-paretic) on anterior fall-recovery performance, and kinematics, and then determine the benefits of fall-recovery training on those outcomes. We hypothesized that compensatory steps with the paretic limb would be associated with worse fall-recovery performance and kinematics, but such aspects would improve with fall-recovery training. We found that between-limb differences in stepping were most pronounced in the frontal plane, with first and second paretic limb steps being placed wider relative to the whole-body CoM. When stepping initially with the paretic limb, we observed training-based increases in anterior step placement relative to the whole-body CoM. When stepping initially with the non-paretic limb, we observed a reduction in trunk forward rotation. To our knowledge, this is the first report of limb-specific kinematic adaptations in anterior fall recovery for those with chronic stroke.

The initial stepping limb did not influence performance-based variables of fall-recovery (Table 2.3). This is a surprising result given the previous observation that survivors of stroke avoid stepping with their paretic limb [45, 51, 52] as well as the link between paretic limb stepping and subsequent falls in the free-living environment [53]. Perhaps a lack of between-limb differences is due to our sample of relatively high-functioning participants (Table 2.1). Alternatively, the effects of the stepping limb on fall-recovery performance may have been diminished by practice that accompanies progressively difficult repetitions within one training session, as well as a protocol that assessed paretic limb stepping after a progression focused on non-paretic limb stepping. Similar reasons may underlie the lack of significant differences between limbs for sagittal plane kinematics (Table 2.4). However, large between-limb effects on step length and forward trunk rotation angle suggest that detrimental effects of stepping with the paretic limb may exist, especially in a more impaired population. We did observe significant effects of the initial stepping limb on frontal plane kinematics. Participants demonstrated wider steps relative to the whole-body CoM when first stepping with the paretic limb (Table 2.4). Previous studies have verified that stepping to recover from an anterior fall challenges lateral stability, likely due to a limited postural adjustment before toe off [83]. In a study of older adults, those who fell in response to anterior waist pulls demonstrated significantly wider initial compensatory steps compared to those who did not fall [84]. We do not know if altering frontal plane kinematics improves fall-recovery success, nor do we know if this is a modifiable variable in those with chronic stroke. Perhaps the wider steps are a conservative means to maintain stability when on the paretic limb, suggesting that steps should not be narrowed without also improving function of the paretic limb. We suggest that lateral stability either be addressed through a lateral perturbation intervention [52, 85], or, if it is advisable to alter lateral step placement, with the use of biofeedback in conjunction with anterior perturbations.

With training, participants improved their fall-recovery performance. From the first to the last session, participants successfully recovered from a higher proportion of recoveries when stepping with either limb. From the first to the last session, participants progressed to larger perturbation magnitudes only when initially stepping with the paretic limb (Table 2.3). When stepping initially with the paretic limb, first step lengths relative to the whole-body CoM became longer with practice. Our results are similar to previous observations in which individuals with unilateral, lower-extremity amputations improved this variable when stepping with their prosthetic limb

[69]. From these results, it is our interpretation that training adaptation likely occurs in the less-affected stance limb during the first step. Initial step length is an important factor, as it has discriminated fallers from non-fallers in older adults [71] as well as in those with chronic stroke [45]. Previous research has shown that, in response to anterior falls, those with stroke take longer compensatory steps to compensate for reduced trunk control [42]. Of note, the disturbance magnitudes in this previous study were small (20 cm). We anticipated that, with larger disturbances that necessitate longer steps, the ability to compensate for poor trunk control may be diminished. With training, peak trunk forward rotation angles were reduced when initially stepping with the non-paretic limb. The ability to limit forward trunk rotation is critical to successful recovery from an anterior fall [44, 45, 71, 76]. On the last training session, when initially stepping with the non-paretic limb, peak forward trunk rotation occurred at 890 (296) ms after disturbance onset, much after initial step placement 437 (41) ms. So, it is likely that the first as well as the second compensatory steps, not just stance limb plantarflexor activity, played a substantial role in reducing trunk forward rotation. This evidence suggests that exercise interventions aimed to reduce anterior falls in this population should include perturbations that necessitate a multiple-stepping response. The computerized treadmill is specifically suited to provide such perturbations.

We do not know if perturbation-based training can reduce falls in those with chronic stroke. Based on a review of 404 older adults and individuals with Parkinson's disease, such training reduces the risk of falling by 30% (risk ratio 0.71, 95% CI 0.52–0.96) [86]. To our knowledge, one study has evaluated fall-recovery training effects in individuals with chronic stroke [64]. In this randomized controlled trial, perturbation-based training and traditional balance-training groups did not differ in their post-training fall rates. However, limitations of this study included participants with relatively high baseline levels of function. Additionally, the therapist-induced perturbations of the previous study, although more feasible than the treadmill-induced ones of our study, were limited in intensity compared to our approach. We also showed kinematic adaptations to the second recovery step, an aspect which may not have been observed in the previous study, which discouraged multistep responses during training. Therefore, further study is warranted on the benefits of fall-recovery training in this population. It appears that the benefit to fall-recovery may be specific to the fall directions applied during training. Older women who underwent training focused specifically on the trip-recovery response, similar to that reported in this study, reduced the rate of trip-related falling in the laboratory by 86% [87]. This form of training also reduced trip-related falls in the free-living environment (rate ratio 0.54, 95% CI 0.30 – 0.97) despite no effects on the number of stumbles experienced (rate ratio 0.82, 95% CI 0.62 – 1.11) [88]. These results demonstrate that training directly improved the fall-recovery response from a trip, and not the awareness and avoidance of tripping hazards. Of note, not all fall causes were reduced with training. For this reason, we partnered this training with one focused on posterior fall-recovery, which is presented in Chapter 3 of this dissertation.

Our study was limited in that we did not conduct a controlled experiment, so we cannot conclude that changes in our outcomes were due to the training itself. It may be that these aspects improved due to confounding influences, such as interactions with study staff or the general benefits of more activity due to study participation. Given that the effects of stroke are dependent on the injury location and severity, initial fitness of the person, and intensity of previous rehabilitation, this population presents with a wide range of function. Aspects such as lower-extremity impairment, or age may alter responsiveness to our training. These factors, then, would serve as ways to stratify groups in a controlled experiment. Further study is needed to identify such factors.

Another limitation of our study is that many of our participants were high-functioning and active survivors of stroke (Table 2.1). Therefore, we must continue to evaluate the feasibility of this training with lower-functioning participants, particularly those with a high fear of falling, low falls self-efficacy, and those that rely on walking aids such as a cane or walker. Walking aids are commonly used by those with stroke [89, 90]. However, the effectiveness and utility of using a walking aid to recover from a fall is not well understood. In some cases, using a cane has been shown to impede compensatory steps needed to successfully recovery from lateral [91] and posterior [92] falls. In contrast, those with Parkinson's disease improved their postural recovery in response to an unpracticed simulated slip while using a cane [93]. Although, the beneficial effects of using the cane were only observed during the initial perturbation exposure. The feasibility, effectiveness, and utility of training those with stroke who rely on walking aids requires further study.

In conclusion, our study demonstrates the specific means by which anterior fall recovery can be modified with practice in those with chronic stroke. This is the first study to do so using large perturbations requiring multiple steps to regain stability. This aspect is important given the similarity of kinematics resulting from treadmill-induced falls and those resulting from an overground trip [44, 45, 71, 76], a common cause of falling in this population [21]. Training that can improve the ability of those with chronic stroke to respond to an anterior fall may serve as a means to prevent injury and enable independence. Further study is required to determine if this form of training is effective at reducing trip-related falls in the free-living environment in survivors of stroke.

Chapter 3

POSTERIOR FALL-RECOVERY TRAINING APPLIED TO INDIVIDUALS WITH CHRONIC STROKE

3.1 Introduction

Chronic stroke is the leading cause of long-term disability in the United States [1]. Those with chronic stroke have a fall risk that is twice that of age- and sex-matched peers [20]. As many as 75% of those living with stroke fall each year [6, 17, 75], and approximately 84% of fractures in this population are due to an accidental fall [23]. Trips and slips cause one-third of post-stroke falls [21]. Therefore, interventions that improve the reaction to these common fall causes are likely to reduce the rate of falls and their related injuries.

Individuals living with chronic stroke have an impaired posterior fall response. Participants with chronic stroke had slip-recovery steps that were shorter and closer to the whole-body center of mass (CoM), resulting in a 71% failure rate compared to 0% in controls [43]. These trends were replicated with treadmill-induced posterior falls [42]. Compared to peers with no stroke history, survivors of stroke have lower posterior multiple-stepping thresholds, defined as the smallest perturbation magnitudes that elicited more than one step [41]. In a study that induced posterior falls, those with stroke who fell into a safety harness had delayed muscle activations in the paretic limb [50]. Such deficiencies were associated with more posterior trunk rotation, resulting in the trunk center of mass positioned further outside the base of support. From these studies, it is apparent that stroke-related neural impairment alters fall-recovery kinematics, likely affecting the capacity to recover successfully.

The asymmetry of stroke-related function likely influences posterior fall-recovery performance. Individuals with chronic stroke prefer to step with their non-paretic limb [94], often avoiding a step with the paretic limb [95]. Against our expectations, previous evidence suggests that posterior fall-recovery performance, as measured by slip-recovery success rates [95] or posterior multiple-stepping thresholds [41], *is not affected* by the initial stepping limb (i.e. paretic or non-paretic). Two previous studies have assessed differences between paretic and non-paretic posterior step kinematics [95, 96]. In response to a standing perturbation, no between-limb differences were observed in the delay in step initiation, as measured by a force plate [96]. Using a between-subjects design, a second study confirmed no limb-specific differences in post-slip step initiation time between limbs [95]. In this study, however, slip-recovery steps with the paretic limb were placed in a more stable position relative to the whole-body CoM [95]. This result is surprising given the general preference for stepping with the non-paretic limb. The greater stability of paretic-limb steps was attributed to better reactive control of the non-paretic stance limb. We suggest that, given the wide range of function present in those with chronic stroke, a within-participant comparison is needed to better understand the role that the stepping limb has on fall-recovery performance and kinematics. Such within-participant comparisons have revealed that, during a feet-in-place response, the paretic limb had a diminished, delayed muscle response characterized by co-contraction [37, 48, 49, 97–104]. Muscle activity of the stance limb during posterior stepping, however, has not been evaluated.

Previous exercise interventions aimed at reducing falls in survivors of stroke have focused on strength, gait, balance, flexibility, and endurance. However, meta-analyses of these exercise modalities found no reduction in the rate of falls or the risk of falling [54, 55, 64]. These previous approaches lack processing specificity [57]. In other words, the required motor responses to arrest a fall are not provoked or effectively modified using traditional modes of exercise. Successful recovery from a slip-induced fall is often dependent upon the skill of compensatory stepping [105], with the recovery step placed behind the whole body CoM, but not too far laterally from it [105–108]. This demand can be recreated using anteriorly directed treadmill-belt accelerations that necessitate posterior steps to arrest a fall [73, 109]. Given that treadmill-induced falls closely resemble slips, practicing recovery from such falls serves as a *specific* means to improving the response to common fall causes in this population. In those without stroke, such training has reduced fall rates from laboratory-induced slips [73], and has reduced falls in the free-living environment [110]. In a laboratory study of repeated slips to the non-paretic limb, individuals with stroke were able to modify the reactive response on the second trial [62]. Within-session adaptations included a more stable position of the whole-body CoM at toe-off of the recovery step, as well as longer steps. The extent to which the posterior stepping response of those with chronic stroke can be improved with practice over multiple sessions, as well as the limb-specific benefits of such practice, is not well understood.

The purpose of this study was to assess the effects of the initial stepping limb (i.e. paretic or non-paretic) on posterior fall-recovery performance and kinematics, as well as to determine the benefits of fall-recovery training on those outcomes. We hypothesized that compensatory steps with the paretic limb would be associated with worse fall-recovery performance and kinematics. We also hypothesized that such aspects would improve with fall-recovery training. Performance was quantified as the proportion of successful recoveries within a series of perturbations, as well as the highest perturbation magnitude within that series. “Worse” kinematic features included shorter and wider recovery steps relative to the CoM, as well as larger peak trunk backward rotation angles and angular velocities [105–108, 111]. In order to explore the neuromuscular mechanisms that may underlie observed between-limb or between-session differences in fall-recovery, we evaluated EMG of the non-stepping, stance limb plantar and dorsiflexors when feasible to do so. We expected the paretic limb to be characterized by a delayed response with less dorsiflexor activity and more co-contraction. We also expected that training would result in less delay, more dorsiflexor activity, and less co-contraction.

3.2 Methods

3.2.1 Participants

From the University of Delaware’s Stroke Studies Registry, we recruited eighteen participants with chronic stroke. Exclusion criteria included other neurologic disorders, musculoskeletal surgeries within the past year, recent cardiovascular events (past three months), or other conditions that preclude safe participation. Participants had a self-reported ability to walk a city block without a gait aid such as a walker or cane. 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) [77]. This screening criterion, which has been used previously in studies of older adults [78], was conservatively in place to reduce the risk of fractures from the impact of fall-recovery steps or falls into the safety harness. Of note, two participants wore articulating ankle foot orthosis during training that they typically wore on a day-to-day basis. We anticipated that removing the orthosis for training may have presented an unreasonable injury risk to the foot or ankle. This study was approved by the University of Delaware's Institutional Review Board, and all participants provided written informed consent prior to participation.

3.2.2 Training Protocol

Our training was garnered from a previous protocol applied to those with no stroke impairment [73]. The perturbations delivered within our training, which consisted of anterior treadmill belt translations, were designed to necessitate rapid posterior steps similar to that of slip-recovery [73, 109]. All participants attempted to complete six sessions of the training protocol. Sessions consisted of two progressions of treadmill belt perturbations (ActiveStep®, Simbex, Lebanon, NH). Progressions within a training session focused on recovery steps with the non-paretic limb (Figure 3.1) or paretic limb. These progressions were limited to either 10 minutes or 18 perturbations, whichever occurred first, with rest periods lasting approximately five minutes between each progression. In addition, two progressions focused on anterior fall recovery were delivered within each session prior to the posterior fall progressions. The results from these former progressions are reported in Chapter 2 of this dissertation. Progression durations were designed to reasonably limit fatigue, keeping training sessions to within one hour. Six training sessions occurred over approximately three weeks.

Participants wore their own well-cushioned, closed-toe athletic shoes with no elevated heels. They were outfitted with a full-body safety harness (Delta™, Capital

Safety, Bloomington, MN) attached to an overhead rail. Support straps were adjusted so that the participant's hands and knees could not contact the treadmill. The harness was instrumented with a force transducer (Dillon, Fairmont, MN), the peak forces of which were recorded for each trial.

When awaiting a perturbation, participants stood self-supported on the treadmill, feet at a comfortable width with toes evenly positioned in the anteroposterior direction (Figure 3.1). Perturbation velocity waveforms were triangular in shape, consisting of 200 ms acceleration and deceleration phases. Participants were instructed to "try to recover in one step" in response to these posterior falls. This step constraint was established because slip recovery is primarily dictated by the first-step features [106], as well as to address the observation that individuals with stroke tend to take more recovery steps compared to those with no stroke [41, 43, 112]. The first perturbation of each progression had an initial acceleration of 0.5 m/s^2, resulting in a displacement of 0.01 m. After a successful recovery, the subsequent perturbation had an initial acceleration +0.5 m/s^2 greater than the previous perturbation [69]. After a failed recovery, the subsequent trial acceleration was reduced by 0.5 m/s^2. Failures were defined as recoveries in which the force transducer recorded more than 20% body weight [79], recoveries in which the participant stepped with the wrong limb, or recoveries in which two or more steps were taken. Non-stepping responses were permissible. Treadmill displacements were 0.64 m or less, peak velocities were 3.2 m/s or less, and peak accelerations were 16 m/s^2 or less. Each perturbation was preceded by a 1-5 s delay, and small perturbations (0.05 m displacement) resulting in an anterior fall were introduced approximately once every six trials to limit anticipatory adjustments. Participants were asked to inform research staff if the training intensity became too much for them to tolerate (i.e. muscle soreness, general fatigue, or uneasiness being on the treadmill). In such cases, training continued at the highest perturbation magnitude tolerated for the remainder of the session. This approach was intended to limit nervousness or discomfort and maintain compliance while promoting practice repetitions.

All trials were recorded with a 12 camera motion capture system operating at 120 Hz (Motion Analysis®, Santa Rosa, CA, replaced mid-study with Qualisys®, Göteborg, Sweden). The positions of 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.

Muscle activity was recorded using surface electromyography (Delsys, Natick, MA, 1200 Hz) of the bilateral ankle dorsi- and plantar-flexors [50, 80], where four wireless bipolar surface electrodes were placed bilaterally on the medial gastrocnemius (MG) and tibialis anterior (TA). Sensors were oriented in the direction of the muscles' fibers and at half the distance between the motor endpoint and the distal end of each muscle [81], Unfiltered EMG signals were shifted to account for a 48 ms delay between EMG sensors and kinematic data (as per Delsys equipment documentation). The signals were then de-meaned, bandpass filtered (10 - 300 Hz), rectified, and lowpass filtered (8th order Butterworth) at 50 Hz for muscle onset latency and 4 Hz for peak activation and co-contraction ratios. A higher frequency cutoff was used for muscle onset latency due to the effect of filtering on time-based measures of muscle activity [82].

3.2.3 Analysis

Fall-recovery performance was quantified from the proportion of successful recoveries and the highest perturbation magnitude achieved. In order to determine how lower-extremity impairment affected performance at baseline, we compared the separate progressions of stepping with the paretic or non-paretic limb within the first training session. In order to evaluate if these measures changed with training, we compared limb-specific outcomes on the first and last training sessions.

Custom LabView software (National Instruments, Austin, TX) was developed to calculate kinematic and EMG variables (Table 3.1, Table B.1). In order to determine if stroke-related lower-extremity impairment affected fall-recovery kinematics, we compared paretic and non-paretic limb stepping responses from the first training session. In addition, to explore if the stance limb muscle response affected posterior fall-recovery we compared EMG of the paretic and non-paretic stance limb. As to remove the confounding effect of perturbation magnitude, we evaluated successful responses to the largest common perturbation magnitude across limbs. In addition, kinematic and EMG outcomes were compared on the first and the last sessions to evaluate a training effect. Within each initial stepping limb, successful responses to the highest common perturbation magnitude across sessions were evaluated. All comparisons of performance-based, kinematic and EMG measures were evaluated using paired t-tests and Cohen's *d* (SPSS 25, IBM, Armonk, NY, alpha = 0.05).

Table 3.1. Posterior fall-recovery kinematic and EMG variables.

Variables	**Definition**
Step length CoM	The posterior distance between the stepping-limb heel marker and the whole-body center of mass at step contact with the treadmill.
Step width CoM	The lateral distance between the stepping-limb heel marker and the whole-body center of mass at step contact with the treadmill.
Peak trunk backward rotation angle	The peak trunk backward rotation angle relative to the standing starting position. Negative values indicate backward trunk rotation relative to the starting position.
Peak trunk backward rotation angular velocity	The peak value of the first time derivative of the trunk backward rotation angle. Negative values indicate trunk backward rotation.
Peak trunk backward rotation angular velocity	The peak value of the first time derivative of the trunk backward rotation angle. Negative values indicate trunk backward rotation.
Co-contraction ratio	The integral of the concurrent activity between the stance-limb tibialis anterior and medial gastrocnemius muscles, scaled to pre-disturbance median activity, from disturbance onset until the first step contact.
Muscle onset latency	The time after disturbance onset at which muscle activity in the stance-limb tibilias anterior exceeded three standard deviations above the median activity 500 milliseconds before disturbance onset, and was sustained for at least 50 milliseconds.
Peak muscle activation	Peak muscle activation was calculated as the maximum amplitude achieved by the stance-limb tibialis anterior muscle within the first step contact, scaled to pre-disturbance median activity.

3.3 Results

Thirteen (10 males, 3 females) of 18 participants completed at least five of six training sessions and were included in this analysis (Table 3.2). As part of protocol development, our first participant's training consisted of posterior falls delivered while walking. Although our treadmill is able to deliver walking perturbations relative to gait events, it is not able to discriminate left and right steps. Therefore, we could not administer a limb-specific, progressively challenging series of perturbations. Starting with the second subject, we changed the training protocol so that posterior falls were induced while the participant was standing. Two participants voluntarily withdrew from the study on the first training session, stating that they were not comfortable continuing with the training. Of note, the perturbation magnitudes that these participants experienced did not elicit a step or cause a fall, and they reported no physical discomfort. A third participant performed three fall-recovery training sessions. This participant, however, withdrew from the study due to an unanticipated seizure that occurred away from the laboratory. One participant did not perform posterior fall-recovery progressions on the first day of training.

Table 3.2. Demographic and clinical assessment data

Measure	**Mean (SD), Range**
Age (Years)	59 (12), 29 – 77
BMI (kg/m^2)	28.9 (3.9), 22.0 – 33.9
Years after stroke	5 (3.5), 2 – 15
Fugl-Meyer LE	24 (6), 8 – 32
Activities Specific Balance Confidence Scale (ABC)	91 (8), 76 – 100
Functional Gait Assessment (FGA)	17 (6), 9 – 29
Berg Balance Scale (BBS)	51 (7), 36 – 56

Note: Prior to starting fall-recovery training, descriptive measures of the Fugl-Meyer Lower Extremity assessment [115], Activities-Specific Balance Confidence (ABC) scale [116], Berg-Balance Scale [117], and the Functional Gait Assessment[118] were used to characterize our participants balance and mobility.

3.3.1 Initial Stepping-Limb Effects

Initial compensatory steps with the non-paretic limb were associated with about a 25% greater success rate (Table 3.3). However, the initial stepping limb did not affect the highest perturbation magnitude successfully achieved in the first session (Table 3.3). Steps with the paretic limb were, on average, placed more than twice as far laterally from the CoM as steps with the non-paretic limb (Table 3.4, Figure B.1B).

No between-limb differences were observed in sagittal-plane trunk or step kinematic variables. We observed between-limb differences in the stance limb EMG. The paretic limb in stance had significantly greater co-contraction and delays in tibialis anterior activity (Table 3.4). It should be noted, the between-limb comparison sample size was only seven for the kinematic variables, and only five for the EMG variables (Table 3.4). The smaller sample size for kinematic and EMG variables was because, on the first training session, 6 of the 13 participants were unable to execute a compensatory step with their paretic limb at a common perturbation magnitude to that of the non-paretic limb. In other words, there were not common perturbation magnitudes in which a step was taken between the paretic and non-paretic limbs in the first session. In addition, the two participants wearing AFOs did not have EMG sensors placed on their paretic limb due to the orthotic. Therefore, we were unable to compare between-limb fall-recovery kinematics or EMG of interest for these participants.

Table 3.3. Posterior fall-recovery performance-based outcomes. (n = 13)

Variable	**Initial Step Limb**	**First Session**	***p*-value (Cohen's d)**	**Change w/ training**	***p*-value (Cohen's d)**
% Successful Trials (%)	Non-Paretic	81 (13)	.015* (1.27)	+8 (13)	.049* (0.54)
	Paretic	55 (27)		+15 (18)	.014* (0.74)
Largest Perturbation (m/s^2)	Non-Paretic	3.5 (1.3)	.131 (0.44)	+0.6 (0.6)	.003* (1.00)
	Paretic	2.7 (1.3)		+0.4 (0.4)	.042* (0.61)

Note: Data from the first session, as well as the change observed on the last training session, are displayed as mean (SD). ***Significant ($p < 0.05$) between-session differences from the first session and the last sessions of training.

Table 3.4. Between-limb posterior fall-recovery kinematic and EMG variables on the first session.

	First Step Kinematics (n = 7)		
	Non-Paretic	**Paretic**	***p*-value (Cohen's d)**
Step length CoM (cm)	20.8 (3.8)	22.2 (10.2)	.742 (0.26)
Step width CoM (cm)	4.9 (3.1)	12.4 (3.8)	.011* (1.50)
	Trunk Kinematics (n = 7)		
	Non-Paretic	**Paretic**	***p*-value (Cohen's d)**
Peak trunk backward rotation angle (deg)	-8.7 (4.4)	-11.3 (8.9)	.469 (0.48)
Peak trunk backward rotation angular velocity (deg/s)	-72.6 (25.6)	-70.9 (37.4)	.907 (0.05)
	Stance Limb EMG (n = 5)		
	Paretic	**Non-Paretic**	***p*-value (Cohen's d)**
Co-contraction ratio	0.54 (0.48)	0.10 (0.09)	.069 (8.37)
Agonist muscle onset latency (ms)	103 (22)	65 (23)	.0002* (5.94)
Peak muscle activation	23.52 (24.07)	41.60 (33.61)	.268 (0.51)

Note: Non-paretic-limb and paretic-limb data are displayed as mean (SD). Between-limb kinematic comparisons were limited to 7 participants, as opposed to 13, because 6 participants did not have similar-magnitude perturbations between stepping limbs. In other words, they did not step with the paretic limb, or they only stepped with the paretic limb in response to small perturbations that did not elicit a step with the non-paretic limb. Between-limb EMG comparisons were further limited to 5 participants because 2 participants wore ankle foot orthosis that prevented EMG sensor placement. *****Significant ($p < 0.05$) between-limb differences on the first session of training at a common perturbation magnitude between-limbs.

3.3.2 Training-Based Changes

From the first to last sessions, participants increased the proportion of successful recoveries and progressed to larger perturbation magnitudes (Table 3.3). Participants increased their proportion of successful recoveries primarily by improving their ability to step with the correct limb (non-paretic or paretic). On the first session, when initially stepping with the paretic limb, approximately 34% of responses were failures due to stepping with the wrong limb, 10% of responses were failures due to taking more than one step, and 1% of responses were failures due to falls into the safety harness. On the last session, failures due to stepping with the wrong limb were reduced to 22%, failures due to taking more than one step were reduced to 8%, and failures due to falls into the safety harness were reduced to 0%. On the first session, when initially stepping with the non-paretic limb, 1% of responses were failures due to stepping with the wrong limb, and 18% of responses were failures were due to taking more than one recovery step. On the last session, failures due to stepping with the wrong limb were reduced to 0% and failures due to taking more than one step were reduced to 11%. One participant experienced three falls into the safety harness. Observationally, this participant later successfully recovered from the same perturbation magnitudes that originally caused them to fall (Figure 3.1). Across all participants, there were no significant changes in kinematic or EMG variables between the first and last sessions (Table 3.5). Figure representations of our results with participant-specific representations are available in the Appendix B. Additional trunk and step kinematic variables, none of which demonstrated significant between-limb or between-session effects, are also presented in the Appendix B.

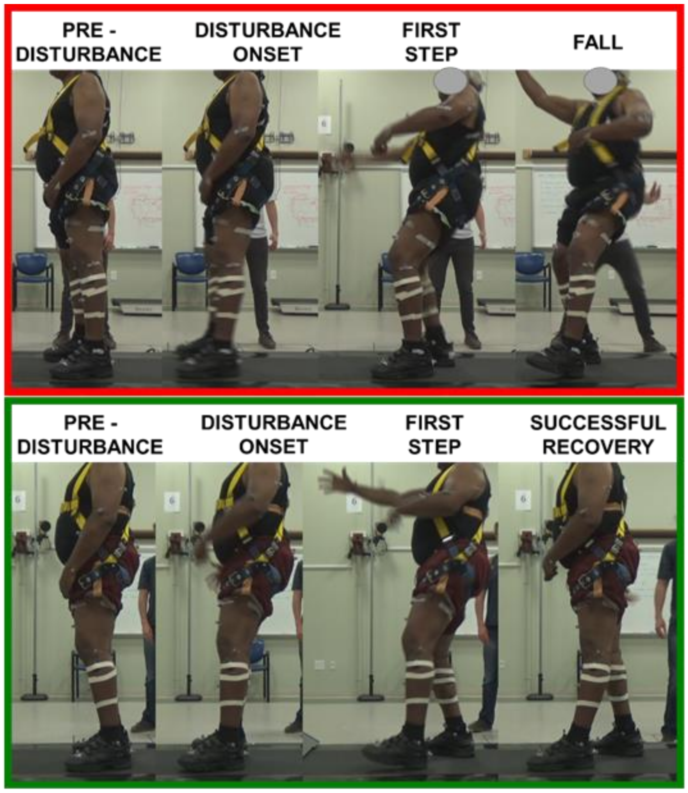

Figure 3.1. An individual with chronic stroke performs slip-recovery training. Treadmill-induced perturbations were applied to standing participants necessitating steps to prevent a fall into the safety harness. The top series (red) shows a posterior fall during the first training session stepping with the non-paretic limb. The bottom series (green) shows a successful fall-recovery at the same initial belt acceleration of 5.0 m/s^2 on the sixth training session stepping with the non-paretic limb.

Table 3.5. Between-session posterior fall-recovery kinematic and EMG variables on the first and last sessions.

	First Step Kinematics: Non-Paretic (n = 13)			First Step Kinematics: Paretic (n = 7)		
	First Session	**Change w/ Training**	***p*-value (Cohen's d)**	**First Session**	**Change w/ Training**	***p*-value (Cohen's d)**
Step length CoM (cm)	19.5 (8.0)	-1.2 (6.1)	.509 (0.18)	23.0 (9.6)	-4.0 (8.1)	.239 (0.48)
Step width CoM (cm)	6.7 (4.5)	-1.7 (5.4)	.274 (0.27)	11.7 (5.0)	-0.7 (3.1)	.577 (0.21)
	Trunk Kinematics			**Trunk Kinematics**		
	First Session	**Change w/ Training**	***p*-value (Cohen's d)**	**First Session**	**Change w/ Training**	***p*-value (Cohen's d)**
Peak trunk backward rotation angle (deg)	-6.7 (8.4)	2.9 (5.3)	.071 (0.51)	-10.6 (7.2)	-1.3 (6.1)	.601 (0.25)
Peak trunk backward rotation angular velocity (deg/s)	-70.3 (30.3)	8.8 (38.5)	.426 (0.34)	-83.7 (55.6)	-15.6 (30.2)	.221 (0.63)
	Stance Limb EMG: Paretic (n = 8)			**Stance Limb EMG: Non-Paretic (n = 7)**		
	First Session	**Change w/ Training**	***p*-value (Cohen's d)**	**First Session**	**Change w/ Training**	***p*-value (Cohen's d)**
Co-contraction ratio	0.35 (0.25)	-0.02 (0.36)	.879 (0.07)	0.29 (0.47)	-0.06 (0.26)	.589 (0.29)
Agonist muscle onset latency (ms)	103 (59)	+51 (174)	.434 (0.52)	85 (23)	+5 (47)	.750 (0.15)
Peak muscle activation	26.58 (22.26)	-3.92 (16.53)	.524 (0.23)	39.81 (38.21)	-4.07 (25.77)	.691 (0.15)

Note: First session and change with training data are displayed as mean (SD). Between-session kinematic comparisons when initially stepping with the paretic limb were limited to 7 participants, as opposed to 13, because 6 participants did not have similar-magnitude perturbations between training sessions. Between-session EMG comparisons of the paretic stance limb were limited to 8 participants because 2 participants wore ankle foot orthosis that prevented EMG sensor placement and 3 participants did not have similar-magnitude perturbations between sessions. *Significant ($p < 0.05$) between-session differences from the first session and the last sessions of training at a common perturbation magnitude within each limb.

3.4 Discussion

The purpose of this study was to investigate the effect of the initial stepping limb (i.e. paretic or non-paretic) on posterior fall-recovery performance and kinematics, and then determine the benefits of fall-recovery training on those outcomes. We hypothesized that compensatory steps with the paretic limb would be associated with worse fall-recovery performance and kinematics, but such aspects would improve with fall-recovery training. This hypothesis was partially supported. At baseline, initial steps with paretic limb were associated with less frequent success and altered frontal plane kinematics. We observed notable performance-based improvements over the course of training, but these observations were not aligned with changes in relevant recovery kinematics. In order to understand the neuromuscular mechanisms that may underlie the observed between-limb and between-session differences in fall-recovery, we evaluated EMG of the stance limb. On the first session, we observed between-limb differences in the stance limb EMG activity, but those differences do not appear to be a primary factor in fall-recovery performance or sagittal plane kinematics.

Stepping with the paretic limb was associated with a lower proportion of successful recoveries (Table 3.3). On the first session, the majority of failed responses were due to an inability to initiate steps with the paretic limb (i.e. steps with the wrong limb). This observation aligns with the previously observed, “aborted” steps of the paretic limb in response to a slip [95]. Although we saw between-limb differences in the proportion of successful recoveries, we did not observe between-limb differences in the largest successful perturbation response (Table 3.3). This result aligns with a previous study that found no between-limb differences in posterior multiple-stepping thresholds in this population [41]. The observed discrepancy in between-limb effects on performance variables in our study may be due to within-session adaptation on the first day. In other words, the highest perturbation magnitude achieved in a session does not reflect the failed responses that were observed earlier in that session. Such rapid, trial-to-trial adaptation of posterior fall-recovery has been previously observed in this population [62]. Such within-session adaptation may also underlie the observed lack of between-limb differences in sagittal-plane kinematics, as our analysis focused on the response to larger perturbations that occurred later in the training session.

Between-limb differences in step kinematics were primarily observed in the frontal plane, with paretic-limb steps placed wider relative to the whole-body CoM. After a slip, a wider step with respect to the CoM decreases the likelihood of successful recovery in individuals with no previous stroke [106]. So, the wider steps characteristic of the paretic limb may be problematic. Compared to volitional stepping, perturbation-evoked stepping is characterized by smaller postural adjustments before step initiation, resulting in more lateral CoM displacement during the step [83]. In turn, wider steps are needed to regain lateral stability. Perhaps the wider steps of the paretic limb are a conservative means to increase lateral stability with that step, as the paretic limb likely has a reduced capacity to generate stabilizing forces upon step completion. Participants demonstrated greater co-contraction, and had longer muscle onset latencies in the paretic stance limb. These results align with similar research that found that during a feet-in-place response, the paretic stance limb had a diminished, delayed muscle response characterized by co-contraction [37, 48, 49, 97–104]. However, the altered muscle response of our participants' paretic limb in stance did not affect fall-recovery performance or sagittal-plane kinematics. Perhaps there could be compensation from the non-paretic limb in swing, or it could be that ankle musculature does not play a substantial role in recovering from a posterior fall with a step.

With practice, those with chronic stroke recovered from a higher proportion of perturbations, and they recovered from larger perturbation magnitudes. These results are similar to a recent report that demonstrating that those with chronic stroke were able to recover from more and larger treadmill-induced falls as a result of their perturbation-based training [113]. In some cases, the participants corrected the inability to take a recovery step with the paretic limb. This ability is relevant, as slips occurring outside the laboratory may require initial steps with either limb, depending on which limb is perturbed. Accordingly, the inability to take a recovery step with the paretic limb has been prospectively related to falls in the free-living environment [53]. Therefore, training effective steps with the paretic limb may directly address an underlying source of the high fall-risk associated with survivors of stroke. Post hoc, between-limb comparisons of the proportion of successful recoveries at the end of training suggest that, despite training based improvements when stepping with the paretic limb, a significant between-limb difference in the proportion of successful recoveries persisted (p = .008, d = 1.40). So, our approach did not eliminate the between-limb disparity in fall-recovery performance associated with paretic-limb steps.

Despite performance-based improvements in fall-recovery, we did not observe kinematic or EMG adaptations. By only comparing successful responses, we may not be considering large enough perturbations to observe training-based benefits. However, we could not practically compare successful responses to failed recoveries, as the latter included non-stepping responses or steps with the wrong limb. It may be that, with our single-step constraint, kinematic adaptations are limited in those with chronic stroke. In a previous study, young adults lengthened their posterior step within a single training session of perturbations, while those with stroke did not change their step kinematics [112]. Instead, those with stroke *shortened* their initial step length, with a preference towards multistep responses. Although a multistep response may be effective when no step constraints are given, it may have limited efficacy in response to falls that are dependent on first step features, such as the response to a slip-induced fall [106]. It may also be that our selected variables did not capture the underlying mechanisms behind performance-based improvements. Undetected mechanisms could include alterations in stepping limb kinetics after foot contact, such as those needed to prevent limb collapse [114]. Alternatively, we may be altering psychological aspects such as the confidence to maintain the resulting posture without a second step (i.e. preventing unnecessary second steps).

Although we have demonstrated that our training has performance-based benefits to posterior fall-recovery, our study is limited by the high-functioning levels of our participants (Table 3.2). Therefore, we must continue to evaluate the feasibility and effectiveness of this training with lower-functioning participants who are likely at the highest risk of falling. In addition, we were not able to identify underlying kinematic mechanisms by which fall-recovery was improved. We demonstrated that, compared to non-paretic limb steps, paretic-limb steps were wider with respect to the whole body CoM. This variable was not implicitly responsive to fall-recovery training, yet it remains to be seen whether narrowing the step width would be beneficial in these circumstances. We suggest that, given these initial results, future studies explore the role that fall-recovery training can play as a means to reduce falls and enable mobility in this population.

3.5 Conclusions

The purpose of this study was to determine the effects of lower limb, stroke-related impairment on anterior and posterior fall-recovery performance, and then determine the benefits of exercise focused specifically on improving fall-recovery skill. We hypothesized that compensatory steps with the paretic limb would be associated with worse fall-recovery performance and kinematics. We also hypothesized that such aspects would improve with specific fall-recovery training.

Our hypotheses were partially supported. There were no between-limb differences in *anterior* fall-recovery performance in the first session, however, there were between-limb differences in compensatory step placement. In response to an anterior fall, steps with the paretic limb were wider relative to the center of mass compared to steps with the non-paretic limb (p = .011, d = 0.72). Previous studies have verified that stepping to recover from an anterior fall challenges lateral stability [83, 84]. We do not know if altering frontal plane kinematics improves fall-recovery success, nor do we know if this is a modifiable variable in those with chronic stroke. Perhaps the wider steps are a conservative means to maintain lateral stability when on the paretic limb, suggesting that steps should not be narrowed without also improving function of the paretic limb.

With training, participants successfully recovered from a higher proportion of anterior falls (p's = .011, Cohen's d's > 0.73), as well as progressed to larger perturbation magnitudes (p's < .065, d's > 0.54). Initial paretic limb steps became longer (p = .034, d = 0.66), and trunk forward rotation was reduced when first stepping with the non-paretic limb (p = .029, d = 0.62). Our results align with previous observations in which individuals with unilateral lower-extremity amputations, another unilaterally impaired population, improved their anterior fall-recovery response with a similar training program [69]. Improvements from simulated-trip training included an increased initial step length and a reduced trunk flexion angle, with benefits limited to steps with the prosthetic limb [69]. The initial compensatory step length is an important factor, as it has discriminated fallers from non-fallers in older adults [71] as well as in those with chronic stroke [45].

There were between-limb difference in *posterior* fall-recovery performance in the first session. Initial posterior steps with the non-paretic limb were associated with a higher proportion of success than initial steps with the paretic limb ($p = .015$, Cohen's d = 1.26). In the first session, the majority of failed responses were due to an inability to initiate steps with the paretic limb (i.e. steps with the wrong limb). This observation aligns with the previously observed, "aborted" steps of the paretic limb in response to a slip [95]. Although we saw between-limb differences in the proportion of successful recoveries, we did not observe between-limb differences in the largest successful perturbation response. This result aligns with a previous study that found no between-limb differences in posterior multiple-stepping thresholds in this population [41].

In response to a posterior fall, steps with the paretic limb were wider relative to the CoM ($p = .011$, d = 1.50). After a slip, a wider step with respect to the CoM decreases the likelihood of successful recovery in individuals with no previous stroke [106]. So, the wider steps characteristic of the paretic limb may be problematic. We do not know if, with more explicit feedback, lateral step placement is a modifiable variable in those with chronic stroke. In addition, we do not know if narrowing the step would improve fall-recovery success. With training, participants successfully recovered from a higher proportion posterior falls (p's < .049, d's > 0.54), as well as progressed to larger perturbation magnitudes (p's < .042, d's > 0.61). In some cases, the participants corrected the inability to take a recovery step with the paretic limb. This ability is relevant, as slips occurring outside the laboratory may require initial steps with either limb, depending on which limb is perturbed. Accordingly, the inability to take a recovery step with the paretic limb has been prospectively related to falls in the free-living environment [53]. There were no significant changes in kinematic variables with posterior fall-recovery training (p's > .071, d's < 0.51). In a previous study, young adults lengthened their posterior step within a single training session of perturbations, while those with stroke did not change their step kinematics [112]. Instead, those with stroke *shortened* their initial step length, with a preference towards multistep responses. Although a multistep response may be effective when no step constraints are given, it may have limited efficacy in response to falls that are dependent on first step features, such as the response to a slip-induced fall [106].

Our study was limited by not being a controlled experiment, so our results should be interpreted with this in mind. Our outcomes were variables measured within training sessions. Therefore, the addition of reliable, yet precise pre-training and post-training balance measures would be needed to conduct such a study. Without an active control group, we cannot conclude that benefits would be due to the training itself. It may be that confounding influences, such as interactions with study staff or the general benefits of more activity underlie training improvements. Many of our participants were high-functioning and active individuals (Table 2.1). Therefore, we cannot assume that our training protocol is feasible with lower functioning participants. The feasibility, effectiveness, and utility of training those with a high fear of falling, low falls self-efficacy, and those that rely on walking aids such as a cane or walker requires further study.

In conclusion, the initial stepping limb affects relevant step kinematics during anterior and posterior fall recovery of high-functioning individuals with chronic stroke. We demonstrated that, because we saw performance-based and kinematic adaptations to the fall-recovery response, our fall-recovery training is a potentially beneficial exercise intervention for those with chronic stroke. Anterior fall-recovery training improved performance and select kinematic outcomes. Although this study provides evidence that the skill of posterior stepping in response to a fall can be improved with practice in those with chronic stroke, we were not able to identify the underlying kinematic mechanisms behind this adaptation. Further study is required to determine if this form of training is effective at reducing trip- and slip-related falls in the free-living environment in survivors of stroke. We intend to extend this work by evaluating the benefits of applying this training to a larger cohort with more impaired function, and evaluating its effects on balance self-confidence, walking activity, and subsequent falls in the free-living environment.

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Appendix A

ANTERIOR FALL-RECOVERY SUPPLEMENTAL DATA

Table A.1. Anterior fall-recovery kinematic variables

Variables	**Definition**
Step length	The anterior distance between the stepping-limb toe marker and the stance-limb toe marker at step contact with the treadmill.
Step length CoM	The anterior distance between the stepping-limb toe marker and the whole-body center of mass at step contact with the treadmill.
Step width	The lateral distance between the stepping-limb toe marker and the stance-limb toe marker at step contact with the treadmill.
Step width CoM	The lateral distance between the stepping-limb toe marker and the whole-body center of mass at step contact with the treadmill.
Step time	The time from the perturbation onset to step contact with the treadmill.
Peak trunk forward rotation angle	The peak trunk forward rotation angle relative to the standing starting position. Positive values indicate forward trunk rotation relative to the starting position.
Peak trunk forward rotation angle time	The time at which peak trunk forward rotation angle occurs relative to perturbation onset.
Peak trunk forward rotation angular velocity	The peak value of the first time derivative of the trunk forward rotation angle. Positive values indicate trunk forward rotation.
Peak trunk forward rotation angular velocity time	The time at which peak trunk forward rotation angular velocity occurs relative to perturbation onset.
Trunk forward rotation angle at first heel strike	The trunk forward rotation angle relative to the standing starting position at first heel strike. Positive values indicate trunk forward rotation.
Trunk forward rotation angular velocity at first heel strike	The first time derivatives of the trunk forward rotation angle were calculated to obtain the trunk forward rotation angular velocity from which the value was recorded at the first heel strike. Positive values indicate trunk forward rotation.

Table A.2. Between-limb anterior fall-recovery kinematic and EMG variables in the first session.

	First Step Limb		
	Non-Paretic	**Paretic**	***p*-value (Cohen's d)**
Step length (cm)	52.6 (14.0)	52.0 (18.7)	.878 (0.05)
Step length CoM (cm)	28.1 (5.4)	25.2 (6.2)	.056 (0.61)
Step width (cm)	28.5 (6.2)	29.9 (5.2)	.407 (0.21)
Step width CoM (cm)	11.5 (3.2)	14.4 (3.4)	.011* (0.82)
Step time (ms)	452 (44)	341 (530)	.451 (1.76)
	Second Step Limb		
	Paretic	**Non-Paretic**	***p*-value (Cohen's d)**
Step length (cm)	46.7 (18.4)	45.3 (21.3)	.751 (0.09)
Step length CoM (cm)	24.5 (8.6)	26.4 (9.6)	.506 (0.19)
Step width (cm)	27.7 (7.2)	26.9 (6.6)	.608 (0.14)
Step width CoM (cm)	15.7 (5.0)	12.3 (3.8)	.011* (0.72)
	Trunk Kinematics		
	Non-Paretic	**Paretic**	***p*-value (Cohen's d)**
Peak trunk forward rotation angle (deg)	24.6 (9.3)	31.4 (19.7)	.092 (1.07)
Peak trunk forward rotation angle time (ms)	799 (247)	860 (227)	.361 (0.24)
Peak trunk forward rotation angular velocity (deg/s)	104.0 (38.8)	105.8 (30.6)	.823 (0.06)
Peak trunk forward rotation angular velocity time (ms)	402 (116)	517 (204)	.124 (0.62)
Trunk forward rotation angle at first heel strike (deg)	15.8 (7.8)	15.3 (9.6)	.866 (0.03)
Trunk forward rotation angular velocity at first heel strike (deg/s)	18.5 (54.7)	-5.9 (21.2)	.151 (0.31)
	Stance Limb EMG		
	Paretic	**Non-Paretic**	***p*-value (Cohen's d)**
Co-contraction ratio	1.1 (0.8)	1.2 (1.1)	.791 (0.08)
Agonist muscle onset latency (ms)	114 (38)	121 (49)	.516 (0.26)
Peak muscle activation	14.5 (11.8)	15.6 (12.9)	.850 (0.06)

Note: Non-paretic limb and paretic limb data are displayed as mean (SD). ***Significant ($p < 0.05$) between-limb differences on the first session of training at a common perturbation magnitude between-limbs.

Table A.3. Between-session anterior fall-recovery kinematic variables from the first and last sessions.

	First Step: Non-Paretic Limb			**First Step: Paretic Limb**		
	First Session	**Change w/ Training**	***p*-value (Cohen's d)**	**First Session**	**Change w/ Training**	***p*-value (Cohen's d)**
Step length (cm)	56.8 (18.3)	+0.03 (10.4)	.991 (0.03)	53.9 (15.9)	+5.9 (13.0)	.111 (0.41)
Step length CoM (cm)	29.1 (6.9)	-0.8 (5.5)	.614 (0.14)	23.8 (6.1)	+4.3 (6.7)	.034* (0.66)
Step width (cm)	28.3 (6.0)	-1.9 (5.6)	.208 (0.39)	29.0 (5.7)	-1.5 (5.2)	.295 (0.34)
Step width CoM (cm)	11.7 (3.4)	-0.9 (3.2)	.315 (0.27)	13.9 (4.7)	-0.9 (4.3)	.423 (0.21)
Step time (ms)	326 (493)	+112 (450)	.401 (0.15)	482 (50)	+31 (96)	.247 (0.41)
	Second Step: Paretic Limb			**Second Step: Non-Paretic Limb**		
	First Session	**Change w/ Training**	***p*-value (Cohen's d)**	**First Session**	**Change w/ Training**	***p*-value (Cohen's d)**
Step length (cm)	51.4 (20.0)	+6.3 (9.8)	.027* (0.62)	52.5 (22.3)	-1.4 (15.1)	.735 (0.09)
Step length CoM (cm)	23.9 (9.2)	+2.7 (7.2)	.174 (0.35)	29.5 (6.8)	+1.0 (7.2)	.623 (0.18)
Step width (cm)	26.3 (9.2)	-2.2 (4.1)	.051* (0.54)	24.8 (7.6)	-0.5 (8.1)	.809 (0.06)
Step width CoM (cm)	13.7 (3.4)	-1.2 (3.1)	.161 (0.43)	11.6 (3.7)	-0.1 (4.0)	.889 (0.34)
	Trunk Kinematics			**Trunk Kinematics**		
	First Session	**Change w/ Training**	***p*-value (Cohen's d)**	**First Session**	**Change w/ Training**	***p*-value (Cohen's d)**
Peak trunk forward rotation angle (deg)	25.2 (8.4)	-2.5 (4.0)	.029* (0.62)	29.0 (12.6)	-3.0 (8.3)	.197 (0.36)
Peak trunk forward rotation angle time (ms)	907 (295)	-20 (324)	.815 (0.06)	827 (159)	+91 (240)	.178 (0.56)
Peak trunk forward rotation angular velocity (deg/s)	114.5 (40.5)	-15.2 (32.0)	.083 (0.46)	108.1 (36.6)	+0.4 (32.3)	.966 (0.02)
Peak trunk forward rotation angular velocity time (ms)	483 (240)	+32 (304)	.688 (0.12)	518 (177)	-22 (257)	.747 (0.09)
Trunk forward rotation angle at first heel strike (deg)	14.6 (8.1)	-2.8 (7.6)	.179 (0.31)	15.4 (7.9)	+0.67 (7.4)	.741 (0.11)
Trunk forward rotation angular velocity at first heel strike (deg/s)	8.3 (52.0)	+7.9 (42.5)	.481 (0.18)	-2.5 (33.0)	-18.1 (34.2)	.069 (0.53)

Note: First session, last session, and change with training data are displayed as mean (SD). *****Significant ($p < 0.05$) between-session differences from the first session and the last sessions of training at a common perturbation magnitude within each limb.

Table A.4. Between-session anterior fall-recovery EMG variables from the first and last sessions.

	Non-Paretic Stance Limb (n = 11)			Paretic Stance Limb (n = 12)		
	First Session	**Change w/ Training**	***p*-value (Cohen's d)**	**First Session**	**Change w/ Training**	***p*-value (Cohen's d)**
Co-contraction ratio	0.8 (0.5)	-0.01 (0.89)	.967 (0.43)	0.9 (0.8)	-0.15 (0.74)	.492 (0.169)
Agonist muscle onset latency (ms)	110 (36)	+16 (138)	.715 (0.32)	132 (65)	-14 (69)	.488 (0.172)
Peak muscle activation (%)	14.1 (6.8)	+1.4 (14.1)	.757 (0.15)	15.3 (7.1)	-0.4 (7.9)	.866 (0.080)

Note: First session, last session, and change with training data are displayed as mean (SD). *****Significant ($p < 0.05$) between-session differences from the first session and the last sessions of training at a common perturbation magnitude within each limb.

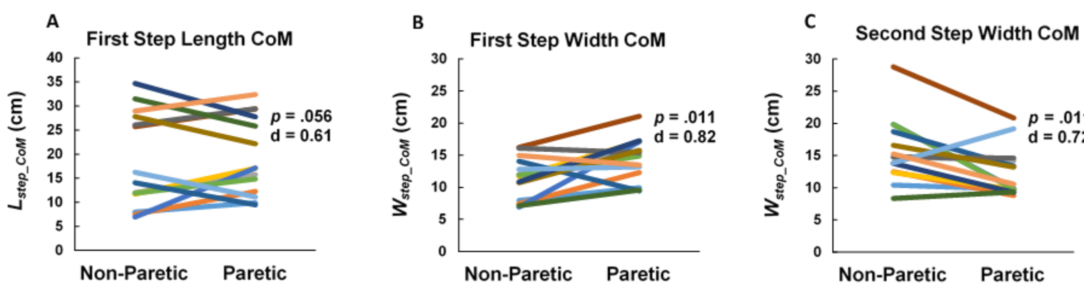

Figure A.1. First step lengths (A), first step widths (B), and second step widths (C) relative to the whole-body center of mass on the first training session at a common perturbation magnitude when stepping with the non-paretic and the paretic limbs.

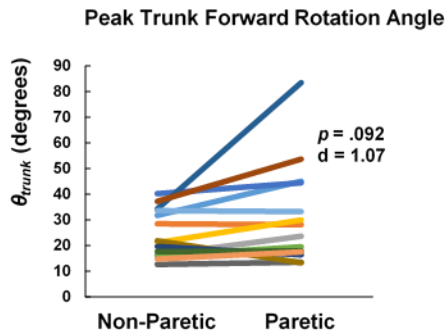

Figure A.2. Peak trunk forward rotation angles on the first training session at a common perturbation magnitude when stepping with the non-paretic and the paretic limbs.

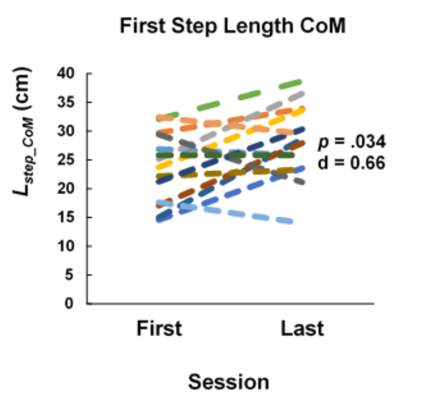

Figure A.3. First step lengths relative to the whole-body center of mass when stepping with the paretic limb on the first and the last session.

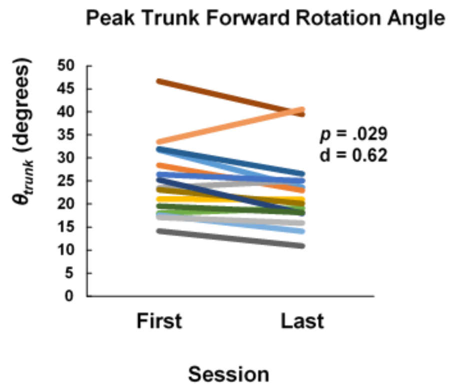

Figure A.4. Peak trunk forward rotation angles when stepping with the non-paretic limb on the first and the last training session.

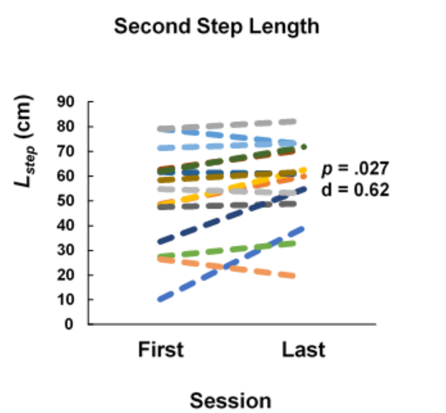

Figure A.5. Second step lengths when stepping with the paretic limb on the first and the last session.

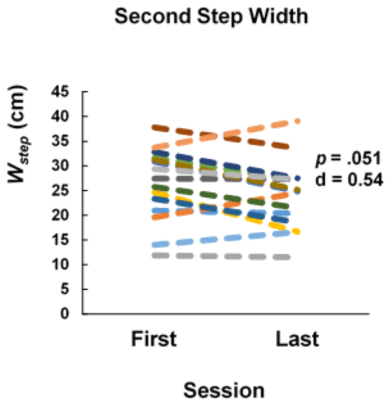

Figure A.6. First step widths when stepping with the paretic limb on the first and the last session.

Appendix B

POSTERIOR FALL-RECOVERY SUPPLEMENTAL DATA

Table B.1. Posterior fall-recovery kinematics variables.

Variables	**Definition**
Step length	The posterior distance between the stepping-limb heel marker and the stance-limb heel marker at step contact with the treadmill.
Step length CoM	The posterior distance between the stepping-limb heel marker and the whole-body center of mass at step contact with the treadmill.
Step width	The lateral distance between the stepping-limb heel marker and the stance-limb heel marker at step contact with the treadmill.
Step width CoM	The lateral distance between the stepping-limb heel marker and the whole-body center of mass at step contact with the treadmill.
Step time	The time from the perturbation onset to step contact with the treadmill.
Peak trunk backward rotation angle	The peak trunk backward rotation angle relative to the standing starting position. Negative values indicate backward trunk rotation relative to the starting position.
Peak trunk backward rotation angle time	The time at which peak trunk backward rotation angle occurs relative to perturbation onset.
Peak trunk backward rotation angular velocity	The peak value of the first time derivative of the trunk backward rotation angle. Negative values indicate trunk backward rotation.
Peak trunk backward rotation angular velocity time	The time at which peak trunk backward rotation angular velocity occurs relative to perturbation onset.
Trunk backward rotation angle at first heel strike	The trunk backward rotation angle relative to the standing starting position at first heel strike. Negative values indicate trunk backward rotation.
Trunk backward rotation angular velocity at first heel strike	The first time derivatives of the trunk backward rotation angle were calculated to obtain the trunk backward rotation angular velocity from which the value was recorded at the first heel strike. Negative values indicate trunk backward rotation.

Table B.2. Between-limb posterior fall-recovery kinematic variables on the first training session.

	First Step Limb		
	Non-Paretic	**Paretic**	**p-value (Cohen's d)**
Step length (cm)	22.9 (8.1)	22.1 (14.0)	.888 (0.08)
Step length CoM (cm)	20.8 (3.8)	22.2 (10.2)	.742 (0.26)
Step width (cm)	17.8 (5.8)	21.2 (7.6)	.404 (0.39)
Step width CoM (cm)	4.9 (3.1)	12.4 (3.8)	.011* (1.50)
Step time (ms)	536 (121)	585 (89)	.323 (0.36)
	Trunk Kinematics		
Peak trunk backward rotation angle (deg)	-8.7 (4.4)	-11.3 (8.9)	.469 (0.48)
Peak trunk backward rotation angle time (ms)	554 (192)	675 (96)	.107 (0.61)
Peak trunk backward rotation angular velocity (deg/s)	-72.6 (25.6)	-70.9 (37.4)	.907 (0.05)
Peak trunk backward rotation angular velocity time (ms)	357 (122)	458 (133)	.195 (0.57)
Trunk backward rotation angle at first heel strike (deg)	-3.7 (5.9)	-8.5 (8.7)	.382 (0.44)
Trunk backward rotation angular velocity at first heel strike (deg/s)	-7.4 (33.5)	-31.8 (36.1)	.266 (0.48)

Note: Non-paretic limb and paretic limb data are displayed as mean (SD). *****Significant ($p < 0.05$) between-limb differences on the first session of training at a common perturbation magnitude between-limbs.

Table B.3. Between-session posterior fall-recovery kinematic variables on the first training session.

	First Step: Non-Paretic Limb (n = 13)			**First Step: Paretic Limb (n = 7)**		
	First Session	**Change w/ Training**	**p-value (Cohen's d)**	**First Session**	**Change w/ Training**	**p-value (Cohen's d)**
Step length (cm)	23.1 (13.9)	+1.7 (10.5)	.563 (0.16)	23.8 (13.9)	-3.8 (8.5)	.286 (0.45)
Step length CoM (cm)	19.5 (8.0)	-1.2 (6.1)	.509 (0.18)	23.0 (9.6)	-4.0 (8.1)	.239 (0.48)
Step width (cm)	20.7 (8.0)	-2.5 (8.3)	.297 (0.38)	20.7 (9.2)	-0.3 (6.7)	.897 (0.05)
Step width CoM (cm)	6.7 (4.5)	-1.7 (5.4)	.274 (0.27)	11.7 (5.0)	-0.7 (3.1)	.577 (0.21)
Step time (ms)	492 (62)	+62 (202)	.290 (0.77)	607 (103)	+27 (66)	.317 (0.40)
	Trunk Kinematics			**Trunk Kinematics**		
	First Session	**Change w/ Training**	**p-value (Cohen's d)**	**First Session**	**Change w/ Training**	**p-value (Cohen's d)**
Peak trunk backward rotation angle (deg)	-6.7 (8.4)	2.9 (5.3)	.071 (0.51)	-10.6 (7.2)	-1.3 (6.1)	.601 (0.25)
Peak trunk backward rotation angle time (ms)	622 (304)	-262 (352)	.020 (0.70)	668 (94)	+4 (134)	.928 (0.03)
Peak trunk backward rotation angular velocity (deg/s)	-70.3 (30.3)	8.8 (38.5)	.426 (0.34)	-83.7 (55.6)	-15.6 (30.2)	.221 (0.63)
Peak trunk backward rotation angular velocity time (ms)	497 (141)	-6 (214)	.916 (0.04)	485 (149)	-40 (108)	.371 (0.33)
Trunk backward rotation angle at first heel strike (deg)	2.4 (8.6)	+1.8 (5.6)	.276 (0.30)	-8 (6.8)	-1.7 (5.5)	.439 (0.49)
Trunk backward rotation angular velocity at first heel strike (deg/s)	-14.1 (57.2)	-6.1 (64.9)	.740 (0.12)	-21.5 (45.5)	-12.1 (54.3)	.576 (0.21)

Note: First session, last session, and change with training data are displayed as mean (SD). *****Significant ($p < 0.05$) between-session differences from the first session and the last sessions of training at a common perturbation magnitude within each limb.

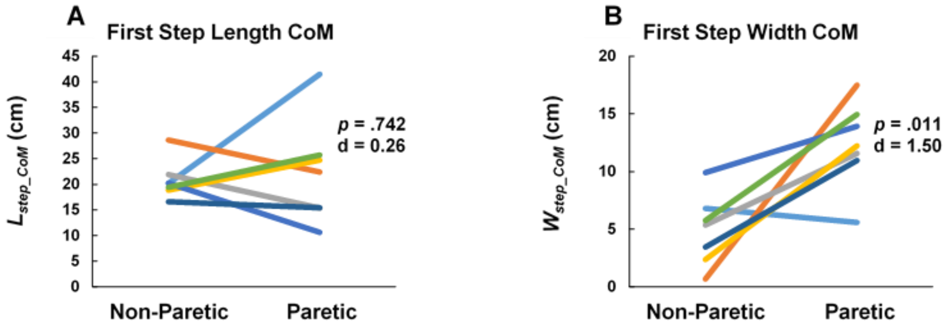

Figure B.1. First step lengths (A), first step widths (B) relative to the whole-body center of mass on the first training session at a common perturbation magnitude when stepping with the non-paretic and the paretic limbs.

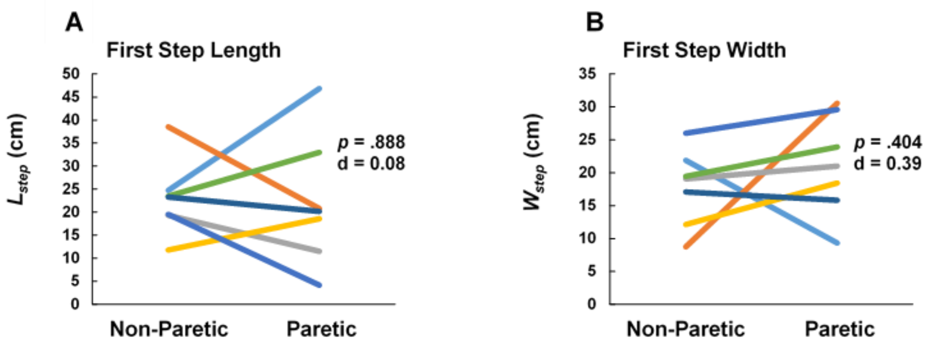

Figure B.2. First step lengths (A), first step widths (B) relative to stance-limb toe on the first training session at a common perturbation magnitude when stepping with the non-paretic and the paretic limbs.

Appendix C

INSTITUTIONAL REVIEW BOARD APPROVAL LETTER

UNIVERSITY OF DELAWARE

RESEARCH OFFICE

210 Hullihen Hall
University of Delaware
Newark, Delaware 19716-1551
Ph: 302/831-2136
Fax: 302/831-2828

DATE: November 5, 2018

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

STUDY TITLE: [791562-9] Fall Recovery Training for Individuals with Chronic Stroke

SUBMISSION TYPE: Amendment/Modification

ACTION: APPROVED
APPROVAL DATE: November 5, 2018
EXPIRATION DATE: August 18, 2019
REVIEW TYPE: Expedited Review

REVIEW CATEGORY: Expedited review per 45 CFR 46.110 (b)(2)

Thank you for your submission of Amendment/Modification 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.

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