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DYNAMIC BALANCE CONTROL DURING TREADMILL WALKING IN CHRONIC STROKE SURVIVORS

by

Eric Walker, B.S.

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Marquette University,
in Partial Fulfillment of the Requirements for
the Degree of Doctor of Philosophy

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ABSTRACT
DYNAMIC BALANCE CONTROL DURING TREADMILL WALKING
IN CHRONIC STROKE SURVIVORS

Eric Walker, B.S.

Marquette University, 2013

Maintaining dynamic balance is an important component of walking function that is likely impaired in chronic stroke survivors, evidenced by an increased prevalence of falls. Dynamic balance control requires maintaining the center of mass (COM) within the base of support during movement. During walking, dynamic balance control is achieved largely by modifying foot placement to adjust the base of support. However, chronic stroke survivors have difficulty with both precision control of foot placement, as well as reduced control of COM movement. The objective of this dissertation was to characterize dynamic balance control strategies during walking in chronic stroke survivors. Additionally, we evaluated whether altered sensory feedback could be used to improve balance control in stroke survivors. Dynamic balance control was characterized during challenging walking conditions in stroke survivors and age-matched neurologically intact individuals. Adaptations to perturbations in frontal plane COM, induced using a custom cable-driven device, were used to further probe mechanisms of dynamic balance control. Despite larger amounts of COM movement and step widths, chronic stroke survivors produced a similar ratio of step width to COM sway, indicating that simply increasing step width does not produce a safer walking pattern for the stroke group. Placement of the paretic limb was unchanged in response to the external perturbations of trunk movement, which might underlie deficits in dynamic balance control. Augmented sensory feedback improved paretic foot placement and COM control, when applied during a stepping or treadmill walking task. These results provide insight into differences in dynamic balance control in stroke while also demonstrating that augmented sensory feedback signals might be used to improve balance control, and thus walking function for chronic stroke survivors.

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CHAPTER 1: INTRODUCTION AND BACKGROUND

1.1 INTRODUCTION

The ability to walk is one of the most important factors for individuals returning to work after suffering a stroke (Vestling *et al.*, 2003). Each year about 610,000 new people suffer a stroke, and about 30% are unable to walk without some assistance (Go et al., 2013). After a stroke, walking ability is related to the individual's ability to control balance (Michael *et al.*, 2005). Balance control is an even stronger predictor of walking ability in individuals that walk more slowly, and have more gait impairments (Patterson et al., 2007). A survey of stroke survivors found that within 6 months of discharge from a rehabilitation center 46% of individuals fell at least once, and 39% of the total number of falls occurred during walking (Mackintosh *et al.*, 2005). The incidence of falls during walking in stroke survivors after traditional rehabilitation protocols, and the importance of balance control in determining walking function, demonstrates a need to further explore changes in walking balance control post-stroke.

The purpose of this dissertation is to characterize modifications in dynamic balance control strategy during walking in chronic stroke survivors. Additionally, we will evaluate whether altered sensory feedback can improve balance control during walking. This chapter will discuss balance mechanisms during walking, as well as sensory and motor deficits that may contribute to impaired balance control post-stroke.

1.2 BALANCE CONTROL

1.1.1 Control of Standing Balance

In general, standing balance can be maintained by keeping the body's center of mass (COM) within the base of support. Forces produced at the ankle and/or hip result in moments about the COM, which act to control COM location in both the sagittal and frontal planes (Winter, 1995). The base of support is determined by stance width, which can vary from person to person (McIlroy & Maki, 1997), but is unchanged during quiet stance. Wider stance widths increase the base of support and reduce the hip force necessary to control frontal plane movement of the COM, but also increase demands on the neural control mechanisms to maintain stability (Bingham *et al.*, 2011). Additionally, stance width impacts energy expenditure during the task (Donelan *et al.*, 2001), and changes the relative contribution of the hip and ankle joints to the maintaining standing balance (Gatev *et al.*, 1999). In addition to determining stance width, control of COM and center of pressure (COP) position can be utilized to maintain balance during quiet stance. Analysis of these movements during stance provides insight into dynamic control strategies used to maintain balance. The magnitude of COP movements provides insight into the precision of balance control, while the consistency of the trajectories indicates the level of focus devoted to balance control (Donker *et al.*, 2007). Control of COM movement is dependent upon the quality and amount of sensory feedback (Oie *et al.*, 2002), the integrity of the neural control system (Maki & McIlroy, 1996), as well as the task demands (Brown *et al.*, 2002). Examination of standing balance control helps to characterize the steady-state performance of the underlying neural control mechanisms.

1.1.2 Maintaining Balance during Walking

Although the focus of both standing and walking balance control remains the same, to keep the COM within the base of support, the control mechanisms are different (Winter, 1995). Forward progression of the COM, and changes in the size and position of the base of support increases the difficulty of maintaining balance during walking. However, walking balance control may be simplified by focusing resources on maintaining balance in the frontal plane. Modeling of leg movement during walking demonstrates an inherent stability in the sagittal plane that can adjust to small perturbations without requiring direct control, but the unstable frontal plane would require active control (Kuo, 1999). These modeling results are supported by larger amounts of foot placement variability in the frontal plane, compared to the sagittal plane, observed during walking with and without the eyes closed (Bauby & Kuo, 2000). Therefore, it is likely that dynamic balance control strategy during walking is largely focused on maintaining stability in the frontal plane.

Similar to quiet stance, maintaining lateral balance during walking can be achieved through controlling COM movement, or through modifying foot placement to adjust the base of support. During walking, the most effective control strategy is to adjust lateral foot placement based upon the COM position and velocity (Hof *et al.*, 2007; Hof, 2008). Young control subjects walked with larger step widths on a treadmill compared to overground, but maintained a similar minimum lateral separation between the COM and the edge of the base of support (Rosenblatt & Grabiner, 2010). This lateral separation was consistent even when perturbations of lateral trunk movement were applied during swing (Hof *et al.*, 2010). Additionally, during treadmill walking in healthy young and old individuals, step width variability was strongly related to variations in COM position and acceleration (Hurt *et al.*, 2010). These studies demonstrate that dynamic balance control during walking is focused upon a lateral foot placement strategy.

1.1.3 Planning, Execution, and Control of Movement

Balance control during walking is dependent upon sensory feedback to both spinal and supraspinal networks to successfully plan and execute movement (for review see Nielsen, 2003). Vestibular, visual, and proprioceptive feedback signals are important for providing information about body position and orientation in the environment. These sensory signals must be integrated with the motor commands to adjust balance control strategy to the current task demands (Lockhart & Ting, 2007). Furthermore, a lateral foot placement strategy requires descending drive from supraspinal centers to ensure proper endpoint control during targeted movements. Animal models have implicated both the motor cortex (Metz & Whishaw, 2002; Friel *et al.*, 2007) and posterior parietal cortex (Lajoie & Drew, 2007) in the precision control of foot placement. During walking, sensory feedback at the spinal level is needed to produce within-step adjustments to the walking pattern in response to changes in the environment (af Klint *et al.*, 2008). Thus, these studies demonstrate that multiple networks contribute to the successful planning, execution, and correction of lower extremity movements.

1.3 BALANCE IMPAIRMENTS POST-STROKE

1.1.4 Reduced COM Control

After a stroke, sensory and motor deficits impair the coordination of movements across the entire body. This reduced movement coordination impacts the ability of the individual to control movement of the COM (Mansfield *et al.*, 2011). Stroke survivors demonstrate increased levels of frontal plane COM movement during quiet stance relative to elderly controls, and this difference is further emphasized when individuals were asked to stand with their eyes closed (Marigold & Eng, 2006a). Additional reductions in postural control post-stroke were observed after stroke survivors completed extended period of walking (Carver *et al.*, 2011). These studies demonstrate that deficits in COM control are larger when the task demands increase. This reduced control is present even though stroke survivors focus more cognitive resources on the balance task (Roerdink et al., 2006). Evaluation of COM control during standing can be used to characterize changes in balance control over the course of rehabilitation (Kirker *et al.*, 2000), as well as differences between fallers and non-fallers post-stroke (Marigold & Eng, 2006a).

Control of COM movement is also impaired during dynamic movements, but few studies have evaluated COM movements in the context of dynamic balance control. Chern et al. (2010) used a full-body reaching task (bend down, pick up object, return to standing) to explore differences in dynamic postural control. They observed that stroke subjects demonstrated larger COM movements and velocities during the task, and were less likely to shift weight onto the paretic limb. This study demonstrates changes in dynamic standing balance control post-stroke, but differences in dynamic balance control during walking are still relatively unexplored. Reduced ability to accurately sense trunk position after movement, would contributed to deficits in dynamic COM control (Ryerson *et al.*, 2008). Stroke survivors have difficulty coordinating movement across body segments during normal walking (Hacmon *et al.*, 2012) and when changing walking direction (Hollands et al., 2010), contributing to altered COM movement during walking. These studies provide evidence suggesting that stroke survivors have difficulty controlling COM movement during walking. This reduced control would impair their ability to maintain balance during movement, and may also contribute to changes in dynamic balance control strategy.

1.1.5 Impaired Foot Placement Control

In addition to deficits in controlling COM movement, foot placement control is also impaired in stroke survivors. Stroke survivors have difficulty making visually guided foot placement corrections, and these deficits were largest when attempting to make medial corrections (Nonnekes et al., 2010). These foot placement control deficits persisted even when support was provided to remove the balance control constraints from the task. The persistence of this reduced control when support was provided indicates that foot placement deficits are likely a contributing factor to altered balance control post-stroke, and not just a result of poor balance. Reduced foot placement control contributes to altered stepping patterns during an obstacle avoidance task, and these altered foot placement locations may compromise balance during the task ((Said *et al.*, 2001)). During normal walking, impaired foot placement control may contribute to increased step widths (Chen *et al.*, 2005*b*), and asymmetrical foot placement in the frontal plane

(Balasubramanian *et al.*, 2010). These locomotor changes are typically associated with stroke survivors selecting a more cautious walking pattern, since both modifications would increase separation between the COM and edge of the base of support. However, further investigation is necessary to understand how these impairments impact dynamic balance control, and walking function post-stroke.

1.1.6 Impact Upon Walking Function

Maintaining dynamic balance is important for the successful completion of many daily activities. After a stroke, an individual's ability to walk in the community has a large impact on their perceived quality of life (Lord *et al.*, 2004). Walking ability post-stroke is at least partially predicted by balance control (Michael et al., 2005; Patterson et al., 2007), but these studies evaluated balance during quiet stance. Changes in dynamic balance control strategy modify the walking pattern of chronic stroke survivors, which could negatively impact walking function. For example, stroke survivors walk with larger step widths, which may be necessary to adjust the base of support due to larger amounts of COM movement during walking. Wider step widths require more energy expenditure (Donelan et al., 2001), further increasing the already high metabolic cost of walking post-stroke (Detrembleur *et al.*, 2003). This increased energy expenditure would cause the individual to fatigue more quickly, limiting the duration of the walking bouts, and further reducing walking function for chronic stroke survivors. Additionally, balance control is further reduced as the stroke survivor begins to fatigue, which would increase the fall risk during walking. Therefore, gait modifications intended to maintain balance may have unintended effects that could reduce walking function and balance control. Characterization of dynamic balance control mechanisms post-stroke will provide valuable knowledge that may be used to help improve walking function for chronic stroke survivors.

1.4 SPECIFIC AIMS

1.1.7 Aim 1: Sensory electrical stimulation improves foot placement during targeted stepping post-stroke

Reductions in the precision control of paretic foot placement likely contribute to impairments in balance control. This reduced foot placement control would greatly limit the effectiveness of using a lateral foot placement strategy to maintain the COM within the base of support during walking. One potential mechanism to improve foot placement control, is to augment sensory feedback signals from the paretic limb. In this aim, we evaluated whether somatosensory stimulation of the paretic foot would improve foot placement during a targeted stepping task. During the targeted stepping task, participants initiated movement with the non-paretic limb, and stepped to one of five target locations projected onto the floor with distances normalized to the paretic stride length. This task enabled the comparison of precision foot placement control of the paretic limb within a stepping movement. Targeting error and lower extremity kinematics were used to assess changes in foot placement and limb control due to somatosensory stimulation. We hypothesized that electrical stimulation of the paretic foot, applied during the task, would decrease foot-targeting error and improve lower extremity kinematics.

1.1.8 Aim 2: Dynamic balance control strategies in stroke survivors.

After a stroke, individuals have an increased fall risk, especially during walking, which can lead to injuries further impairing mobility. Multiple studies have evaluated balance control during standing, but few have examined changes in dynamic balance control post-stroke. The goal of this study was to characterize dynamic balance control strategy by assessing walking performance during challenging walking conditions. Experimental conditions challenging walking performance were created by either removing visual feedback from the lower visual field, or by having to complete a a moving, and stationary head-targeting task while walking. Changes in locomotor performance were compared across the walking conditions, and between ten chronic stroke and ten age-matched neurologically intact individuals. We hypothesized that visual feedback of body movement would reduce frontal plane COM movement in chronic stroke survivors during walking, with the largest improvements found when a stationary reference was provided

1.1.9 Aim 3: Locomotor adaptations to frontal plane trunk perturbations in young adults.

Balance control responses to environmental factors can involve locomotor modifications aimed at increasing the base of support and/or reducing COM movement. In order to better understand balance control responses during walking, a novel cable driven device was constructed to directly perturb COM movement. This device enabled the characterization of balance control responses to changes in COM movement in the frontal plane. Locomotor adaptations to continuous frontal plane perturbations of trunk motion were evaluated during treadmill walking in ten young, healthy adults. Lower limb kinematics and kinetics were used to characterize modifications to different perturbation types (accentuating vs. resisting), perturbation magnitudes, and the impact of holding onto a handrail hold. We hypothesized that individuals utilize a lateral foot placement strategy to maintain dynamic balance, increasing step width for accentuating perturbations, and decreasing step width for resisting perturbations.

1.1.10 Aim 4: Locomotor adaptations to continuous, external perturbations of the trunk in stroke survivors.

Sensory and motor deficits post-stroke impact balance control during walking, and likely produce changes in dynamic balance control strategy. One potential strategy to compensate for reduced foot placement control is to focus upon controlling COM motion. We utilized the same cable-driven device from Aim 3 to perturb trunk movement during walking in chronic stroke survivors. These perturbations required the individual to modify their walking pattern in order to maintain balance. These locomotor modifications were compared to those made my age-matched, neurologically intact individuals, to characterize potential changes in the underlying dynamic balance control strategy post-stroke. We hypothesized that altered balance control strategy post-stroke would result in reduced foot placement adjustments in response to external perturbations of frontal plane trunk motion during walking.

CHAPTER 2: SENSORY ELECTRICAL STIMULATION IMPROVES FOOT PLACEMENT DURING TARGETED STEPPING POST-STROKE

2.1 INTRODUCTION

The precision control of foot placement location is an important component of locomotion. For example, step-by-step modification of foot placement is important for dynamic balance control during walking (Hof et al., 2007; 2010), and much of the focus of this control is centered upon the frontal plane (O'Connor & Kuo, 2009). Additionally, accurate control of foot placement is important for adapting the walking pattern to environmental conditions, such as when stepping over obstacles. This control of foot placement requires the integration of visual and proprioceptive feedback signals, and involves brain structures such as the primary motor cortex (Bretzner & Drew, 2005) and posterior parietal cortex (Marigold *et al.*, 2011). After stroke, damage to these and other brain structures can disrupt sensorimotor integration, impairing the control of foot placement during stepping.

Impairment in sensorimotor control of foot placement might substantially impact walking function in stroke survivors. Walking dysfunction post-stroke includes slower walking speeds (Turnbull *et al.*, 1995), decreased walking endurance (Michael et al., 2005), and increased risk of falls (Mackintosh et al., 2005). Impairments in control of foot placement appear to contribute to these functional losses. For example, foot placement asymmetries in both the frontal and sagittal plane during walking correlate with functional impairments post-stroke (Balasubramanian et al., 2010). Additionally, stroke survivors modify foot placement location relative to an obstacle, providing additional time for the paretic limb to clear the obstacle, but also potentially compromising balance (Said et al., 2001). Stroke survivors also have difficulty making medial foot placement adjustments mid-step; however, their ability to make these adjustments improves when balance assistance is provided during the task (Nonnekes et al., 2010). These studies demonstrate that the control of foot placement is associated with balance control and walking function. Therefore, increased walking function might be achieved through techniques aimed at improving foot placement control in stroke survivors.

Augmenting sensory feedback provides a potential mechanism to improve foot placement. Somatosensory electrical stimulation applied to the paretic wrist improves hand function for a period of time after stimulation in stroke survivors (Wu *et al.*, 2006). Applying vibratory stimulation to the paretic wrist during movement improves endpoint stability during both planar reaching (Conrad *et al.*, 2011*a*) and tracking tasks (Conrad *et al.*, 2011*b*). Sensory stimulation has also been used in the lower extremity to improve standing and walking function. Increased plantar sensory feedback, through the use of a textured insole, improves standing balance in neurologically intact individuals when visual feedback is removed (Corbin et al., 2007). Additionally, sub-sensory threshold vibration of the plantar surface of the foot improves standing balance control in stroke participants, with the largest improvements observed in participants with the greatest balance impairments (Priplata et al., 2006). Foot sole vibration also improves walking function in Parkinson's patients when applied during stance (Novak & Novak, 2006). Delivering electrical stimulation to the paretic foot and ankle during movement improves both walking speed and standing balance in chronic stroke survivors (Tyson *et al.*, 2013). These studies demonstrate that augmented sensory feedback, through various techniques, can improve the control of upper and lower extremity movements. In this study, we used electrical stimulation to augment sensory feedback from the paretic foot, which might be useful for improving foot placement control post-stroke.

The purpose of this study was to quantify the effects of sensory stimulation, provided by an electrical stimulus applied to the paretic foot, on foot placement during a stepping task. We hypothesized that electrical stimulation of the paretic foot, would decrease foot targeting error and improve lower extremity kinematics.

2.2 METHODS

2.2.1 Participant Information

Twelve chronic (> 6 months) stroke participants (age 47 – 63) with unilateral brain injury participated in this study. All twelve participants reported a vascular origin of their injury. Stroke information is included in Table 2-1. Exclusion criteria included inability to obtain informed consent, diagnosis of other neurologic disorders or cognitive deficits, recent (< 3 months) use of botulinum toxin, and inability to walk independently (with or without the use of an assistive device). A licensed physical therapist conducted a clinical evaluation of each individual consisting of the lower extremity Fugl-Meyer Test (Fugl-Meyer *et al.*, 1975), Berg Balance Assessment (Berg *et al.*, 1992), and 10 meter walking test (Mudge & Stott, 2009). Participant characteristics are summarized in Table 2-1. All procedures were approved by the Institutional Review Board at Marquette University, and all participants provided written informed consent.

Participant	Age (yrs)	Sex	Time Since Stroke (months)	Affected Side	Stroke Location	LE FM[1]	Berg[2]	Self-Selected Walking speed (m/s)	Paretic limb Monofilament Threshold[3]
S101	48	M	77	R	cort	22	51	1.173	3.84
S102	61	M	101	L	subcort	21	39	0.502	5.07
S103	60	F	61	R	cort	32	49	1.270	5.07
S104	63	F	236	L	n/a	32	55	1.298	3.61
S105	49	M	26	L	cort	24	45	0.743	5.18
S106	58	M	55	R	subcort*	29	46	1.361	5.88
S107	53	M	72	L	cort	24	49	0.988	4.56
S108	54	F	90	R	subcort	31	56	1.271	3.61
S109	64	M	30	R	subcort	28	52	1.043	4.31
S110	55	M	194	R	cort	32	54	1.576	4.31
S111	61	F	60	L	cort^	19	46	0.837	4.17
S112	58	F	285	L	cort	23	38	0.626	3.22

Table 2-1: Participant Characteristics: Time post injury (TPI). Lower extremity Fugl-Meyer (LE FM) maximum score 34. Berg Balance Test (Berg) maximum score 56. Self-selected overground walking speed (Ten Meter). Paretic limb monofilament sensory threshold, Normal ≤ 3.61, Loss of Protective Sensation ≥ 5.07. *carotid stroke; all others middle cerebral artery. ^hemorrhagic stroke; all others ischemic.

2.2.2 Data Collection

Kinematic data from the lower extremities were collected using a six camera Vicon Mx motion capture system (Vicon Motion Systems Ltd, Oxford, UK). Fifteen passive infrared reflective markers were placed at anatomical locations according to the Plug-In-Gait model (Davis *et al.*, 1991). All signals were collected using the Vicon Nexus software at 100Hz.

2.2.3 Experimental Protocol

Participants were placed in a ceiling-mounted fall arrest system. Participants started from a standing position, aligning both feet with two lines projected onto the floor to keep the starting location consistent across trials. One line aided in aligning the paretic foot in the medial lateral direction, while the other line aided in positioning both feet in the anterior posterior direction. Participants initiated each trial with the non-paretic limb, stepped to the projected target with the paretic limb, and then completed one more step each with the non-paretic and paretic limb. This sequence produced one complete, goal directed stride for each limb. During each trial, a circular target (r = 20mm) was projected onto the floor 500ms after a buzzer sounded indicating the start of the trial. Target locations were normalized to a percentage of the participant's paretic limb stride length, determined at the beginning of the session. Close, normal, and far targets were located in line with the paretic limb at a distance of 80%, 100%, and 120% of the paretic limb stride length, respectively. Two additionl targets were located 20% of the paretic stride length medial or lateral to the paretic limb starting location, at an anterior-posterior distance equal to the paretic stride length (Figure 2-1). Participants performed one practice trial to each target location to ensure they could complete the stepping sequence, and to reduce possible practice effects.

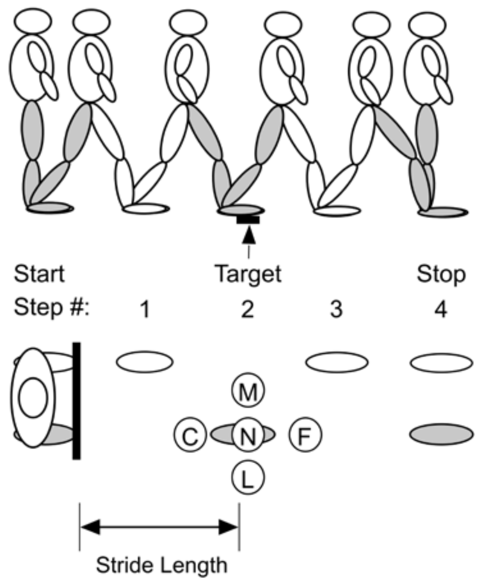

Figure 2-1: Diagram of Targeted Stepping Task. Participant started from rest, initiated movement with the non-paretic limb, stepping to the projected target with the paretic limb, finishing the sequence stepping the non-paretic then paretic limbs. Steps one and three were completed with the non-paretic limb, while steps two and four were completed with the paretic limb. Top view of experiment depicting target locations, a single target location was projected for each trial. Shaded limb/foot represents the paretic limb.

The testing was conducted in 3 blocks. During each block, targets were presented in a randomized order, and each target location was repeated 4 times, resulting in 20 trials in each experimental block. During the second of the three blocks, a 30 Hz electrical stimulation was applied to the medial plantar nerve of the paretic limb, providing evaluation of stepping before, during, and after stimulation. The stimulation began one second before target projection and remained on for the duration of the trial (6 s). A constant current stimulator (DigitimerDS7A, Digitimer Ltd, Hertfordshire, England) delivered biphasic pulses to two surface electrodes (Vermed Inc, Bellows Falls, VT) placed posterior to the medial malleolus on the paretic foot. Stimulation intensity was set to 95% of motor threshold of the abductor hallucis. This intensity produced a tactile sensation on the plantar surface of the foot, without producing a palpable contraction in the foot. The final, third, experiment block was conducted without stimulation to evaluate any potential aftereffects from the stimulation. A custom LabVIEW (National Instruments, Austin, TX) program was used to control timing of the Vicon data collection, target presentation, and electrical stimulation.

2.2.4 Data Analysis

Processing of the marker trajectories was completed using the Plug-In-Gait model in Vicon Nexus to obtain lower extremity kinematics and kinetics. Further data analysis was completed in Matlab (Mathworks, Natick, MA). Marker trajectories were low pass filtered at 15Hz prior to analysis. The analysis produced joint angles for each joint in three planes (sagittal, frontal, and transverse), foot placement locations, stance and swing timing, and stride and step lengths. Initially, stepping performance was assessed by the error magnitude between the projected target location and the toe marker location during paretic limb stance. Targeting error measures were calculated separately for the anterior-posterior and medial-lateral directions. Hip frontal plane motion during swing was quantified further by integrating the paretic limb frontal plane angle while the limb was in abduction during swing. The area of the frontal plane hip angle provided a measure of limb circumduction during swing, and was sensitive to changes in both the magnitude and duration of abduction. A measure of swing time symmetry was obtained by dividing the paretic by the non-paretic swing duration. A value of one indicated perfect swing time symmetry between the two limbs, and a value greater than one indicated that the paretic limb spent more time in swing compared to the non-paretic limb.

Separate univariate ANOVAs were completed to assess the effect of the electrical stimulation on error magnitude and frontal plane hip motion. Bonferroni post-hoc tests were used to examine differences between pre-stimulation, stimulation, and post-stimulation blocks. Pearson correlation analyses were completed to examine the relationships between the changes in hip frontal plane motion, lower extremity Fugl-Meyer, Berg Balance score, self-selected walking speed, and swing time symmetry. A correlation analysis between targeting error and trial number was performed for each participant to test for the presence of learning effects in the pre-stimulation block. All statistical tests were conducted with a significance level of $\alpha = 0.5$, and were completed using SPSS 16.0 software (IBM, Endicott, NY).

2.3 RESULTS

2.3.1 Targeting Error

Foot placement locations for all steps completed to the far target location for a single participant are shown in Figure 2-2. In general, participants reduced the distance between their foot placement and target location when somatosensory stimulation was applied. These changes in the control of foot placement due to electrical stimulation were quantified by the targeting error magnitude in both the medial-lateral and anterior-posterior directions (Figure 2-3). A significant main effect of stimulation condition ($p=0.008$) was observed across all targets for targeting error in the medial-lateral direction, while no significant effect was observed in the anterior-posterior direction. Post-hoc analyses indicated that medial-lateral targeting error was significantly greater in the pre-stimulation block compared to the stimulation ($p=0.006$) and post-stimulation blocks ($p=0.035$), as shown in Figure 2-3A. No significant correlations between targeting error and trial number in the pre stimulation block were observed for any of the 12 participants, indicating that the decrease in targeting error was not due to a learning effect.

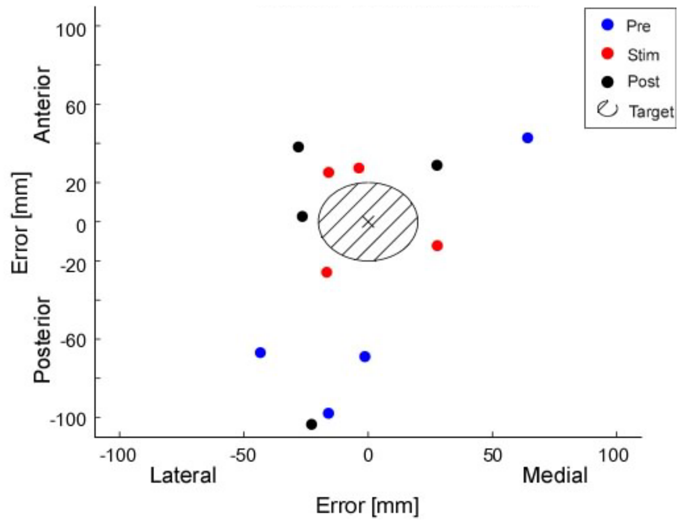

Figure 2-2: Single Subject Foot Placement for Far Target. Foot placement locations for S105 when stepping to far target, located at 120% of the paretic limb stride length. Somatosensory stimulation reduced medial-lateral distance from projected target (shaded circle) during stimulation and post-stimulation blocks.

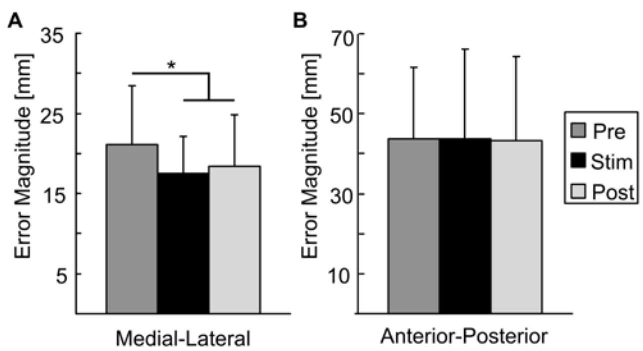

Figure 2-3: Targeting Error. Group average (+std) targeting error magnitude in medial-lateral (A) and anterior-posterior (B) directions across all targets. Medial-lateral targeting error was significantly reduced during the stimulation and post stimulation trials (Bonferroni post-hoc, $p < 0.05$).

2.3.2 Joint Kinematics

In addition to reductions in medial-lateral targeting error, 7/12 participants displayed decreases in magnitude and duration of hip abduction during swing (mean trajectories for participants S04 and S05 are shown in Figure 2-4). These seven participants demonstrated sustained hip abduction through late swing during the pre-stimulation block (Figure 2-4A) that was not present in the other five participants (Figure 2-4B). The presence of increased hip abduction during late swing is indicative of a hip circumduction compensatory strategy (Kerrigan *et al.*, 2000). When sensory stimulation was applied to the paretic limb, we observed decreases in this circumduction pattern that remained in the post-stimulation trials (Figure 2-4A).

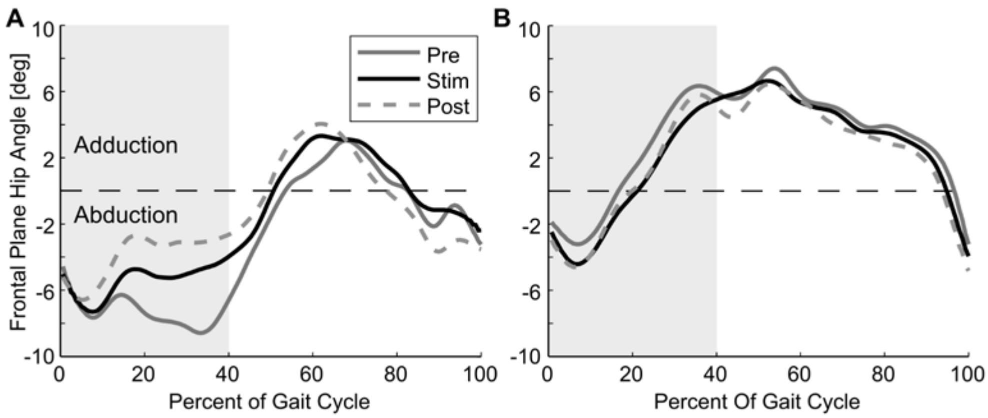

Figure 2-4: Subject Hip Frontal Plane Angle During Targeted Step. Frontal plane hip motion of the paretic limb from two representative participants when stepping to the normal target location (A: S105, B: S104). Shaded region represents swing phase. The somatosensory stimulation reduced the amplitude and duration of hip abduction during late swing for individuals presenting with a circumduction movement pattern (A), but had no effect on hip abduction for the non-circumduction group (B).

To evaluate the differential effects of stimulation on frontal plane hip motion, we correlated changes in frontal plane hip area from the pre-stimulation to stimulation block with clinical and functional measures. This change in hip abduction area significantly correlated with lower extremity Fugl-Meyer score ($r=0.752$, $p=0.005$), self-selected walking velocity ($r=0.609$, $p=0.024$), and swing time asymmetry ($r=-0.702$, $p=0.011$) (Figure 2-5). Reductions in hip abduction area during swing were observed in individuals with lower Fugl-Meyer scores (< 29) and slower self-selected walking speeds (< 1.2 m/s). These seven participants also presented with hip circumduction movement patterns during the pre-stimulation block, which were not observed in the other five participants. These seven individuals (circumduction group) showed a significant effect of stimulation condition ($p = 0.008$), and post-hoc analyses indicated that there was a significant decrease in the stimulation and post-stimulation blocks compared to the pre-stimulation block ($p < 0.001$) (Figure 2-6). There were no significant effects of stimulation condition for the non-circumducting group ($n = 5$).

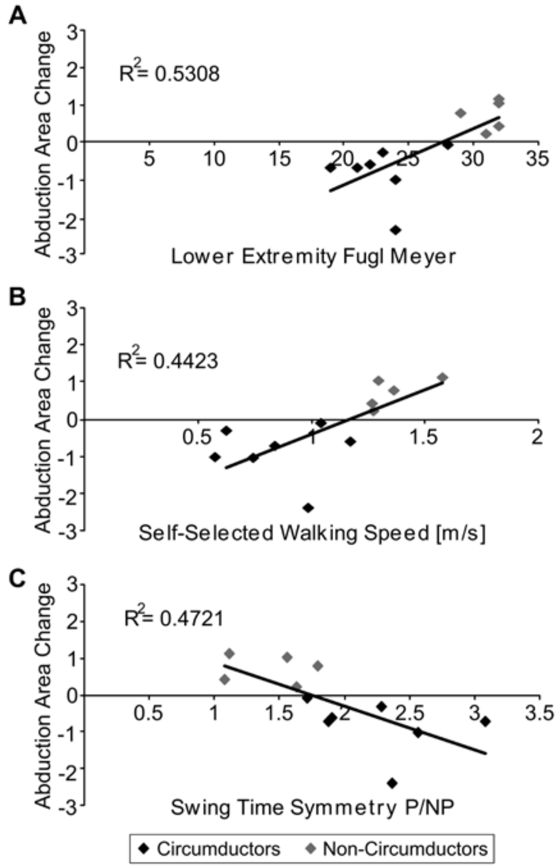

Figure 2-5: **Change in Abduction Area Correlations.** Correlation of average change in abduction area from stimulation to pre-stimulation block with lower extremity Fugl-Meyer (A), self-selected walking speed (B), and swing time symmetry ratio (B). A negative value represents a decrease in circumduction when stimulation was applied. The change in area significantly correlated with all three metrics, with reductions in circumduction area observed in patients with lower Fugl-Meyer scores, slower walking speeds, and more swing time asymmetry.

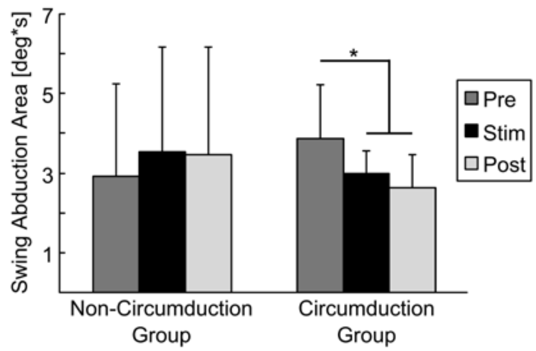

Figure 2-6: Effect of Stimulation on Abduction Area in Circumducting and Non-circumducting Groups. Average hip abduction area during swing for the two participant groups: those presenting with hip circumduction movement pattern (n=7), and those without hip circumduction movement pattern (n=5). Swing abduction area significantly decreased in both the stimulation and post-stimulation block compared to the pre-stimulation trials only for the circumduction group.

2.4 DISCUSSION

Application of somatosensory, electrical stimulation to the paretic foot produced improvements in frontal plane control of the paretic leg during a targeted stepping task. Specifically, we observed significant reductions in medial-lateral targeting error during the stimulation and post-stimulation blocks (Figure 2-3), suggesting improvement in the control of foot placement post-stroke. Somatosensory stimulation of the paretic limb also reduced hip abduction area during swing for participants presenting with a circumduction walking pattern (7/12), suggesting changes in frontal plane limb control. These results indicate that somatosensory stimulation might provide a mechanism to improve walking function post-stroke, especially in more impaired individuals.

The observation of locomotor changes in the frontal plane may be attributed to the manner in which supraspinal structures actively control walking. During walking, leg movement is inherently stable in the sagittal plane, and therefore supraspinal resources are likely focused upon control of frontal plane motion to optimally ensure balance and stability while walking (O'Connor & Kuo, 2009). Similarly, somatosensory, electrical stimulation applied to the paretic wrist improves hand function by inducing changes at the cortical level (Kaelin-Lang et al., 2002). It is plausible that our somatosensory stimulation paradigm activated a similar cortical mechanism, despite being applied to the lower extremity. Somatosensory stimulation of the paretic foot may be acting to enhance sensorimotor integration in areas such as the posterior parietal cortex, which are important to the execution of visually guided locomotor movements (Marigold et al., 2011). Further research is needed to understand the potential mechanisms behind these improvements in locomotor control in order to maximize its effect for stroke survivors.

The observed improvements in paretic leg control might also be associated with stimulation-induced changes in hip and knee synergy patterns that reduce circumduction. After stroke, increased multi-joint coupling between the paretic hip and knee (Lewek *et al.*, 2007) contributes to both reduced gait speeds as well as increased pelvic compensatory movements (Cruz *et al.*, 2009). The persistence of abnormal hip abduction movements during robot-assisted gait (Neckel *et al.*, 2008; Sulzer *et al.*, 2010) suggests that measures must be taken to reduce this coupling in order to restore normal kinematic patterns. The observed decreases in hip abduction area during swing in this study may represent changes in functional coupling of the hip and knee muscles due to the somatosensory stimulation. This reduced frontal plane hip movement could contribute to observed reductions in targeting error by enabling participants to take a more direct path to the target location. However, we did not observe any significant correlations between hip abduction area and frontal plane targeting error. Since hip circumduction only reduced in the circumducting group, but both groups showed improvements in foot placement control, we do not attribute reduced targeting error solely to reductions in hip circumduction. Improved frontal plane biomechanics, especially in more impaired stroke survivors, and improved locomotor planning likely act together to improve foot placement control during the task.

It is important to note that the targeted stepping task used in this study is analogous to, albeit different from continuous walking. In our task, participants initiated gait with the non-paretic limb, stepped to a projected target with the paretic limb, and finishing with a series of two more steps. This design ensured that participants completed this goal directed movement within the context of a walking task. Unlike previous studies that have evaluated foot placement during obstacle avoidance (Said et al., 2001) or targeted foot placement during walking (Alexander *et al.*, 2011), which allowed for modification of the walking pattern over a series of steps, we wanted to evaluate the ability of stroke survivors to execute a targeted movement within a single gait cycle of the paretic limb. However, due to the fact that participants started this task from rest, larger demands were placed on the paretic limb to generate forward momentum to initiate walking (Hesse et al., 1997), which has been shown to have reduced propulsive output post-stroke (Bowden *et al.*, 2006). The increased propulsive demands placed on the paretic limb during the step to the target, relative to normal walking, may result in larger improvements than those expected during continuous walking. We were unable to obtain sufficient ground reaction forces in this experiment to quantify paretic limb propulsion during the baseline task performance, or the influence of the somatosensory stimulation on paretic propulsion. However, it is unlikely that the improvements in targeting error were only due to changes in paretic propulsion, since deficits in the frontal plane control of foot placement were also observed during medial step corrections made with the paretic limb (Nonnekes et al., 2010). Another potential limitation is the goal directed nature of the targeting stepping task, which has been shown to produce higher firing rates in the motor cortex compared to normal locomotion in cats (Beloozerova *et al.*, 2010). Therefore, it is unknown how these improvements in frontal plane foot placement transfer to continuous walking with somatosensory stimulation. However, it is likely that the greatest benefits will be observed when continual adjustments are needed during walking, such as walking over an uneven surface or through a cluttered environment.

The results of this study demonstrate the potential for including somatosensory stimulation of the paretic foot into traditional rehabilitation techniques to further improve walking function in stroke survivors. Stroke survivors possess the ability to produce symmetric walking patterns (Reisman *et al.*, 2009), but the prevalence of asymmetries in the walking pattern post-stroke suggests a significant contribution of abnormal control mechanisms. Applying somatosensory stimulation to the paretic foot during the walking task improved the precision of paretic foot placement, as well as reducing hip circumduction in more impaired individuals. Furthermore, these reductions in hip abduction correlated with both clinical and functional metrics, suggesting that somatosensory stimulation will likely have the largest effect in individuals with the most impaired walking function. Similarly, a ceiling effect was observed when somatosensory stimulation was applied to the paretic wrist (Kaelin-Lang et al., 2002), supporting the use of somatosensory stimulation with more impaired patients. Additionally, these improvements in frontal plane control remained when the stimulation was removed, suggesting at least a short-term (20 stepping trials) change in locomotor control (i.e. aftereffects). Further research is needed to determine the duration of these plastic changes in stepping function, as well as to identify the impact of somatosensory stimulation of the paretic foot on continuous walking.

CHAPTER 3: DYNAMIC BALANCE CONTROL STRATEGIES IN STROKE SURVIVORS

3.1 INTRODUCTION

The purpose of this study was to compare dynamic balance control strategies of stroke survivors and age-matched controls under challenging balance conditions during treadmill walking. After a stroke, standing balance control is a strong predictor of walking function (Michael et al., 2005). This association likely reflects a change in dynamic balance control, which we define as balance during walking. Dynamic balance control is also likely to be impaired in stroke survivors; however, balance is rarely measured during walking despite the functional implications. Dynamic balance control is critical to function because errors in balance during walking can lead to falls, which have significant health effects in stroke survivors. Even after completion of a rehabilitation protocol, stroke survivors have a higher occurrence of falls (Jørgensen *et al.*, 2002), and many of these falls occur during walking (Mackintosh et al., 2005). Potential injuries sustained from a fall, as well as an increased fear of falling again (Watanabe, 2005), can further decrease the already impaired walking function post-stroke. This potential negative impact upon walking ability demonstrates the need to better characterize deficits in walking balance control post-stroke.

Deficits in the ability to accurately place the foot at a targeted location during walking is likely a key factor in dynamic balance control in stroke survivors. Frontal plane control of balance is challenging due to medial-lateral movement of the body's center of mass (COM), and the varying size and position of the base of support during gait. In contrast, the sagittal plane has an underlying dynamic stability, which results in balance control that is largely focused on the frontal plane (Bauby & Kuo, 2000). One mechanism to maintain frontal plane balance is to modify lateral foot placement location to keep the body's center of mass within the base of support (Hof, 2008). Difficulty in making medial-lateral step corrections (Nonnekes et al., 2010) may impair this control mechanism in stroke survivors. A potential compensatory mechanism for impaired foot placement control is to shift the body's COM further away from the paretic limb, and over the nonparetic limb, where medial-lateral corrections are available. In fact, this strategy has been observed as an asymmetric medial-lateral foot placement relative to the pelvis during walking in stroke survivors (Balasubramanian et al., 2010). Furthermore, when stepping over an obstacle, stroke survivors modify foot placement location to keep the COM closer to the stance limb to help with balance control in case the paretic limb contacts the obstacle (Said et al., 2001; 2008). Thus, impairments in the ability to place the paretic foot likely cause changes in the dynamic balance control strategy.

In addition to difficulties controlling foot placement, changes in COM movement likely impact balance control during walking in stroke survivors. During quiet standing, stroke survivors demonstrate increased levels of frontal plane COM movement compared to elderly controls, and this difference is enhanced when individuals stand with their eyes closed (Marigold & Eng, 2006b). Further increases in postural sway are also observed after stroke survivors complete an extended period of walking (Carver et al., 2011). Deficits in accurate trunk position sense (Ryerson et al., 2008) would also impact the accuracy of frontal plane foot placement during walking (Hof et al., 2007).

Consequently, stroke survivors might rely more strongly upon visual feedback to estimate body position. Providing visual feedback of center of pressure location during standing significantly reduced frontal plane sway in chronic stroke survivors, although sway was still larger in comparison to young and old controls (Dault *et al.*, 2003). Feedback of trunk position reduces sway in healthy young individuals (Verhoeff *et al.*, 2009) however, it is unknown whether visual feedback of body movement can improve dynamic balance control during walking in stroke survivors.

In this study we assessed walking performance during challenging walking conditions to gain further insight into dynamic balance control deficits post-stroke. Additionally, we evaluated the impact of providing a visual cue related to body motion on dynamic balance control in stroke survivors. We hypothesized that visual feedback of body movement would reduce frontal plane COM movement in chronic stroke survivors during walking, with the largest improvements when a stationary reference was provided.

3.2 METHODS

3.2.1 Participants

Ten chronic (> 6 month) stroke survivors with unilateral brain injury, and ten age and sex-match neurologically intact individuals participated in this study. Exclusion criteria for this study included recent use of botulinum toxin in the lower extremity, inability to walk independently (with or without use of an assistive device), lesion to brainstem centers, diagnosis of other neurologic disorders, or inability to provide informed consent. Prior to beginning the experimental session, a licensed physical therapist conducted a clinical evaluation of the stroke participants, consisting of the lower extremity Fugl-Meyer Test (Fugl-Meyer et al., 1975), Berg Balance Assessment (Berg et al., 1992), dynamic gait index (Jonsdottir & Cattaneo, 2007), and 10 meter walking test (Mudge & Stott, 2009). Only the 10 meter walking test was completed for control participants, to assess their comfortable overground walking speed. Participant characteristics are summarized in Table 3-1. The Marquette University Institutional Review Board approved all experimental procedures, and written informed consent was obtained from all individuals participating in this study.

ID	Sex	Age [yrs]	Time Post Stroke [months]	Affected Side	LE FM	Berg	DGI	Walking Speed [m/s]	Treadmill Speed [m/s]
S201	M	54	71	L	24	49	15	0.988	0.55
S202	F	62	317	L	19	46	21	0.837	0.36
S203	F	55	30	R	31	56	24	1.271	0.63
S204	M	54	42	L	30	43	17	1.136	0.48
S205	F	65	117	L	32	55	23	1.298	0.60
S206	F	62	144	R	32	49	21	1.270	0.58
S207	M	62	95	L	21	39	14	0.502	0.29
S208	M	59	120	R	29	46	21	1.361	0.75
S209	F	54	68	L	28	41	17	0.635	0.30
S210	M	65	7	R	27	54	19	0.995	0.65
C201	M	56	-	-	-	-	-	1.471	1.00
C202	F	62	-	-	-	-	-	1.212	0.96
C203	F	54	-	-	-	-	-	1.212	0.85
C204	M	57	-	-	-	-	-	1.515	0.90
C205	F	66	-	-	-	-	-	1.242	1.00
C206	F	61	-	-	-	-	-	1.299	0.75
C207	M	63	-	-	-	-	-	1.429	0.95
C208	M	58	-	-	-	-	-	1.333	0.90
C209	F	54	-	-	-	-	-	1.325	0.95
C210	M	63	-	-	-	-	-	0.980	0.84

Table 3-1: Participant Characteristics. Lower extremity Fugl-Meyer (LE FM) maximum 34, Berg Balance maximum 56, Dynamic Gait Index maximum 24.

3.2.2 Experimental Protocol

Walking trials were conducted on an instrumented split-belt treadmill (FIT, Bertec Inc, Colombus, OH) with both belts set to the same speed. Belt speed was determined at the beginning of the session during a familiarization period, during which the treadmill speed was slowly increased until participants self-selected the speed that felt most comfortable. This self-selected belt speed was used for all the subsequent walking trials, and is included in Table 3-1. Individuals were placed in a fall arrest harness, and held onto a side handrail with the non-paretic hand for safety. The handrail was instrumented with a six DOF load cell (MC3A-250, AMTI, Watertown, MA) to quantify handrail forces and torques throughout the trials. Control participants held onto the handle with the hand opposite the test leg, to keep handrail hold consistent between groups.

Walking performance was evaluated under testing conditions where visual information was altered to change sensory feedback signals during walking. Reduced visual feedback was achieved by having the individual wear a pair of goggles with black tape obstructing the lower half of the visual field. These goggles blocked the view of the participant's legs, while still providing some visual feedback of their location in the room. Additional visual feedback of body motion during walking was provided using a laser that was attached to a headband worn by the participants. The laser produced a visible circle (r = 0.01m) on the wall in front of the treadmill (3.8m), and the movement of the circle was related to the movement of the participant's head (and body) during walking. Trials were conducted with no altered visual information (normal walking), normal walking with the laser, reduced visual feedback, and reduced visual feedback with the laser. In these laser-walking trials, the laser was turned on and the participant was given no explicit instruction on what to do with the laser. After these trials were completed, two laser targeting trials were conducted. During these targeting trials, a projector mounted above the treadmill was used to display a target on the wall in front of the treadmill that either remained stationary or moved during the trial. The stationary targeting trial consisted of a large circular target (r= 0.22m) that the participant was instructed to keep the laser within, while walking. During the moving targeting condition, a smaller target (r= 0.06m) randomly moved through a 1.5 by 1.0m area on the wall in front of the participant, with the position changing every 1.0 to 2.0s. The center of the stationary target, and middle of the moving target area was approximately at the center of the participant's visual field when looking straight ahead. The order of these two targeting trials was randomized across participants.

Throughout all walking trials, walking performance was characterized over a period of 100 gait cycles with the paretic or test leg. Fifteen passive infrared reflective markers were placed at anatomical locations according to the Plug-In-Gait model (Davis et al., 1991), with an additional seven markers placed at the left and right shoulder, C7, and four markers placed on the head. A six camera Vicon motion capture system (Vicon Motion Systems Ltd, Oxford, UK) recorded marker location at 100Hz. Treadmill ground reaction forces, and handrail forces were collected at 1000Hz using a Vicon Mx Giganet to synchronize the analog and video data.

3.2.3 Data Analysis

The data were initially processed in Vicon Nexus software to label markers, visually indicate gait events, and run the lower extremity Plug-In-Gait model. Additional data analysis was completed in Matlab (Mathworks, Natick, MA). An eight-segment model consisting of the foot, shank, thigh, pelvis, and trunk was used to estimate whole body COM location (Winter, 2009). COM sway measured the extent of COM movement in the frontal plane over a gait cycle. Step width and foot placement location relative to the pelvis COM (Balasubramanian et al., 2010) at heel strike were calculated to characterize foot placement in the frontal plane. The ratio of step width to COM movement (SW/COM) was calculated as a metric to compare the size of the base of support to the extent of COM movement. Additionally, temporal and spatial gait parameters were calculated to characterize changes in walking performance during the different testing conditions.

Statistical analyses were conducted using SPSS 20.0 (IMB, Armonk, NY). Measures of walking performance were averaged across all gait cycles within each trial to obtain the participant's typical response to each experimental condition. A repeated measures ANOVA was conducted separately for each variable, to evaluate differences between both the experimental conditions and groups. A Greenhouse-Geisser correction was used to correct for non-spherical data when comparing within-subject effects. Post-hoc analyses were carried out for significant factors using a Sidak correction to account for multiple comparisons. A Pearson correlation analysis was carried out between the percent change in SW/COM ratio and the clinical tests, to understand how changes in dynamic balance control post-stroke related to standard clinical measures.

3.3 RESULTS

3.3.1 Balance Measures

Frontal plane movement of the COM and center of pressure (COP) over the first 20s of the baseline walking and stationary targeting trials are shown for a representative control, and two stroke participants in Figure 3-1. In general, stroke participants walked with a larger COM movement in the frontal plane (Group, $p=0.003$) and larger step widths (Group, $p=0.001$) compared to age and gender-matched neurologically intact individuals (Figure 3-2). Stroke survivors also placed their paretic foot more lateral to the COM at heel strike compared to controls (Group, $p<0.001$), but no difference was observed between groups for the non-paretic limb. Despite these baseline differences in step width and COM movement, stroke participants maintained a similar SW/COM ratio (Group, $p>0.958$).

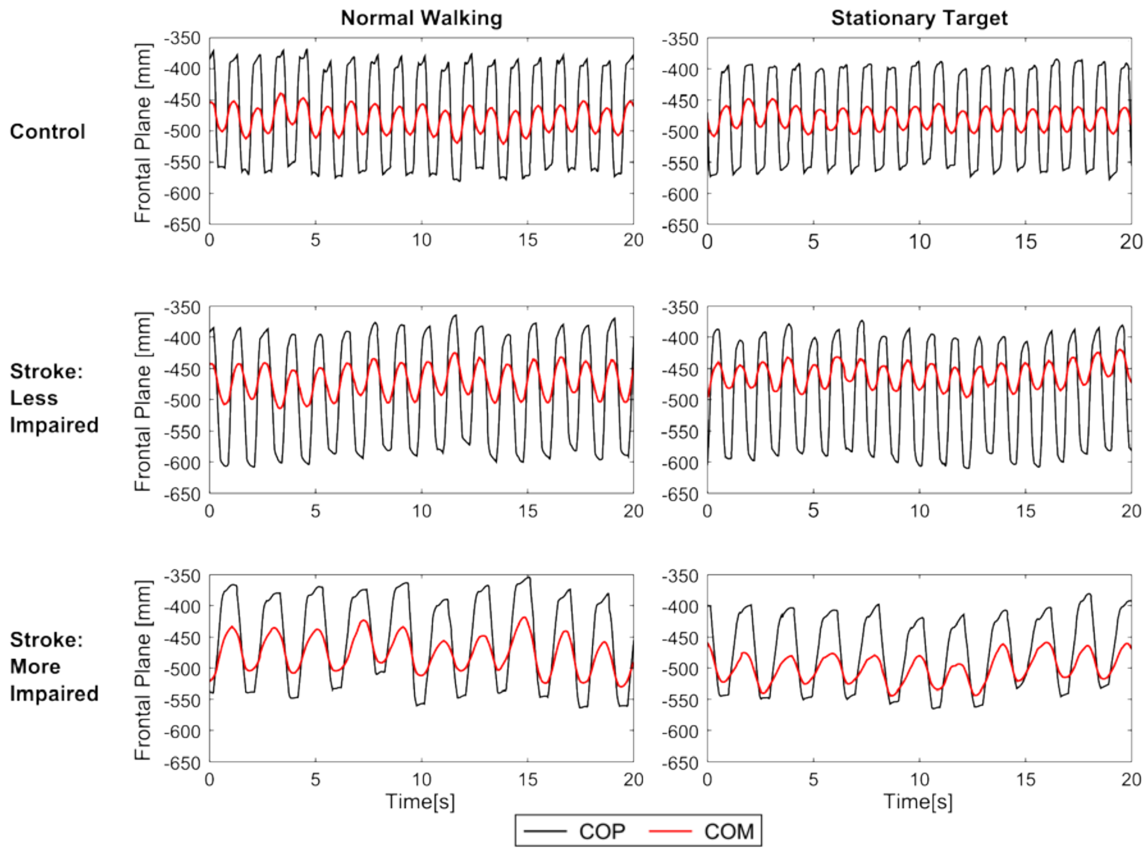

Figure 3-1: Example Participant Frontal Plane COM and COP Movement. COP and COM traces from first 20s of normal and stationary targeting trials from a representative control (C203), less impaired stroke (S205), and more impaired stroke (S207) participant. The less impaired stroke participant shows the greatest reduction in COM range of motion during the stationary targeting trial.

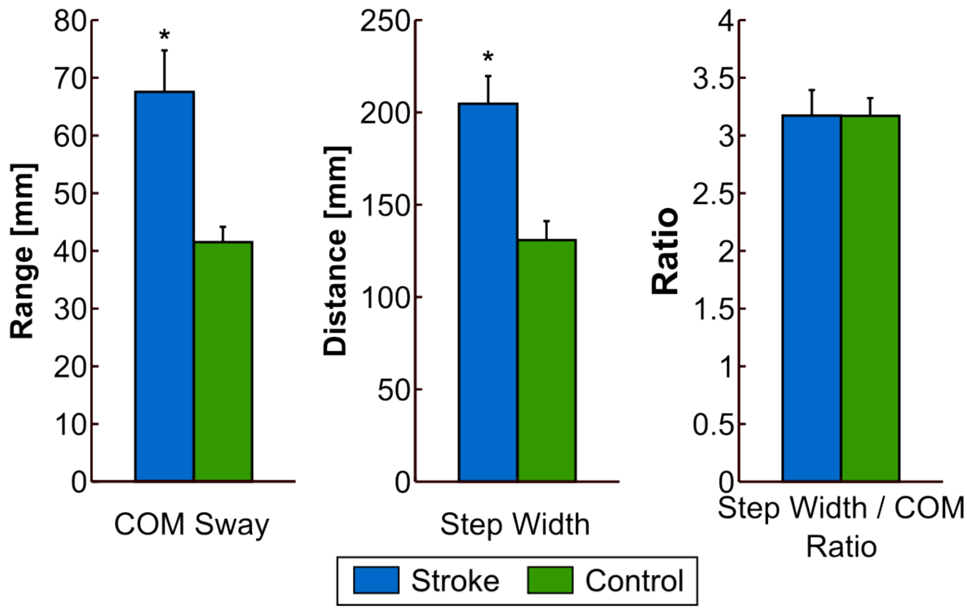

Figure 3-2: Group Differences In Measures Of Frontal Plane Balance Control. Stroke participants walked with larger amounts of frontal plane COM movement and step widths compared to controls across all testing conditions. The ratio of step width to COM movement was not different between groups. (* ANOVA, Group $p<0.05$)

The COM sway (Condition, $p<0.001$) and the SW/COM ratio (Condition, $p=0.002$) was statistically different between experimental conditions, but these experimental conditions did not impact step width ($p=0.243$) or frontal plane foot placement (paretic $p=0.371$, non-paretic $p=0.211$). Changes in COM sway were different between the stroke and control groups (Condition*Group, $p=0.034$) (Figure 3-3). The stationary targeting condition resulted in lower amounts of COM sway compared to both normal ($p=0.034$) and reduced visual feedback walking ($p=0.016$) trials without the laser. Additionally, adding the laser feedback to the normal walking and reduced visual feedback trials slightly reduced COM sway compared to the no laser trials, but these differences were not statistically significant for either the stroke ($p=0.227$) or control ($p=0.396$) group.

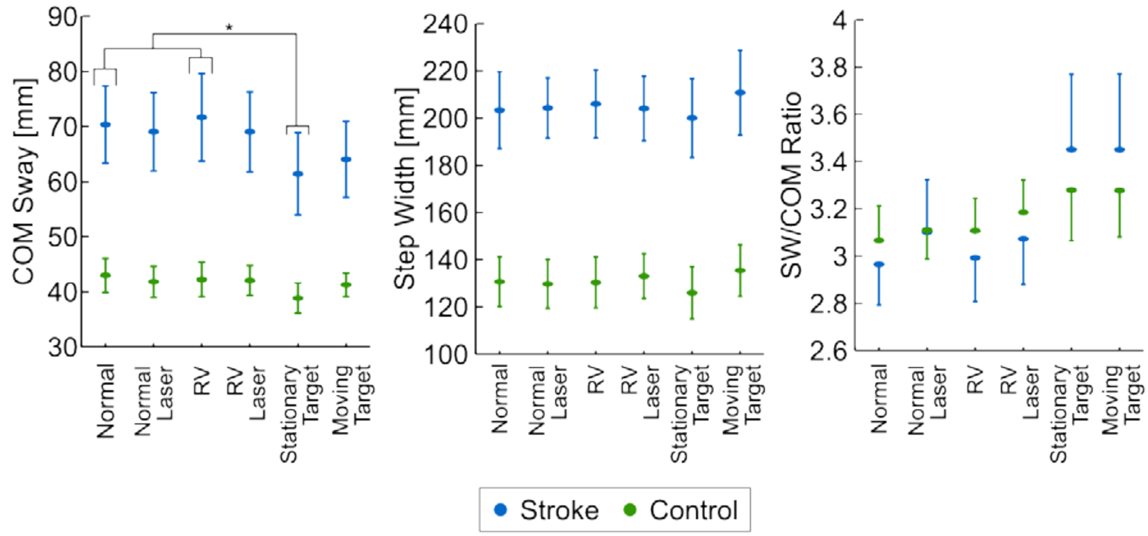

Figure 3-3: Effect of Testing Condition on COM Sway and Step Width. Group average (± standard error) frontal plane COM movement for each testing condition. Significant reductions in COM sway were observed in the stroke group for the stationary targeting condition compared to normal and reduced visual feedback (RV) trials without the laser.

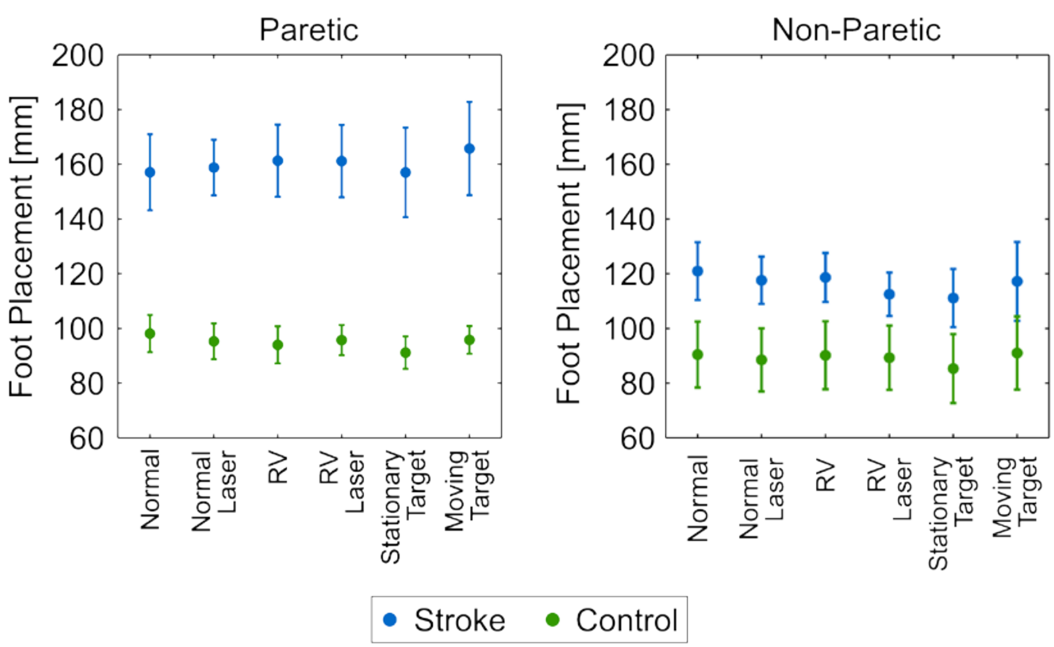

Figure 3-4: Frontal Plane Foot Placement Across Testing Conditions. Average (± standard error) frontal plane foot placement location relative to pelvis COM at heel strike for paretic and non-paretic limbs. Stroke participants placed the paretic foot more lateral to the pelvis than controls. The stroke group tended to maintain paretic limb foot placement location across all conditions, compared to reductions during the stationary targeting condition for the non-paretic, and both limbs in the control group.

The ratio of step width to COM sway (SW/COM) provided insight into the frontal plane balance strategy by relating the base of support relative to the COM range of movement across the gait cycle (Figure 3-2). This ratio increased as the walking tasks became more challenging (Condition, $p=0.002$), with the larger values observed during the stationary (post-hoc, $p=0.025$) and moving (post-hoc, $p=0.041$) targeting trials when compared to baseline walking. Larger ratios could indicate a more conservative balance strategy, with a larger base of support chosen for a given amount of COM movement. However no significant changes in step width (Figure 3-3) or frontal plane foot placement (Figure 3-4) were observed across testing conditions, indicating that changes in this ratio were mainly influenced by COM sway. The percent change in the ratio from baseline walking to the stationary targeting condition correlated with the lower extremity Fugl-Meyer score ($r=0.866$, $p=0.001$) and self-selected overground walking speeds ($r=0.652$, $p=0.041$) (Figure 3-5). As lower extremity Fugl-Meyer scores and walking speeds increased, individuals demonstrated larger percent increases in this ratio.

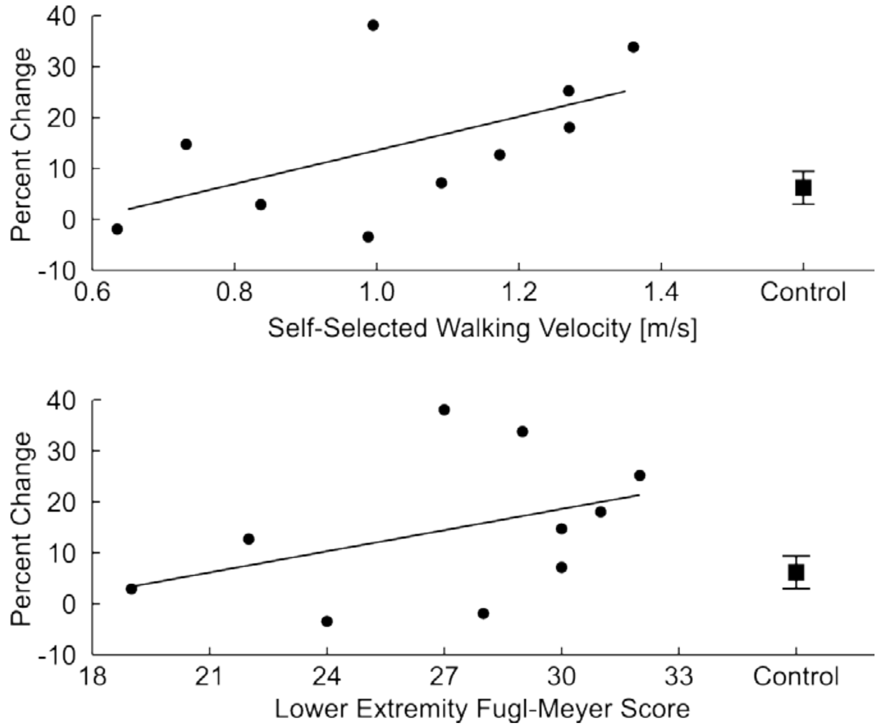

Figure 3-5: Stationary Targeting Percent Change in Ratio Of Step Width to COM Sway Correlates with Clinical Measures. The percent change in the ratio of step width to COM sway during the stationary targeting condition correlated with self-selected walking velocity and lower extremity Fugl-Meyer score.

3.3.2 Temporal Parameters

The stroke group had longer gait cycle durations for both the paretic and non-paretic legs (Group, p=0.014) compared to controls, due to their slower treadmill walking speeds. Both groups decreased cycle time for both legs during the moving targeting trial compared to normal walking with (p=0.005) and without (p=0.014) the laser, reduced visual feedback without the laser (p=0.015), and stationary targeting (p=0.005) trials. These results were strongly driven by changes in the stroke group, but no significant interaction effect of group was observed (Group*Condition, p=0.101).

The changes in cycle time were also accompanied by associated changes in cadence between the testing conditions (Condition, p=0.003). The moving targeting trial increased cadence compared to normal walking with the laser ($p=0.003$) and the reduced visual feedback without the laser ($p=0.003$). Changes in cadence were largely driven by the stroke group (Group*Condition, $p=0.096$), which displayed increases in cadence as the difficulty of the task increased.

3.3.3 Spatial Parameters

Step lengths were shorter for both the paretic (Group, $p=0.002$) and non-paretic (Group, $p<0.001$) limbs of the stroke group compared to the control group. In general, as the difficulty of the walking task increased, step lengths tended to decrease for both the paretic (Condition, $p=0.035$) and non-paretic limbs (Condition, $p=0.001$). Significant reductions in step length were observed during the moving targeting condition relative to the other conditions (post-hoc, $p<0.05$) for the non-paretic and non-test limbs only. These reductions appear to be largely driven by the stroke group (Group*Condition, $p=0.134$), which demonstrated a larger percent reduction (10.8%) compared to controls (2.6%).

3.5 DISCUSSION

The results of this study demonstrate that visual feedback during walking is an important aspect of dynamic balance control post-stroke. Feedback of body movement impacted COM movement during walking, but only for stroke survivors. Furthermore, this effect was task specific, and required the presence of a stationary target to produce significant decreases in COM sway. This reduction in COM sway increased the SW/COM ratio, with the percent change correlating with clinical measures of walking speed and sensorimotor recovery. Additionally, although stroke survivors walked with greater movement of the COM and larger step widths, the ratio between these measures was similar between groups. These results support our initial hypothesis that providing visual feedback of trunk movement can help stroke survivors reduce COM sway.

The additional sensory feedback supplied by the head mounted laser provides a potential mechanism to improve dynamic balance control post-stroke. This visual cue likely has a larger impact in the stroke group due to an increased reliance upon visual feedback for balance control post-stroke (Marigold & Eng, 2006b). The laser provided feedback of body movement during walking, which might be used to compensate for impaired sense of trunk position (Ryerson et al., 2008). Providing additional feedback of trunk movement, through multiple sensory modalities, has been shown to reduce sway during both standing (Huffman *et al.*, 2010) and walking (Verhoeff et al., 2009) in young adults. In our study, the control group showed a trend towards decreased COM sway during the stationary targeting task, but the lack of a significant reduction in sway suggests that neurologically intact participants were less reliant on visual feedback for dynamic balance control compared to chronic stroke survivors. It is important to note that the effectiveness of this feedback signal is dependent upon context of the task. Simply turning on the laser during walking, or providing a moving target, did not provided the appropriate context for the visual cue to have a significant impact upon COM sway.

Analysis of changes in the SW/COM ratio provided insight into the overall balance control strategy in response to the different experimental conditions. Both groups demonstrated the general trend of increasing this ratio as the walking tasks became more challenging, which likely represents the selection of a more conservative walking pattern to reduce fall risk. However, we did notice that the stroke group had a larger percent increase in the SW/COM ratio from baseline walking to the stationary targeting condition, with this percent change positively correlated with the lower extremity Fugl-Meyer score and self-selected overground walking speed. Larger percent changes were observed for individuals with faster walking speeds, and greater levels of sensorimotor recovery. Further examination revealed differences in how stroke survivors achieved these changes in the SW/COM ratio during the stationary targeting task. Higher functioning participants made larger reductions in COM sway, compared to lower functioning participants. The lack of these COM sway changes in the more impaired participants suggests an inability to adapt COM movement to the task demands, which may also explain increased fall incidence. Additionally, despite changes in frontal plane movement of the COM, stroke subjects did not make significant adjustments to lateral foot placement, suggesting deficits in lateral foot placement control. This reduced control may bias stroke subjects towards the selection of a more conservative dynamic balance strategy to reduce the risk of falls.

Interestingly, despite baseline differences in step width and COM sway, the ratio of these parameters is preserved after a stroke. Step width and frontal plane COM movement are strongly associated in both the biomechanics of walking, and in balance control strategy, making it difficult to determine which parameter is driving the observed baseline differences. Increased COM sway could be due to deficits in the control COM movement (Marigold & Eng, 2006a), or due to slower walking speeds post-stroke (Orendurff et al., 2004). However, it is unlikely that slower walking speeds are the sole source of increased COM sway post-stroke, since larger step widths are observed when walking speeds are matched between groups (Chen *et al.*, 2005*b*). Increased step width, and a greater separation between the COM and paretic foot, would help to minimize the contribution of the paretic limb in maintaining balance. Wider step widths also reduce the muscle activity needed to redirect COM movement (Henry *et al.*, 2001), but the neural feedback gains must be adjusted to maintain stability (Bingham et al., 2011). Increased muscle activation latencies in the paretic limb (Kirker et al., 2000) potentially limit the ability of the underlying neural control to maintain stability at these wider step widths, which could explain the increased incidence of falls despite a wider step width post-stroke.

Taken together, these results provide further insight into walking balance control strategy post-stroke. Chronic stroke survivors maintain a similar ratio between COM movement and step width, but walk with greater baseline levels of both variables compared to neurologically-intact individuals. Visual feedback of body movement coupled with a stationary reference point improved frontal plane COM control during walking. However, stroke survivors did not alter step width or lateral foot placement of the paretic limb when the additional feedback was provided. Further research into the dynamic control of foot placement during walking is needed to fully understand changes in walking balance control post-stroke.

CHAPTER 4: LOCOMOTOR ADAPTATIONS TO FRONTAL PLANE TRUNK PERTURBATIONS IN YOUNG ADULTS

4.1 INTRODUCTION

Adaptations to dynamic balance control are an important component of adjusting to novel ambulation environments. Maintaining dynamic balance during walking is a challenging control task for the central nervous system due to the bipedal nature of human locomotion. In general, balance is achieved by maintaining the body's center of mass (COM) within the base of support; however, the height of the body's COM from the ground, and the constantly changing base of support complicate this task (Winter, 1995). Previous modeling and human experimentation has demonstrated that human locomotion is passively stable in the sagittal plane, suggesting that active balance control strategies focus on balance in the unstable fontal plane (Kuo, 1999; Bauby & Kuo, 2000). Perturbations of visual feedback (O'Connor & Kuo, 2009), and oscillation of the support surface (Mcandrew *et al.*, 2010), during treadmill walking support this theory, showing that neurologically intact individuals are more sensitive to perturbations in the frontal as opposed to sagittal plane. The goal of this study was to apply continuous, frontal plane perturbations to the trunk to identify how individuals adjust their walking cycle to maintain dynamic balance.

The control of movement is focused upon adjusting the motor plan to meet the specific demands of the task and the environment. These adjustments usually occur very rapidly for skilled movements that are repeatedly preformed, such as reaching or walking, making it difficult to gain insight into the underlying motor control strategy. Performance of these skilled movements within a novel environment is a valuable technique, which can be used to gain deeper insight into how these movements are controlled. External force fields have been used to in reaching to characterize how the individual adapts their reaching pattern to account for the forces in order to follow a desired movement trajectory (Shadmehr & Mussa-Ivaldi, 1994). Novel task environments have also been used to evaluate locomotor control. A rotating support surface has been demonstrated to produce a podokinetic afterrotation, in which individuals, when blindfolded, produce a curved overground walking trajectory (Gordon *et al.*, 1995). Further analysis of these aftereffects have provided additional insight into the role of the vestibular system in locomotor control (Earhart & Hong, 2006), as well as support for a single neural center responsible for locomotor trajectory control (Mcneely & Earhart, 2010). Split-belt treadmill adaptation studies, where each limb is moving at a different speed, have provided further insight into locomotor control. Different adaptation rates for intralimb and interlimb locomotor parameters to split-belt walking suggest that separate neural networks are responsible for the control of these parameters during walking (Reisman *et al.*, 2005). Additionally, altering the level of attention to the adaptation task, affected adaptation rates of spatial but not temporal parameters, suggesting spatial parameters may be controlled by more cortical centers during walking (Malone & Bastian, 2010). Based upon the information obtained from these adaptation studies, we postulate that altering the dynamics of trunk motion during walking will provide a mechanism to evaluate dynamic balance control strategy.

Lateral trunk perturbations have been previously used to characterize the utilization of a lateral foot placement strategy to maintain balance during walking (Hof et al., 2010). However, the perturbations used by Hoff et. al. were of a short duration, and only characterized the initial corrective response to the perturbation. In this study, we created a novel dynamic balance environment using continuous, cyclical force perturbations to the trunk while study participants stepped on a split belt treadmill. Walking trials were conducted in a block design, with right-left trunk forces continuously delivered throughout the middle block using a cable-driven system. Adaptation to this force field was measured using catch trials. The effects of perturbation type (augmenting vs. resisting), perturbation magnitude, and handrail hold on the locomotor adaptations were also evaluated. We hypothesized that individuals would increase step width in response to augmenting perturbations, and would decrease step width for resisting perturbations. Furthermore, we anticipate that these adaptations will rapidly occur to prevent a loss of balance.

4.3 METHODS

Ten individuals (5 male, 5 female, ages 21-30) with no reported neurological injury or disease participated in this study. The Marquette University Institutional Review Board approved all experimental procedures, and written informed consent was obtained from all individuals prior to participating in this study.

Fifteen passive infrared reflective markers were placed at anatomical locations according to the Plug-In-Gait model (Davis et al., 1991) to capture lower extremity movement. Additionally, markers were placed bilaterally on the wrist, elbow, shoulder, front and back head, and on the C7 vertebra to quantify movements of the upper extremity and head. Marker locations were recorded at 100Hz using an eight camera Vicon motion capture system (Vicon Motion Systems Ltd, Oxford, UK). Ground reaction forces were recorded from an instrumented, split-belt treadmill (FIT, Bertec, Colombus, OH). A custom adjustable handle, instrumented with a six degree of freedom load cell (AMTI, MC3A-250, Watertown, MA), was attached to a front handrail of the treadmill to quantify handrail hold forces. Handle forces were amplified at 1,000 V/V, and low pass filtered at 500 Hz prior to collection (Gen5, AMTI Inc., Watertown, MA). Perturbation forces were measured using a load cell (MLP-300, Transducer Techniques Inc., Temecula, CA) attached in line with the cable. Signals were amplified at 450V/V and lowpass filtered at 250Hz prior to collection (TMO-1-24, Transducer Techniques Inc., Temecula, CA). Ground reaction forces, handle forces, and cable perturbation forces were all sampled at 1000Hz using a Vicon Mx Giganet, which synchronized the analog and video data.

4.3.1 Cable-driven Perturbations

A novel cable-driven device (similar to Wu *et al.*, 2011) was constructed to deliver medial-lateral perturbations to the trunk during treadmill walking. The cable-driven device consisted of a servomotor system (AKM-33H, AKD-0606, Kollmorgen, Radford, VA) that drove an aluminum spool with a light stainless steel cable attached (Figure 4-1). The system was capable of delivering pulls up to 100N, and a device was placed on the left and right side of the treadmill to deliver both left and right perturbations. Each cable ran through a pulley and attached to the belt of the fall arrest harness worn by the individual, with the harness and pulley height adjusted to have the cable connections near the top of the pelvis. This location enabled us to deliver external perturbations near the approximate location of the participant's center of mass.

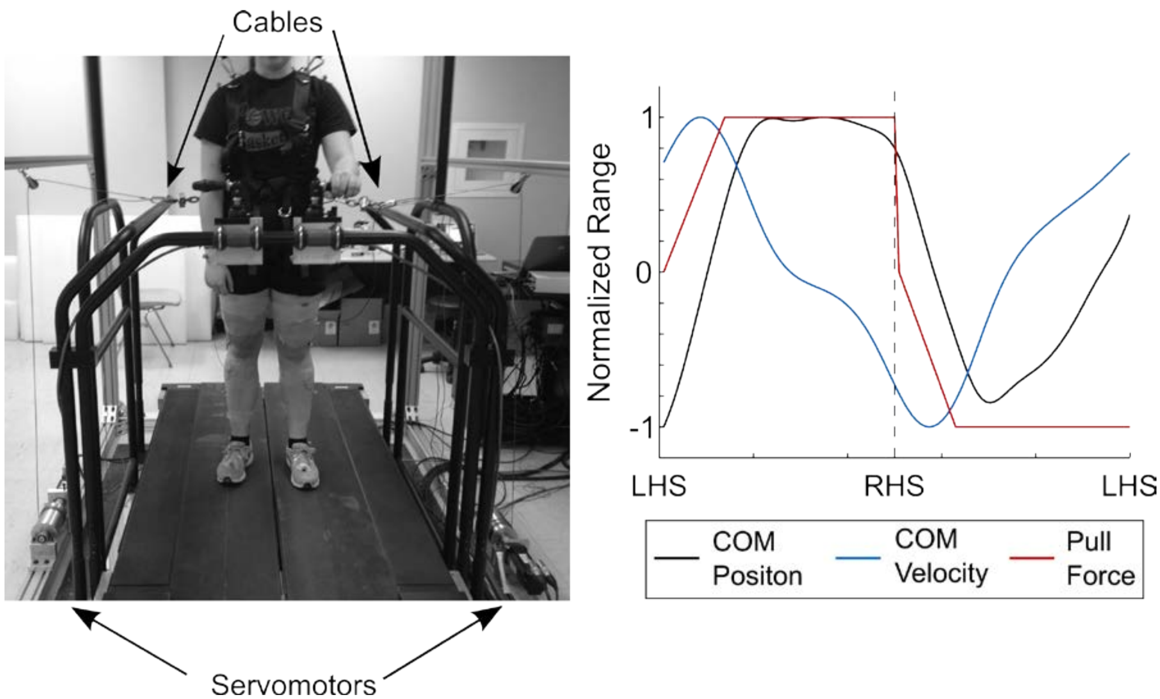

Figure 4-1: Experimental Setup. Participants walked on a split-belt treadmill at their self-selected speed. Two servomotor systems were used to drive a cable spool, with the cables connected to the waist belt on the fall arrest harness. Example net perturbation force for the augment perturbation type is shown on right with sample COM position and velocity over one gait cycle. Perturbation timings were based upon time between successive heel strike events.

Perturbations were controlled using a custom LabVIEW (National Instruments, Austin, TX) program, which separately recorded treadmill ground reaction forces from the instrumented split-belt treadmill to time the perturbation forces to the participant's walking pattern. These recorded forces were used to calculate the whole body center of pressure (COP) in the frontal plane, which was used to detect the heel strike occurrences while the participant walked on the treadmill. Approximate timing of heel strike events were identified by detecting the large changes in the derivative of the medial-lateral COP signal that take place as the participant begins to shift their weight from one leg to the other in early stance. The times between successive steps were calculated on a step-by-step basis, and a running average of the past ten steps was used to time the motor pulls with the participant's walking pattern. This resulted in the timing of the perturbation profile being phased with the medial-lateral COM velocity, while the magnitude of the perturbation was determined by the participant's body weight.

4.3.2 Experimental Protocol

Walking trials were conducted at the participant's self-selected treadmill walking speed, which was determined by slowly increasing the belt speed during an initial familiarization trial until the participant verbally indicated a comfortable pace. The initial two walking trials were used to assess baseline walking over a total of 100 gait cycles per leg, first without the cables connected to the participant, and then with the cables connected. These two trials enabled the characterization of any changes in the walking pattern related to the bilateral baseline perturbation force (~6N) necessary to keep the cables taut during walking. Perturbation trials were conducted while the participant walked at their self-selected treadmill walking speed for a total of 312 gait cycles with the test limb. Each trial was dived into three blocks of 104 gait cycles, with continuous perturbations of frontal plane COM motion applied during the middle block. This block design enabled us to characterize walking changes before, during, and after perturbations were applied, while the continuous trials allowed us to characterize the time course of any adaptation and/or de-adaptation to the perturbations. Additionally, a total of four catch trials were included in each block, occurring randomly every 25-35 steps. During these catch trials, the perturbations were either applied or removed for a single gait cycle, to further characterize adaptations. A total of four perturbation conditions were tested: two force magnitudes, 2.5% and 5% of the participant's body weight, and timed to either accentuate or resist COM motion. Additionally, the effects of a handrail hold were tested by having the participant either hold onto the instrumented handrail with the hand opposite the test leg only, or to hold onto the harness with both hands. These experimental conditions resulted in a total of eight perturbation trials (2 force x 2 type x 2 hold) that were presented in a randomized order. A final normal walking trial of 100 gait cycles with the cables connected was completed at the end of the experiment to evaluate any changes in baseline walking performance from the perturbations.

4.3.3 Data Analysis

Video data were initially processed in Vicon Nexus software to label markers, and run the lower extremity Plug-In-Gait model. Gait events were automatically determined in Matlab (Mathworks, Natick, MA) using a custom algorithm that combined ground reaction force and kinematic event detection methods described by Zeni et al. (2008). Additional data analysis was completed using custom algorithms in Matlab. An eight-segment model consisting of the foot, shank, thigh, pelvis, and trunk was used to estimate whole body COM location (Winter, 2009). COM sway was used to characterize the magnitude of frontal plane movement of the COM, and was calculated as the range of COM movement in the frontal plane over each gait cycle. Temporal and spatial gait parameters were calculated to characterize locomotor adaptations in response to the external COM perturbations.

Statistical analyses were conducted using SPSS 20.0 (IMB, Armonk, NY). Paired t-tests were used to compare step width and COM sway between the two initial walking conditions, to assess any differences due to connecting the cables to the participant. Average responses were obtained within each testing block for both the perturbation condition and the catch trials. A repeated measures ANOVA was carried out separately for COM sway, step width, and cycle duration, to characterize within-subject changes due to the experimental factors of perturbation type, force magnitude, handrail hold, and perturbation block. This analysis method enabled the evaluation of potential interaction effects between the testing conditions, such as the influence of handrail hold during the perturbation block. Since the augment and resist perturbation types had opposite effects on COM movement, separate repeated measures ANOVAs were completed for each perturbation type to simplify the interpretation of the results. If the data for a certain experimental factor was not spherical, a Greenhouse-Geisser correction was used for the within-subject effects. Post-hoc analyses were carried out for significant factors using a Bonferroni correction to account for multiple comparisons. Significance was accepted for $p<0.05$.

4.5 RESULTS

4.5.1 Device Evaluation

The effect of connecting the individual to the cable-driven device was evaluated to identify changes in walking associated with the baseline forces necessary to maintain cable tension. Significant decreases in COM sway ($p<0.001$) and step width ($p<0.001$) were observed when the cables were connected to the trunk (Figure 4-2). Although both cables were equally pulling with a light (~6N) force, this tension force altered COM movement and foot placement in the frontal plane. Locomotor changes were assessed within each experimental trial, and not with respect to the baseline walking trials.

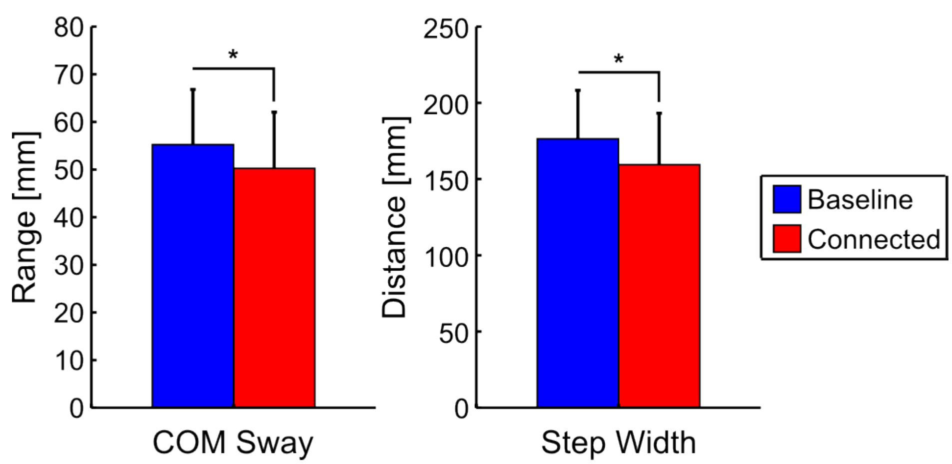

Figure 4-2: Baseline Cable Tension Alters Walking. Baseline cable forces needed to maintain tension within the cables resulted in significant decreases in frontal plane COM movement and step width (* $p<0.05$, paired t-test).

The custom control program was able to correctly count the number of steps taken with each leg, despite the participant simultaneously stepping on both treadmill belts. There was a slight delay between the event identified from the COP and the actual heel strike event from the vertical ground reaction forces (~250ms), but the observed cycle and step times were similar. Using the observed step times resulted in a perturbation profile that phased with COM velocity in the frontal plane. The augmenting perturbations were in phase with COM velocity, while the resisting perturbations were approximately 180° out of phase with COM velocity (Figure 4-3). The cable driven device was able to deliver controlled perturbations of frontal plane COM motion that synchronized with individual walking patterns.

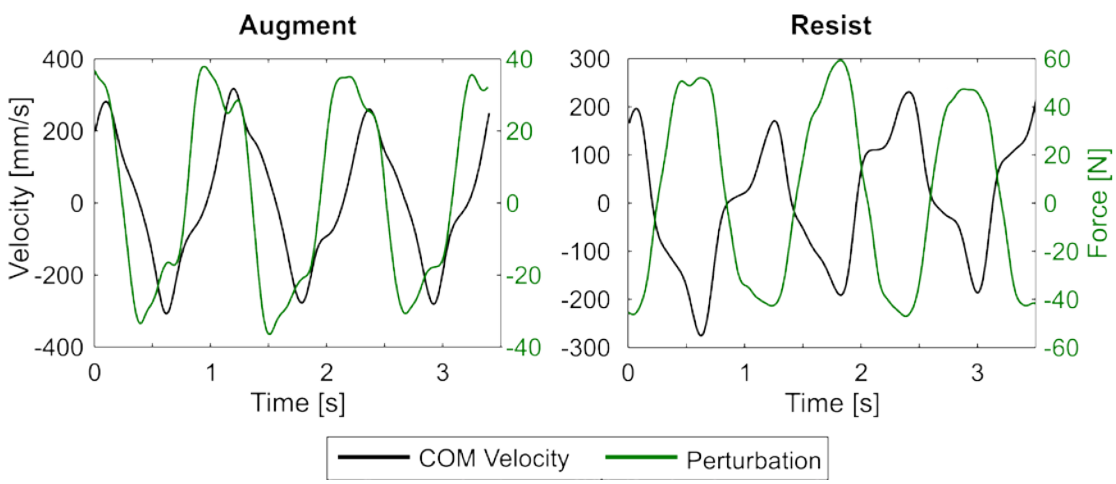

Figure 4-3: Perturbation Force Timing. COM velocity (black line) and net perturbation force (green line) from three consecutive gait cycles in the perturbation block from a single participant (AD306). The perturbation force was in phase with the COM velocity during the augmenting perturbations, and 180° out of phase with COM velocity for the resisting perturbations.

4.5.2 Response to Trunk Perturbations

Example step widths and COM sway for two representative participants are presented in (Figure 4-4 and Figure 4-5). The augment and resist perturbation types had opposite effects upon trunk movement. This difference resulted in a significant main effect of perturbation type for COM sway ($p<0.001$), step width ($p<0.001$), and cycle duration (p<0.001). In order to reduce the number of interaction effects, simplifying the interpretation of the data, separate repeated measures ANOVAs were carried out for the two perturbation types for each experimental measure. The results for each perturbation type are presented separately below.

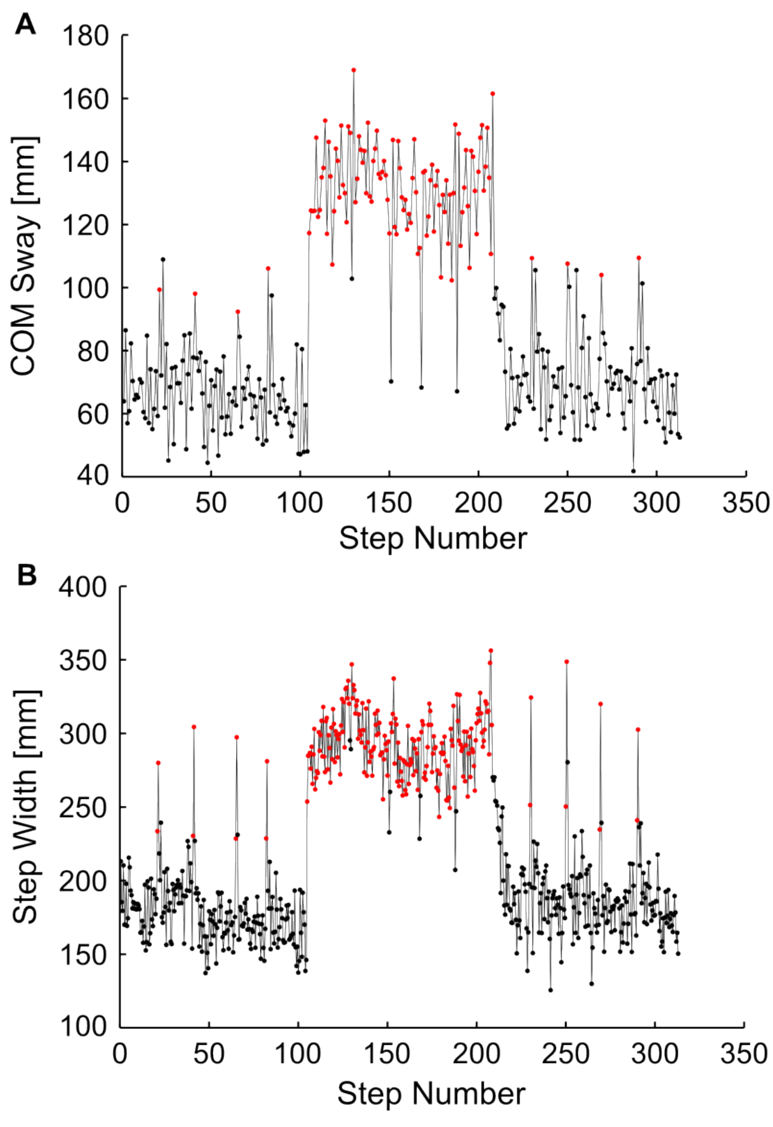

Figure 4-4: Response to Augmenting Perturbation for a Single Participant. COM Sway (A) and step width (B) during the 5% BW perturbation force trial from a single participant (AD302). Red circles indicated the perturbation was applied during the gait cycle. Both COM sway and step width increased in response to the augmenting perturbation condition.

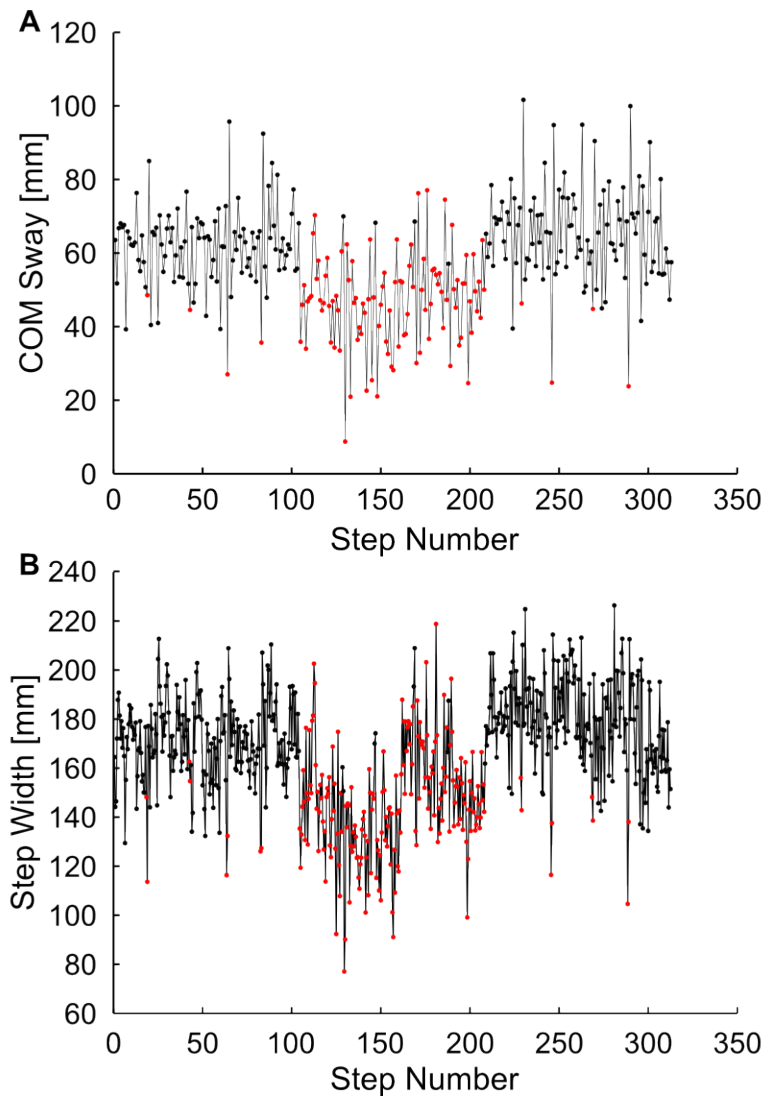

Figure 4-5: Single Participant Response to Resisting Perturbations. COM Sway (A) and step width (B) during the resisting, 2.5% BW perturbation force trial from a single participant (AD303). Red circles indicate the perturbation was applied during the gait cycle. Both COM sway and step width decreased when perturbations resisting COM movement were applied.

4.5.2.1 Augment Perturbations

Comparison of the average COM sway within each testing block for the augment perturbation condition revealed significant main effects of force magnitude ($p=0.002$) and testing block ($p<0.001$), as well as a significant interaction effect between these two main effects ($p < 0.001$). COM sway was significantly higher in the perturbation block compared to the pre and post ($p<0.001$) perturbation blocks, while no average differences in COM sway were observed between the pre and post perturbation blocks. Increasing the magnitude of the force from 2%BW to 5%BW increased the amount of COM sway in the perturbation block (Figure 4-6). The interaction between force magnitude and testing block is likely due to the increase in sway during the perturbation block, since no differences were found between the pre and post perturbation blocks for the two force magnitudes. There was no significant effect of handrail hold observed for this perturbation type ($p=0.074$).

The observed increased COM sway was accompanied by increases in step width when augmenting perturbations were applied. Main effects of force magnitude ($p=0.015$), handrail hold ($p=0.001$), and testing block ($p<0.001$), and an interaction between force magnitude and testing block ($p=0.003$) were observed. Handrail hold had a general effect of reducing step width across the three testing blocks. The augmenting perturbations resulted in larger step widths during the perturbation block compared to the pre ($p=0.001$) and post blocks ($p=0.001$) (Figure 4-6). As force magnitude increased, step width also increased, but only during the perturbation block, which explains the block and force magnitude interaction effect. No significant differences were observed between the pre and post stimulation blocks for step width.

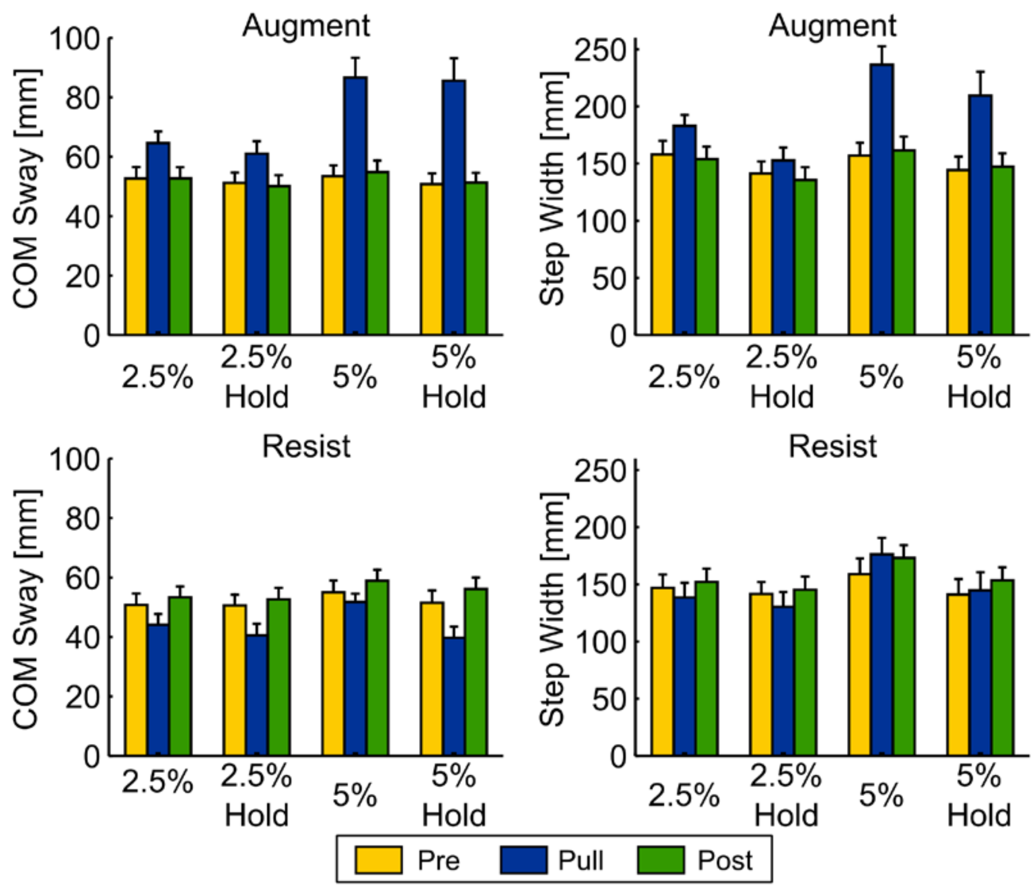

Figure 4-6: Average COM sway and Step Width Response to Perturbations. The augment perturbations (top) increased COM sway and step width, with larger changes observed at higher perturbation forces. Conversely, the resist perturbations reduced COM sway, with smaller reductions observed in step width. Handrail hold had a general effect of reducing step width across the entire trial, but a significant effect was only observed for the resist perturbation type.

In addition to the changes in frontal plane gait parameters, cycle duration was also altered when accentuating perturbations were applied. The application of accentuating forces resulted in decreased cycle duration during the perturbation block compared to the pre ($p<0.001$) and post ($p<0.001$) perturbation blocks (Figure 4-7). As the perturbation magnitude increased, the cycle duration further decreased, but only when the perturbation was applied. Holding onto the handrail had the general effect of slightly increasing cycle duration across all testing blocks ($p=0.001$).

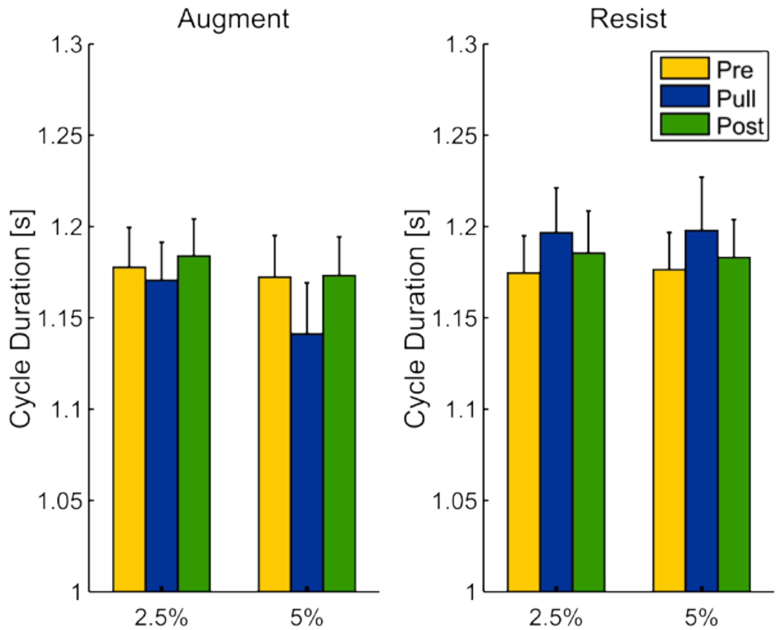

Figure 4-7: Group Gait Cycle Duration. Gait cycle duration decreased in response to the augment perturbation, and increased in response to the resist perturbation. An effect of force magnitude was only observed in for the augment perturbation type.

4.5.2.2 Resist Perturbations

The resisting perturbations acted to reduce COM sway when applied during walking (Figure 4-6). Significant main effects of perturbation magnitude (p=0.013), handrail hold (p =0.003), and testing block (p <0.001) were observed. Additionally, interaction effects were observed between handle hold and block (p < 0.001), and handle hold, force magnitude, and block (p = 0.005). When the perturbation was applied during walking, COM sway was reduced compared to the pre perturbation (p=0.005) and post perturbation block (p=0.001). Removal of the resisting perturbations resulted in larger amounts of COM sway compared to the pre perturbation block (p=0.004). Holding onto the handrail caused further reductions in COM sway, but this effect only occurred during the perturbation blocks.

In contrast to the accentuating perturbations, changes in COM sway were not coupled with step width changes for the resisting perturbations (Figure 4-6). An interaction between force magnitude and testing block ($p=0.031$) is likely due to an observed trend towards reduced step width at the 2.5% force magnitude, but not 5%. Similar to the accentuating perturbations, handrail hold had the general effect of reducing step width across the entire trial ($p<0.001$).

The resisting perturbations acted to increase cycle duration (Block, $p=0.006$), with the perturbation ($p=0.011$) and post ($p=0.001$) testing blocks having a longer cycle duration compared to the pre perturbation block. Handrail hold further increased cycle duration when perturbations were applied ($p=0.001$).

4.5.3 Locomotor Adaptations

We examined the cycle-by-cycle responses to characterize any short-term changes occurring within the each block. Additionally, catch trials were evaluated to characterize any potential adaptation to the perturbations. No significant differences in the average COM sway or step width were observed between the pre and post perturbation blocks for either perturbation type.

4.5.3.1 Augment Perturbations

Group average COM sway and step width over entire trial are presented in Figure 4-8, and a subset of the steps at the block transition points are shown in Figure 4-9. Step width and COM sway both rapidly increase in response to the perturbation. COM sway remains consistent over the course of the perturbation block, but there is a decreasing trend within the perturbation block for step width. A paired t-test was used to compare the average step width of the first and last 15 cycles in the perturbation block. Significant differences were observed without the handrail hold for both force levels (2.5% p=0.003, 5% p=0.021), but no difference was observed when the participant held onto the handrail (2.5% p=0.153, 5% p=0.548). When the perturbations were removed COM sway values returned back to baseline levels within five cycles, while step widths took about ten cycles to return to baseline values (Figure 4-9).

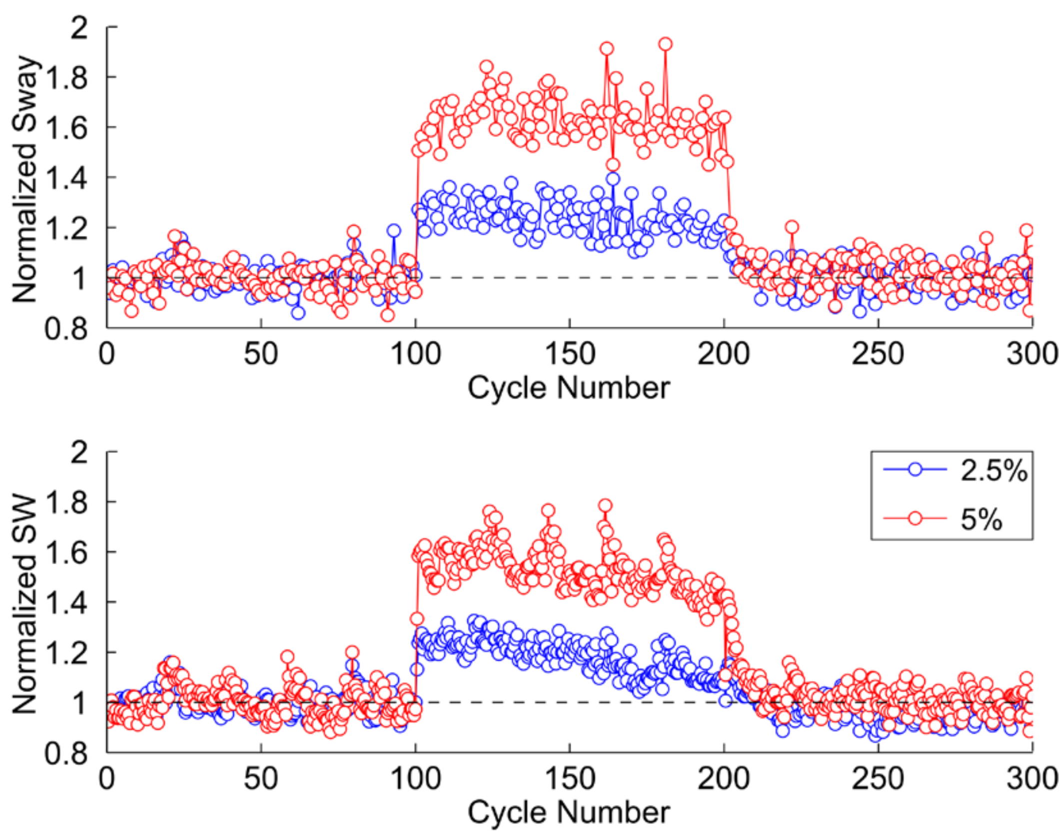

Figure 4-8: Time Course of Locomotor Changes to Augment Perturbation. Ensemble averaged group response to augment perturbation type without the handrail hold. Catch trials were removed, and values were normalized to each participant's average response in pre perturbation block prior to ensemble averaging. Perturbations were applied during cycles 101 to 200. COM sway and step width quickly increased when perturbations were applied, and also quickly returned to baseline levels when perturbations were removed.

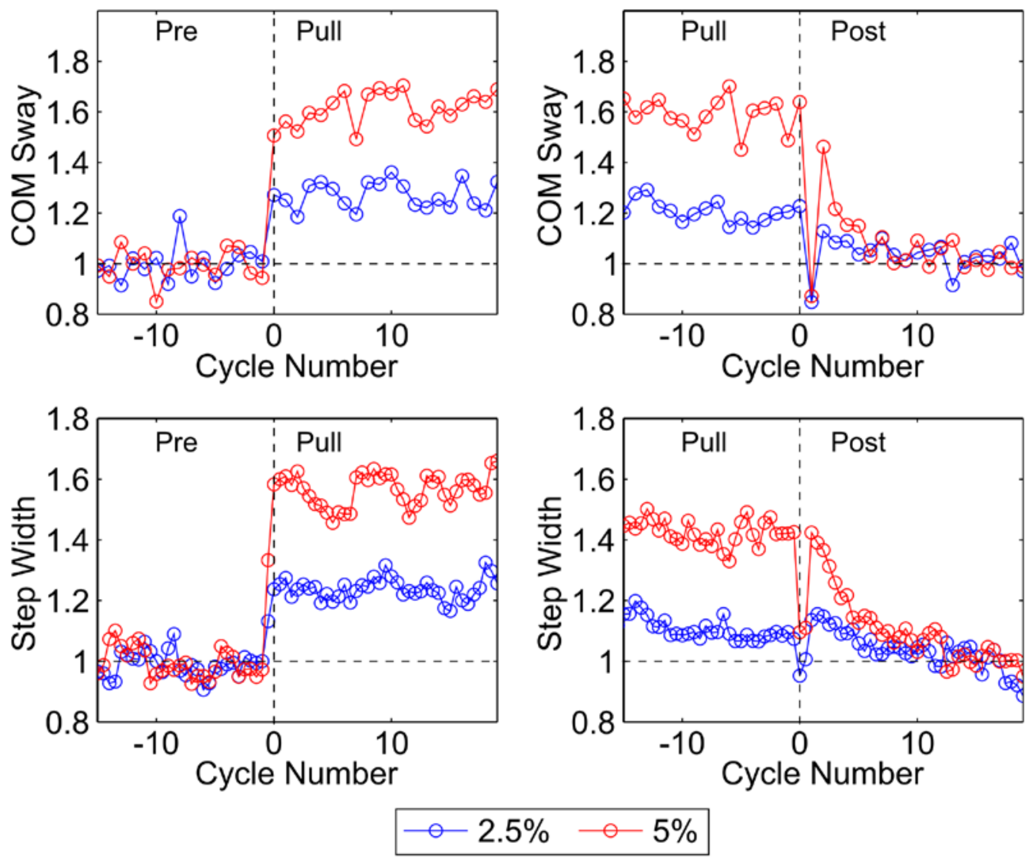

Figure 4-9: Response to Augment Perturbations at Block Transitions. Time course of adaptation (left) and de-adaptation (right) to augment perturbations without the handrail hold. Values are normalized to average of pre perturbation block. Longer rates of de-adaptation compared to adaptation are likely due to increased fall risk of increased COM movement.

Comparison of perturbation catch trials (Figure 4-10) during the pre and post blocks demonstrated no significant differences between the continuous and single-step perturbations for COM sway (Block, p=0.170). As the force magnitude increased, the continuous perturbations produced larger amounts of COM sway compared to the single cycle perturbations (Force*Block, p=0.041). No significant differences were observed in the pre and post block catch trials for step width (Block, p=0.137). There was a trend towards smaller step widths in the post compared to the pre perturbation block at the 2.5% force level (Force*Block, p=0.057). Catch trials during the perturbation block

(Figure 4-10) produced significantly lower amounts of COM sway compared to the pre and post perturbation blocks (Block, p=0.033). There was a significant interaction effect of force level and testing block for step width (p=0.015). This interaction effect is due to larger step widths, when the perturbation was removed, only at the 5%BW force level.

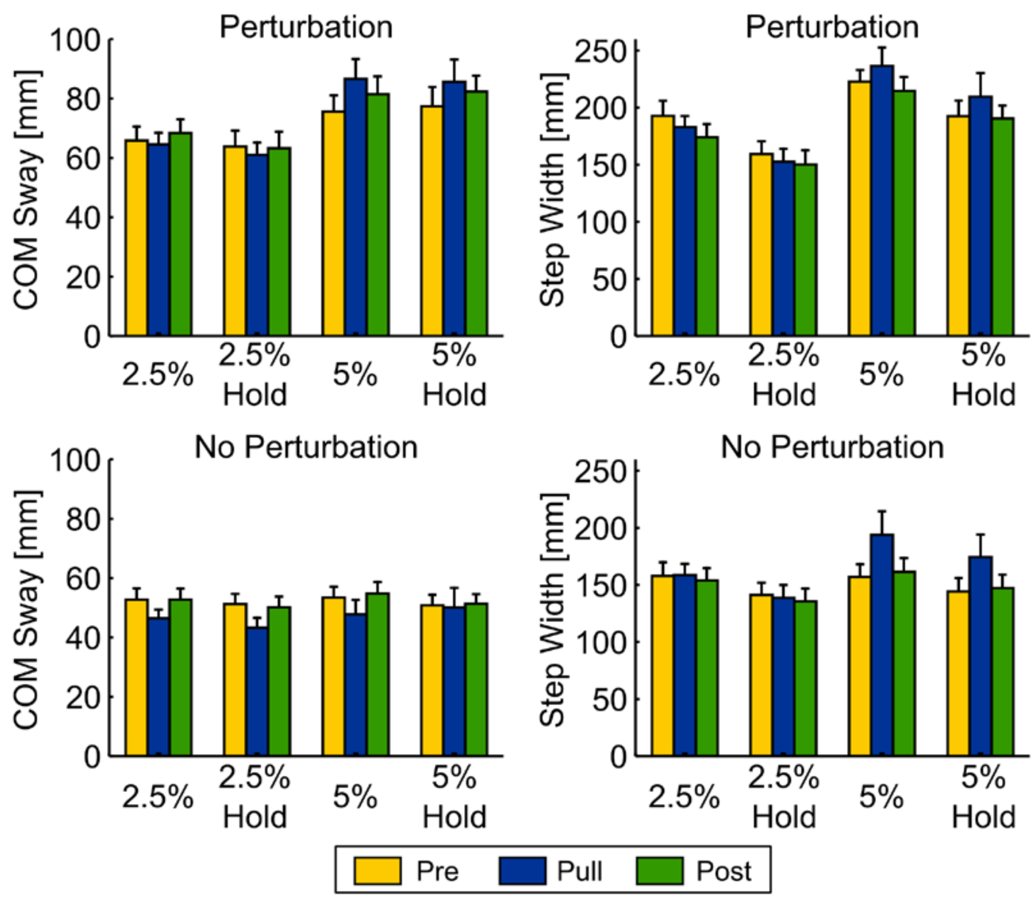

Figure 4-10: Augmenting Perturbation Catch Trials. Catch trials from augment perturbation trials. Top plots compare catch trials applying perturbation in pre and post blocks with average response from perturbation block. Bottom graphs present catch trials removing perturbation during pull block, with average response from the pre and post perturbation trials. Perturbation catch trials similar responses in COM sway and step width compared to continuous perturbations. No perturbation catch trials resulted in lower COM sway, and larger step widths compared to baseline walking.

4.5.3.2 Resist Perturbations

The resist perturbations did not demonstrate any significant changes in COM sway or step width over the course of the perturbation block (Figure 4-11). A short-term aftereffect of the resist perturbations can be observed in step width, but only at the 5% force level (Figure 4-12). Removal of the stimulation produces an initial increase in step width, which returns back to baseline levels within ten to fifteen cycles.

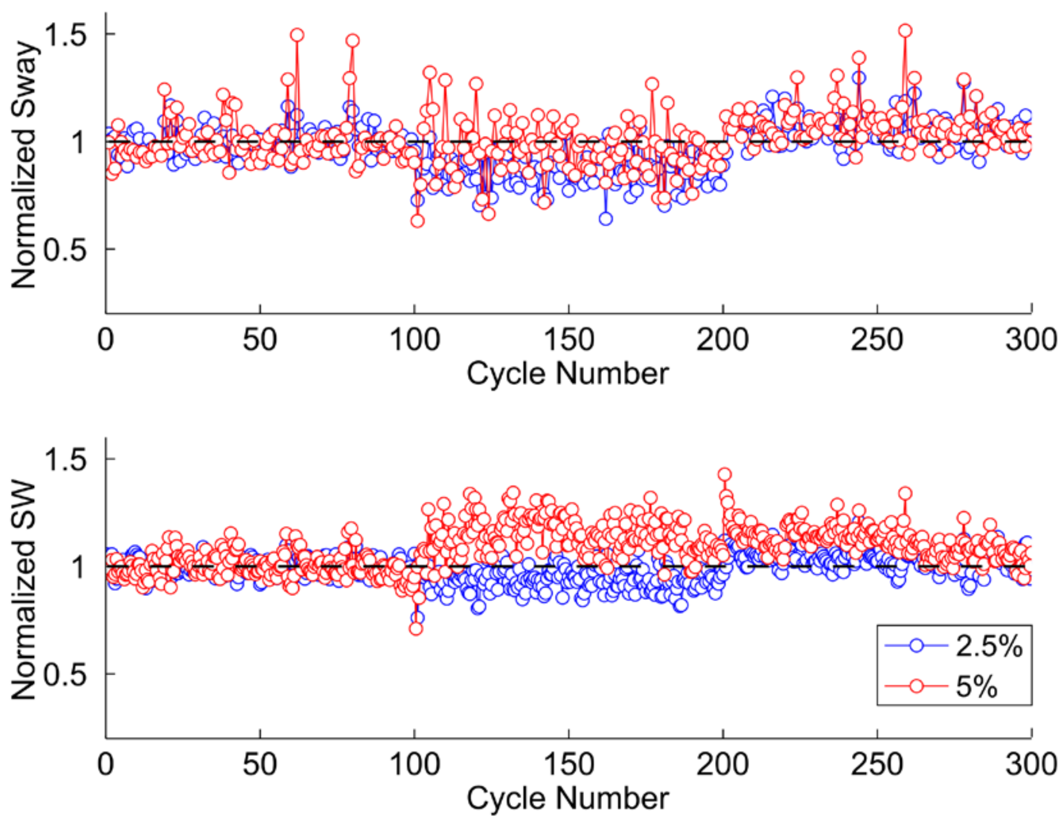

Figure 4-11: Time Course of Average Group Response to Resist Perturbation. Ensemble averaged group response to resist perturbation type (applied during cycles 101 to 200). Catch trials were removed, and values were normalized to each participant's average from pre perturbation block prior to ensemble averaging.

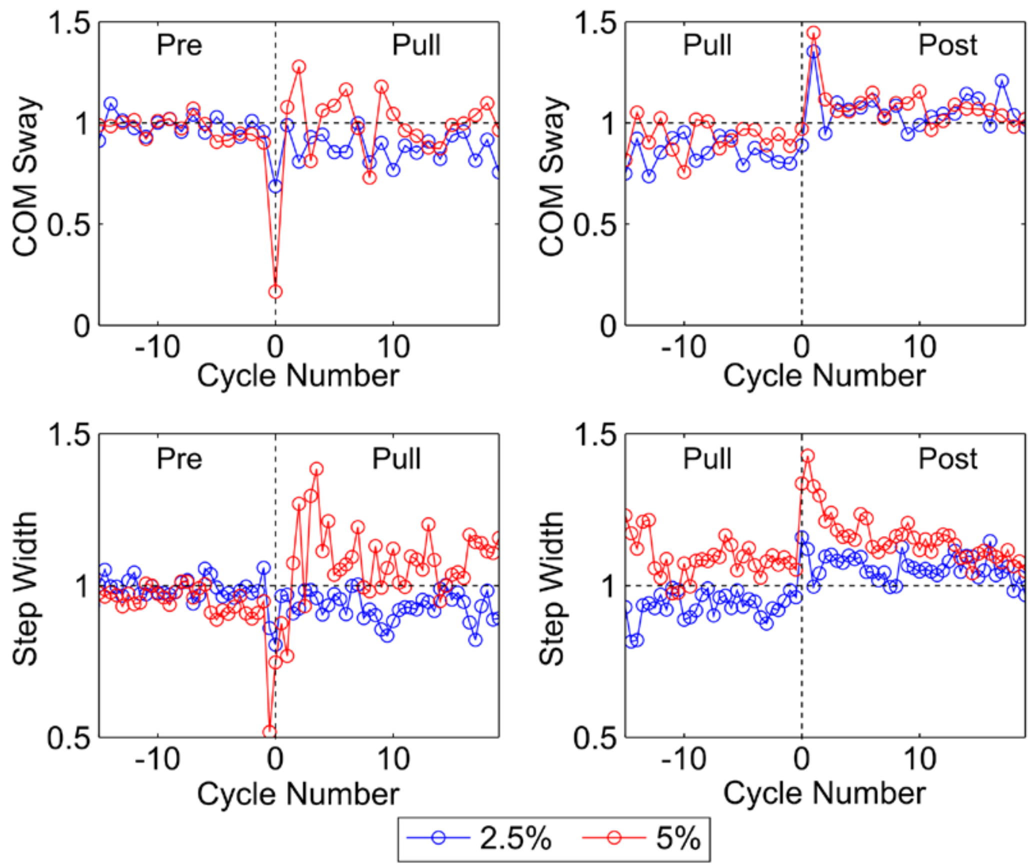

Figure 4-12: Temporal Response to Resist Perturbations at Block Transitions. Averaged group response to resist perturbations for 15 cycles before, through 20 cycles after transition between testing blocks (cycle 0). Removal of perturbation, especially at 5%, demonstrates short-term aftereffect of increased step width for 10-15 cycles.

Perturbation catch trials during the pre and post blocks resulted in significantly lower amounts of COM sway compared to the perturbation block (Block, $p<0.001$), with larger decreases observed at the 5% force level (Force*Block, $p<0.001$). Additionally, step widths were also lower during the perturbation catch trials (Block, $p=0.037$), and were further decreased at the 5% force level (Force*Block, $p=0.015$). Catch trials removing the perturbations resulted in increased levels of COM sway compared to the pre and post blocks (Block, $p<0.001$). COM sway further increased when the perturbations were removed at the 5% force level (Force*Block, $p<0.001$). In response to these catch trials, step widths were larger than those in the pre and post blocks (Block, p<0.001), while the handrail hold reduced this difference (Hold, p<0.001; Hold*Block, p=0.015).

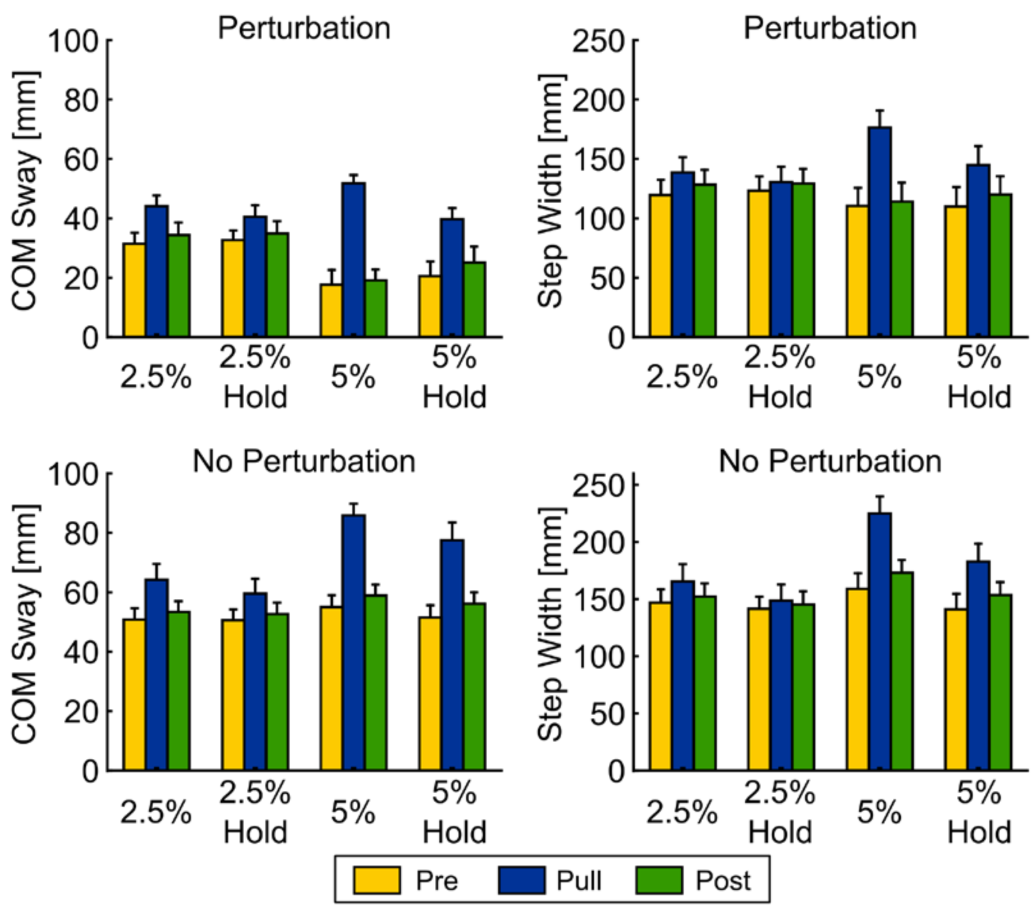

Figure 4-13: Catch Trial Comparison for Resist Perturbation. Larger reductions in step COM sway were observed for catch trials applying perturbation in pre and post blocks, compared to continuous perturbation in pull block (Top plots). COM sway was also larger for catch trials removing perturbation compared to baseline levels in pre and post blocks (lower plots).

4.6 DISCUSSION

Results from this experiment demonstrate our cable-driven device was able to apply medial-lateral perturbations to the trunk, which altered frontal plane movement of the COM. Perturbations intended to resist COM movement in the frontal plane decreased COM sway, while perturbations designed to accentuate COM movement increased COM sway for all participants. Participants were able to adapt to these external perturbations, maintaining dynamic balance largely though adjustments made to step width. The type of perturbation applied, either to augment or resist COM movement, had the largest impact upon the gait adjustments made. Perturbation force magnitude and handrail hold scaled the magnitude of the response. The timing of these adjustments indicates that dynamic balance control strategy quickly reacts to conditions challenging balance, while taking a more conservative approach to conditions reducing balance demands.

This study provides a unique insight into how individuals adapt to continuous external perturbations accentuating trunk movement in the frontal plane. Medial-lateral perturbations have been previously applied during walking, demonstrating that individuals placed their foot more lateral to account for the increased lateral trunk movement (Hof et al., 2010). However, these perturbations were only over a single step, and do not provide insight into the time course of the adaptations. Similar to previous studies, we observed that participants primarily modified lateral foot placement to account for changes in COM movement. Continuous accentuating perturbations resulted in larger step width increases compared to those observed during the single cycle catch trials. Most of the foot placement adaptation was achieved over one or two steps, but additional increases were observed over the next three to five steps. The observed adaptations occurred within the first five steps, while de-adaptation to the accentuating perturbation took approximately ten to fifteen steps (Figure 4-9). Reisman et al. (2005) observed that intralimb parameters (stance time and stride length) quickly adapted to split-belt speed perturbations and demonstrated post-adaption aftereffects, while interlimb parameters slowly (double support time) adapted with no aftereffects. They hypothesized that the slower rates of adaptation in the interlimb parameters may be done to restore a symmetric walking pattern, after the intralimb parameters change to adjust to the speed differences. Overall, in our study we observed much faster rates of adaptation and de-adaptation than the rates observed for the interlimb parameters during split-belt walking. These fast adaptation rates are likely due to the need to quickly adjust the base of support to prevent a fall, supporting the theory that the central nervous systems focuses more upon on controlling frontal plane balance (Bauby & Kuo, 2000). However, we did observe slower de-adaptation rates for all accentuating conditions, as well as a significant decrease in step width over the course of the perturbation block at the 2% force level. These observations suggest that dynamic balance control strategy takes a more conservative approach to reducing the base of support when balance demands are reduced.

Analysis of the catch trials provided further insight into the locomotor adjustments made in response to the balance perturbations. Augment perturbation catch trial adjustments were similar to those observed during the continuous perturbations. The smaller step width responses observed during catch trials at the 5%BW force level is likely due to the increased COM sway when repeated accentuating perturbations are applied. Contrastingly, catch trials removing the assist perturbation produced lower levels of COM sway, but larger step widths at the 5% force level, when compared to baseline levels in the pre and post perturbation blocks. The presence of lower COM sway indicates that participants are making other adjustments, likely faster gait cycle times, to reduce COM movement in response to the assist perturbations. Increased step width when the perturbation is removed, supports the idea of a more conservative balance control strategy towards reducing the base of support. Resisting perturbation catch trials produced larger decreases in COM sway than the continuous perturbations, likely due to participants taking a quick step to maintain balance in response to the perturbation pulling the COM back towards the swing leg. Participants were able to adapt to the continuous resist perturbations, as evidenced by larger amounts of COM sway when perturbations were removed during the catch trial. Similar to the augmenting perturbations, adjustments increasing gait cycle duration in response to the continuous perturbations would act to increase COM sway when perturbations are removed. Differences in COM sway when both the accentuating and resisting perturbations were removed, indicates that although step width control was the primary mechanism for dynamic balance control, participants also made spatial and temporal adjustments to help control COM movement.

Modest decreases in step width were observed when resisting forces were applied and the individual held onto the handrail. Lateral stabilization of the trunk during walking decreases both step width and energy consumption in young (Donelan *et al.*, 2004) and old neurologically intact individuals (Dean *et al.*, 2007). The smaller decreases in step width observed in our study may be due to the difference in how the forces were applied to the trunk. Donelean et al. (2004) used springs attached to the trunk that were intended to stabilize the trunk in a certain location. The perturbations in our study were timed to the walking cycle, and not based upon stabilizing COM location in the center of the treadmill. Interestingly, we did observe that simply connecting the participant to the cable robot did cause a reduction in step width and COM movement. It is likely that the light (~6N), lateral forces did provide some stabilization of trunk movement.

The effect of handrail hold on the magnitude of the locomotor adaptations provides important insight for future studies. Treadmill walking studies often involve the use of the handrails to ensure participant safety, especially for the elderly or individuals with a neurologic disorder. Handrail hold has been shown to reduce step length and width variability during treadmill walking (Owings & Grabiner, 2004). Additionally, a light touch force when using a cane is sufficient to stabilize movement of the pelvis in stroke survivors (Boonsinsukh *et al.*, 2009). Therefore, it is possible that holding onto the handrail could significantly alter how individuals adjust to balance perturbations. We observed a trend towards the handrail hold impacting step width responses to the augment (Hold*Block, p=0.067) and resist (Hold*Block, p=0.08) perturbations . The handrail hold augmented locomotor changes in step width and cycle time when resisting perturbations were applied. However, handrail hold may have a more significant effect for individuals with neurological disorders, since the handrail could be used to provide postural support or assist with controlling COM movement.

In conclusion, this study has demonstrated that a cable-driven device can be used to deliver perturbations of lateral COM movement that are phased to the participant's walking pattern. Additionally, young healthy participants were able to make the necessary modifications to step width to maintain dynamic balance during treadmill walking. Application of forces to accentuate COM movement produced more robust balance adjustments, which were not strongly influenced by holding onto the handrail. Differences in the rates of adaptation and de-adaptation suggest that dynamic balance control strategy prioritizes adjustments to prevent falls, but is more conservative with making adjustments when the balance demands are lessened. These results validate the use of the cable-driven system to create novel balance environments to study dynamic balance control during walking. Additionally, this device may be useful to examine changes in dynamic balance control strategy for individuals with neurological disorders, such as stroke.

CHAPTER 5: LOCOMOTOR ADAPTATIONS TO CONTINUOUS, EXTERNAL PERTURBATIONS OF THE TRUNK IN STROKE SURVIVORS.

5.1 INTRODUCTION

Rehabilitation techniques are largely focused upon improving walking speed in chronic stroke survivors, but compensatory strategies may enable stroke survivors to regain walking speed with non-normal kinematic patterns (Huitema et al., 2004). These compensatory strategies may ultimately limit walking function through their contribution to spatial and temporal asymmetries, as well as increased energy expenditure during walking (Chen *et al.*, 2005*a*). Evaluation of balance control during walking may provide deeper insight into the impact of specific walking patterns on walking function. The purpose of this study was to evaluate locomotor changes made by chronic stroke survivors to maintain balance in response to external perturbations of trunk motion.

Successful locomotor control requires that individuals continuously adjust their walking pattern to meet the current environmental demands. In a controlled setting, such as the laboratory, specific aspects of the environment and walking task can be selectively altered to provide valuable insight into the underlying locomotor control mechanisms and strategies. Healthy individuals have been shown to modify their step length and width in response to changes in optic flow during treadmill walking (O'Connor & Kuo, 2009). Individuals are also able to adjust their walking pattern to adapt to speed differences between the legs when walking on a split-belt treadmill (Dietz *et al.*, 1994; Reisman *et al.*, 2005) Chronic stroke survivors also demonstrate the ability to adapt to belt speed differences between the legs, but this adaptation occurs at a slower rate compared to neurologically intact individuals (Reisman *et al.*, 2007). When the treadmill belts returned to the same speed, chronic stroke survivors demonstrated a short-term aftereffect, resulting in a more symmetric gait pattern compared, which also transferred to overground walking (Reisman et al., 2009). Therefore, in addition to providing insight into the control of walking, novel experimental environments may also have a potential therapeutic effect to improve walking function in chronic stroke survivors.

The application of controlled perturbations during standing or moving provides a means to characterize the underlying balance control strategy. Lateral perturbations of the support surface during standing have been used to characterize responses over the course of rehabilitation (Kirker et al., 2000), as well as differences between fallers and non-fallers post-stroke (Marigold & Eng, 2006a). During walking, short duration perturbations of lateral trunk movement have been delivered in young adults (Hof et al., 2010), confirming that a lateral foot placement control strategy is used to maintain dynamic balance in the frontal plane (Hof, 2008). However, reduced ability to accurately sense trunk position post-stroke (Ryerson et al., 2008), and difficulties making frontal plane step adjustments with the paretic limb (Nonnekes et al., 2010) would impair the planning and execution of lateral foot placement during walking. These functional impairments would reduce the effectiveness of using foot placement control strategy to maintain dynamic balance.

In this study, we proposed the use of external perturbations of frontal plane trunk movement to evaluate potential changes in dynamic balance control strategy post-stroke. Perturbations were applied within a block design to characterize both the locomotor adaptations made to maintain balance, as well as the time course of these adaptations. We hypothesized that stroke survivors would demonstrate less foot placement modulation, indicating a shift in balance control strategy from placement control towards COM movement control.

5.2 METHODS

5.2.1 Participants

Ten chronic (> 6 month) stroke survivors with unilateral brain injury and ten age and sex-matched individuals with no reported neurological deficits participated in this study. Exclusion criteria for this study included recent use of botulinum toxin in the lower extremity, inability to walk independently (with or without use of an assistive device), lesion to brainstem centers, diagnosis of other neurologic disorders, or inability to provide informed consent. Prior to beginning the experimental session, a licensed physical therapist conducted a clinical evaluation of the stroke participants, consisting of the lower extremity Fugl-Meyer test (Fugl-Meyer et al., 1975), Berg Balance Assessment (Berg et al., 1992), dynamic gait index (Jonsdottir & Cattaneo, 2007), and 10 meter walking test (Mudge & Stott, 2009). For controls, only the 10 meter walking test was completed to assess their comfortable overground walking speed. Participant characteristics are summarized in Table 5-1. The Marquette University Institutional Review Board approved all experimental procedures, and written informed consent was obtained from all individuals prior to participating in this study.

Subject ID	Age [yrs]	TPI [months]	Affected Side	DGI	LE FM	Berg	Ten Meter [m/s]	Treadmill Speed [m/s]
CRS301	66	330	L	21	19	46	0.837	0.45
CRS302	55	74	L	15	24	49	0.988	0.45
CRS303	56	130	R	24	31	56	1.271	0.60
CRS304	66	10	R	19	27	54	0.995	0.70
CRS305	60	123	R	21	29	46	1.361	0.50
CRS306	63	275	R	21	32	49	1.270	0.45
CRS307	55	79	L	17	28	41	0.635	0.40
CRS308	62	38	L	15	30	46	0.732	0.30
CRS309	50	125	R	17	22	51	1.173	0.45
CRS310	76	47	R	23	30	34	1.091	0.70
CRC301	60	-	-	-	-	-	0.962	0.75
CRC302	58	-	-	-	-	-	1.515	1.00
CRC303	55	-	-	-	-	-	1.212	1.05
CRC304	64	-	-	-	-	-	0.980	1.00
CRC305	59	-	-	-	-	-	1.333	0.90
CRC306	62	-	-	-	-	-	1.299	0.85
CRC307	55	-	-	-	-	-	1.325	1.00
CRC308	64	-	-	-	-	-	1.429	1.10
CRC309	54	-	-	-	-	-	1.389	1.10
CRC310	71	-	-	-	-	-	1.141	0.65

Table 5-1: Participant Characteristics. Dynamic Gait Index (DGI) maximum score 24. Lower extremity Fugl-Meyer (LE FM) maximum score 34. Berg Balance Test (Berg) maximum score 56.

5.2.2 Cable-driven Perturbations

A novel cable-driven device (similar to Wu *et al.*, 2011) was constructed to deliver medial-lateral perturbations to the trunk during treadmill walking. The cable-driven device consisted of a servomotor system (AKM-33H, AKD-0606, Kollmorgen, Radford, VA) that drove an aluminum spool with a light stainless steel cable attached (Figure 5-1). The system was capable of delivering pulls up to 100N, and a device was placed on the left and right side of the treadmill to deliver both left and right perturbations. Each cable ran through a pulley and attached to the belt of a fall arrest harness worn by the individual, with the harness and pulley height adjusted to have the cable connections near the top of the pelvis. This location enabled us to deliver external perturbations near the approximate location of the participant's center of mass.

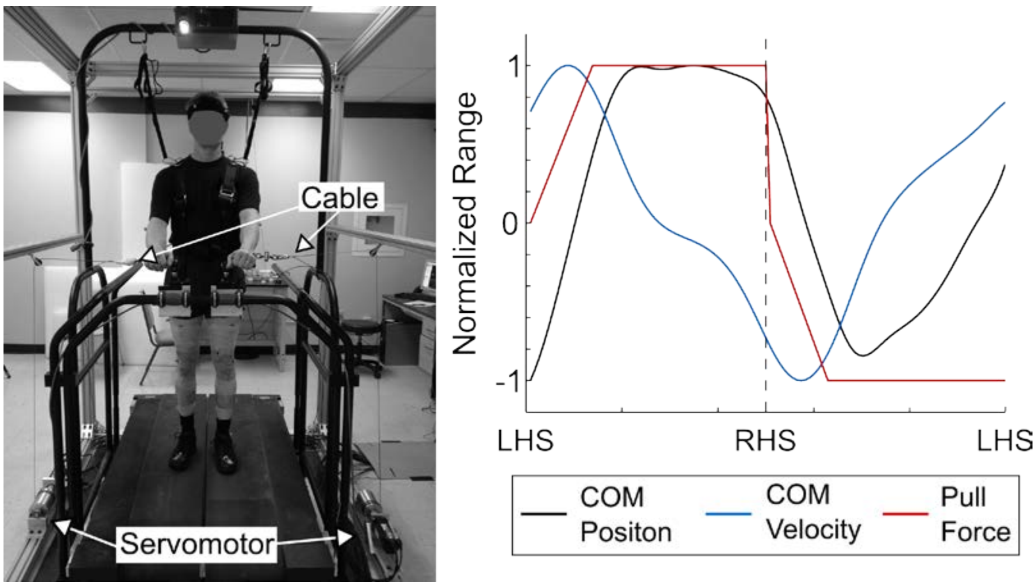

Figure 5-1: Experimental Setup. Perturbations were delivered to the participant through stainless-steel cables that were attached to the waist belt of the harness. The cables were connected to a servomotor system on the left and right ride of the participant, which controlled the timing and magnitude of the pulling forces. An example of the force profile for the augmenting perturbations is shown in the right graph. This profile is based upon the timing of successive heel strike events, and was phased with fontal plane COM velocity.

Perturbations were controlled using custom LabVIEW (National Instruments, Austin, TX) software, which used treadmill ground reaction forces from the instrumented split-belt treadmill to time the perturbation forces to the individual's walking pattern. Ground reaction forces were used to determine the location of whole body center of pressure (COP) in the frontal plane every 20ms. The occurrence of heel strike events were approximated by detecting the peak changes in the derivative of the medial-lateral COP signal, which take place as the participant begins to shift their weight from one leg to the other in early stance. This method produced similar timings between the approximated and actual heel strike events, without constraining the participants from stepping with one leg on each treadmill belt. Timing between successive steps were calculated on a step-by-step basis, with a running average of the past ten steps was used to time the motor pulls to the individual's walking pattern, including any temporal asymmetries. This algorithm resulted in the perturbation profile being phased with the medial-lateral COM velocity. The direction of perturbation forces could be set to either augment or resist COM movement, while the perturbation magnitude was determined by the participant's body weight.

5.2.3 Data Collection

Fifteen passive infrared reflective markers were placed at anatomical locations according to the Plug-In-Gait model (Davis et al., 1991) to capture lower extremity movement. Additionally, markers were placed bilaterally on the wrist, elbow, shoulder, front and back head, and on the C7 vertebra to quantify movements of the upper extremity and head. Marker locations were recorded at 100Hz using an eight camera Vicon motion capture system (Vicon Motion Systems Ltd, Oxford, UK). Ground reaction forces were recorded from the instrumented, split-belt treadmill (FIT, Bertec, Colombus, OH). A custom adjustable handle instrumented with a six degree of freedom load cell (AMTI, MC3A-250, Watertown, MA), was attached the front handrail of the treadmill to quantify handrail hold forces and torques. Handle forces were amplified at 1,000 V/V, and low pass filtered at 500 Hz prior to collection (Gen5, AMTI Inc., Watertown, MA). Perturbation forces were measured using a load cell (MLP-300, Transducer Techniques Inc., Temecula, CA) attached in line with the cable. Signals were amplified at 450V/V and lowpass filtered at 250Hz prior to collection (TMO-1-24, Transducer Techniques Inc., Temecula, CA). Ground reaction forces, handle forces, and cable perturbation forces were all sampled at 1000Hz using a Vicon Mx Giganet, which synchronized the analog and video data.

5.2.4 Experimental Protocol

Walking trials were conducted at the participant's self-selected, comfortable walking speed. All participants were placed in a fall arrest harness, and held onto an instrumented handle in front of the treadmill with the non-paretic (non-test) hand for safety. Self-selected treadmill speed was determined during an initial familiarization trial by slowly increasing the belt speed until the participant identified a comfortable pace. An initial walking trial assessed baseline walking over a total of 105 gait cycles per leg. Next a perturbation familiarization trial was conducted to ensure participants could safely participate in the experiment. During this trial, forces were applied to assist COM motion, with the force level starting at 1% of the subjects body weight (BW), and increased to 1.5% BW then 2% BW every 30 steps. After this familiarization trial, each perturbation trial was conducted in three blocks of 35 gait cycles, with continuous perturbations of frontal plane COM motion applied during the middle block. This block design enabled us to characterize walking changes before, during, and after perturbations were applied. Three perturbation force levels were tested for the accentuating forces: 2%, 3.5%, and 5% BW. At the 3.5% force level, perturbations augmenting and resisting COM motion, and the effects of keeping a head mounted laser within a stationary target projected onto the wall in front of the treadmill were tested. These testing conditions resulted in a total of six perturbation trials that were presented in a randomized order.

5.2.5 Data Analysis

Video data were initially processed in Vicon Nexus to label markers, and run the lower extremity Plug-In-Gait model. Gait events were automatically determined in Matlab (Mathworks, Natick, MA) using a custom algorithm that combined ground reaction force and kinematic event detection methods described by Zeni et al. (2008). COM location was calculated using an eight-segment model consisting of the foot, shank, thigh, pelvis, and trunk (Winter, 2009). COM sway was calculated as the range of COM movement in the frontal plane over each gait cycle. Spatial gait parameters were calculated to characterize locomotor adaptations in response to the external COM perturbations. Step width was calculated on step-by-step basis as the frontal plane distance between the COP at mid-stance between the current and previous steps. This measure was used to quantify changes in the base of support over the course of the walking trial. The ratio of step width to COM sway was used to normalize the base of support to the amount of COM movement, and provided insight into dynamic balance control strategy. Foot placement locations for each limb were normalized to COM location at heel strike (Balasubramanian et al., 2010) to quantify the control of foot placement. Since walking speed was constrained by treadmill belt speed, cadence was calculated for each testing block to quantify temporal gait changes. Forces from the instrumented handrail were examined to evaluate whether the handrail was used to aid in maintaining balance during the perturbations. Handrail forces were first low pass filtered at 10Hz using a 4^{th} order, zero phase, Butterworth filter, and the mean force during stance and swing phase was calculated for each gait cycle.

Prior to statistical analysis, the average response within each testing block was calculated from all steps within the block. Separate repeated measures ANOVAs were used to characterize locomotor changes in response to applied perturbations both within and between the stroke and control groups, as well as differences between the testing blocks. One ANOVA was used to evaluate the impact of different force magnitudes of the perturbations accentuating COM movement. A second ANOVA was used to compare the effect of the augment and resist perturbation types, and use of the head-mounted laser. When the data was not spherical, a Greenhouse-Geisser correction was used for the within-subject effects. Post-hoc analyses were carried out for significant factors using a Bonferroni correction to account for multiple comparisons. Statistical analyses were conduced using SPSS 20.0 (IMB, Armonk, NY), and significance was accepted for $p<0.05$.

5.3 RESULTS

5.3.1 COM Sway

Changes in COM sway across the testing blocks are displayed in Figure 5-2 for both the stroke and control groups. In general, stroke participants walked with larger amounts of COM sway compared to controls (Group, $p=0.008$). Despite larger amounts of sway, the cable-driven system delivered consistent COM perturbations between groups with no significant interactions observed between group and perturbation type ($p=0.192$) or force level ($p=0.608$). The type of perturbation had a significant effect upon COM motion with the augment perturbations increasing COM movement, and the resist perturbation reducing COM movement during the perturbation block (Type, $p<0.001$; Type*Block, $p<0.001$). Changes in COM movement were isolated to the perturbation block only, with no significant differences observed between the pre and post perturbation blocks. Increasing the force of the assisting perturbations also increased the COM sway during the perturbation blocks (Force, $p=0.003$; Force*Block, $p<0.001$). The head mounted laser and stationary target acted to reduce COM movement across all testing blocks (Laser, $p<0.001$) for both groups, but a trend towards a larger decrease with the laser was observed in the stroke group (Group*Laser, $p=0.093$).

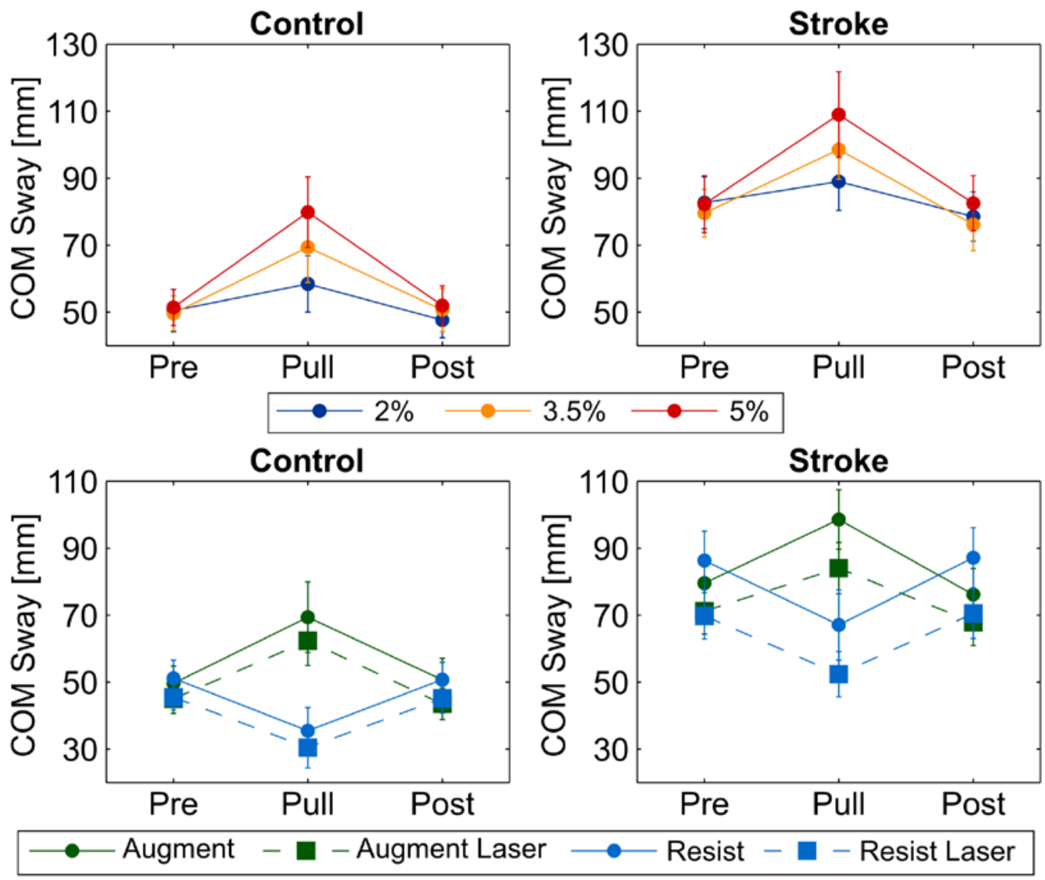

Figure 5-2: Group COM Sway. Group average (± standard error) COM sway in response to different force magnitudes of assisting perturbations (Upper), and effects of laser and perturbation type (Lower). Augmenting perturbations increased COM sway, while resisting perturbations reduced COM sway. Changes in COM movement were consistent between groups, despite larger amounts of baseline sway in the stroke group.

No trends were observed in the time courses of adaptation and de-adaptation to the applied perturbations in either group (Figure 5-3). COM sway rapidly changed when the perturbations were applied, increasing for the accentuating perturbations, or decreasing for the resisting perturbations. These changes remained relatively consistent throughout the perturbation block, indicating that participants did not focus balance control strategy on actively resist the applied perturbations. Removal of the pulls results in a short (one to two) step change in the opposite direction of the adaptation, but the values quickly return to baseline levels.

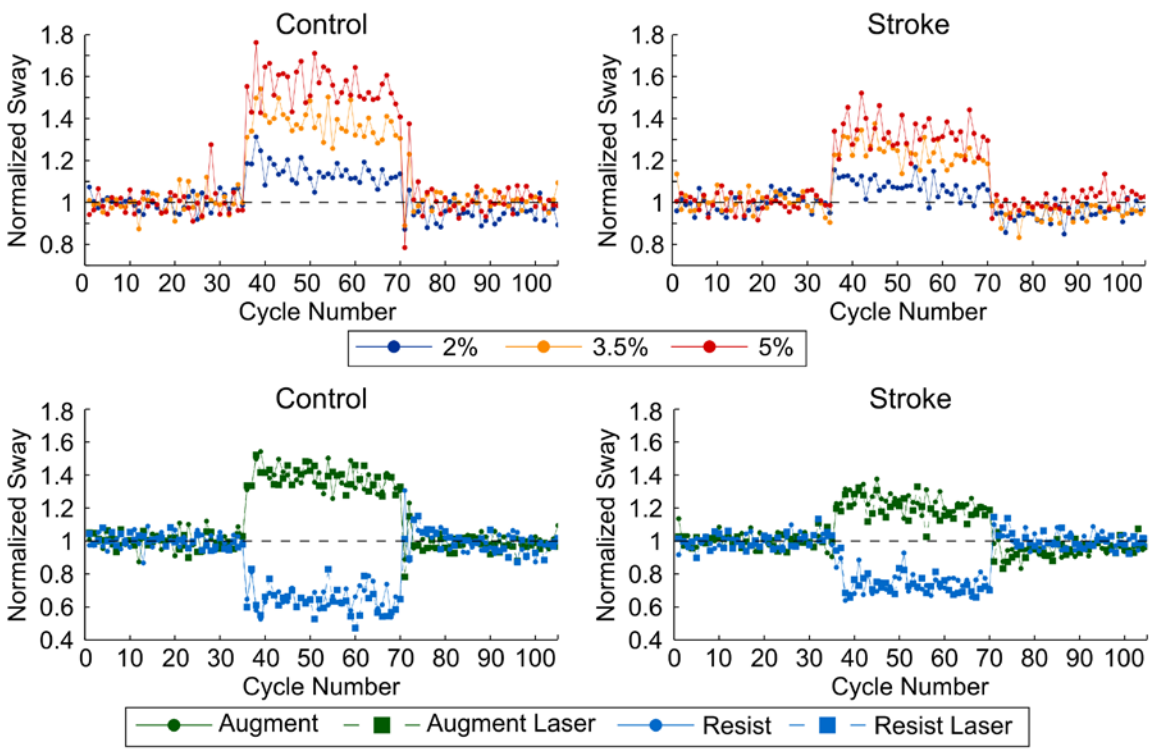

Figure 5-3: COM Sway Temporal Response. Average COM sway across entire trial for each group. Values are normalized to average sway of pre-pull block. COM sway quickly changes in both groups when pulls are applied (cycles 36-70), and quickly return to baseline levels when the pulls were removed.

5.3.2 Step Width

The effects of perturbation type and force level for the stroke and control groups are shown in Figure 5-4. The perturbation type had a significant effect upon step width (Type, $p=0.008$), increasing step width during the augment perturbations, and decreasing step width during the resist perturbations. Step width was only altered during the perturbation block (Type*Block, $p<0.001$). In addition baseline differences in step width (Group, $p=0.019$), a significant interaction between group and perturbation type was observed (Group*Type, $p=0.048$; Group*Type*Block, $p=0.003$). This interaction effect is likely due to the smaller increase in step width when the accentuating perturbations were applied (22.6±7.3% control, 5.5±2.9% stroke), and smaller reduction in step width

(-17.8±4.8% control, -7.6±3.7% stroke) for the resisting perturbations. Larger force magnitudes for the accentuating perturbations produced larger increases in step width (Force*Block, p=0.001). Combining the laser and stationary target had the general effect of reducing step width across all testing blocks (Laser, p=0.004), and a trend towards reducing the magnitude of step width change for the assisting perturbations (Type*Laser*Block, p=0.062).

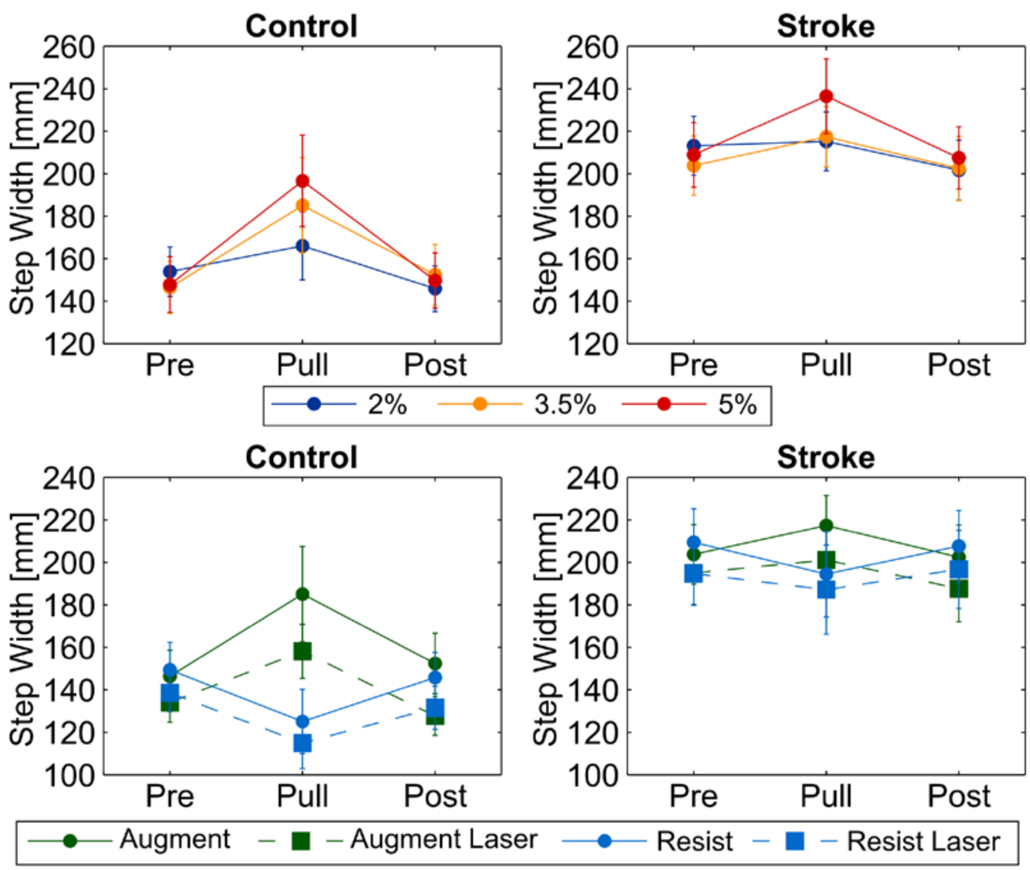

Figure 5-4: Group Step Width. Changes in step width in response to different force magnitudes of augmenting perturbations (Upper), and effects of laser and perturbation type (Lower). Resisting perturbations reduced step widths, while augmenting perturbations increased step width. The stroke group demonstrated smaller changes in step width, compared to controls, when perturbations were applied.

Similar to COM Sway, both groups demonstrated rapid adaptations in step width when the trunk perturbations were applied (Figure 5-5). Removal of the accentuating perturbations produced a slower de-adaptation in step width (5-10 steps), which was similar between groups. When the resisting perturbations were removed, both groups demonstrated a small overshoot, slightly increasing step width before returning to baseline. Interestingly, both groups displayed a post-adaptation effect of lower step widths in post compared to pre block for the 2%BW augment perturbations (paired t-test, stroke p=0.027, control p=0.026).

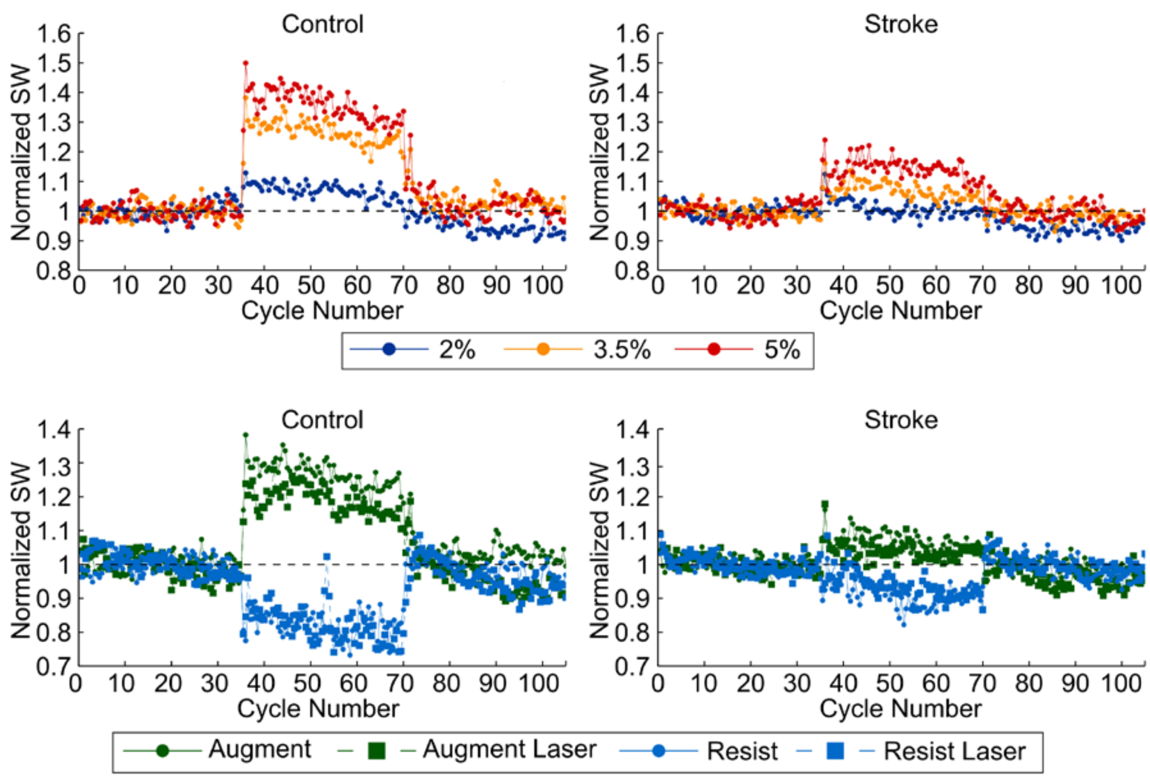

Figure 5-5: Step Width Temporal Response. Ensemble averaged step width across each testing condition. Values were normalized to participant's average step width in pre-perturbation block. Both groups demonstrate fast adaptions in step width to application of perturbation (cycles 36-70), and after-effect of reduced step in response to assisting perturbations delivered at 2%BW.

5.3.3 SW/COM Ratio

Changes in the ratio of step width to COM sway are shown in Figure 5-6. Although the stroke group walked with larger COM movement and step widths, the SW/COM ratio was not significantly different between groups ($p=0.143$). Accentuating perturbations reduced the ratio, while resisting perturbations increased the ratio (Type, $p<0.001$). This effect was only observed during the perturbation block, with no significant differences after the perturbation was removed (Type*Block, $p<0.001$). The laser-targeting task increased the ratio across all trials (Laser, $p=0.001$), and also had a significantly larger impact during the resisting perturbations (Type*Laser, $p=0.002$). As the force of the accentuating perturbations increased, the SW/COM ratio decreased during the perturbation block (Force*Block, $p<0.001$).

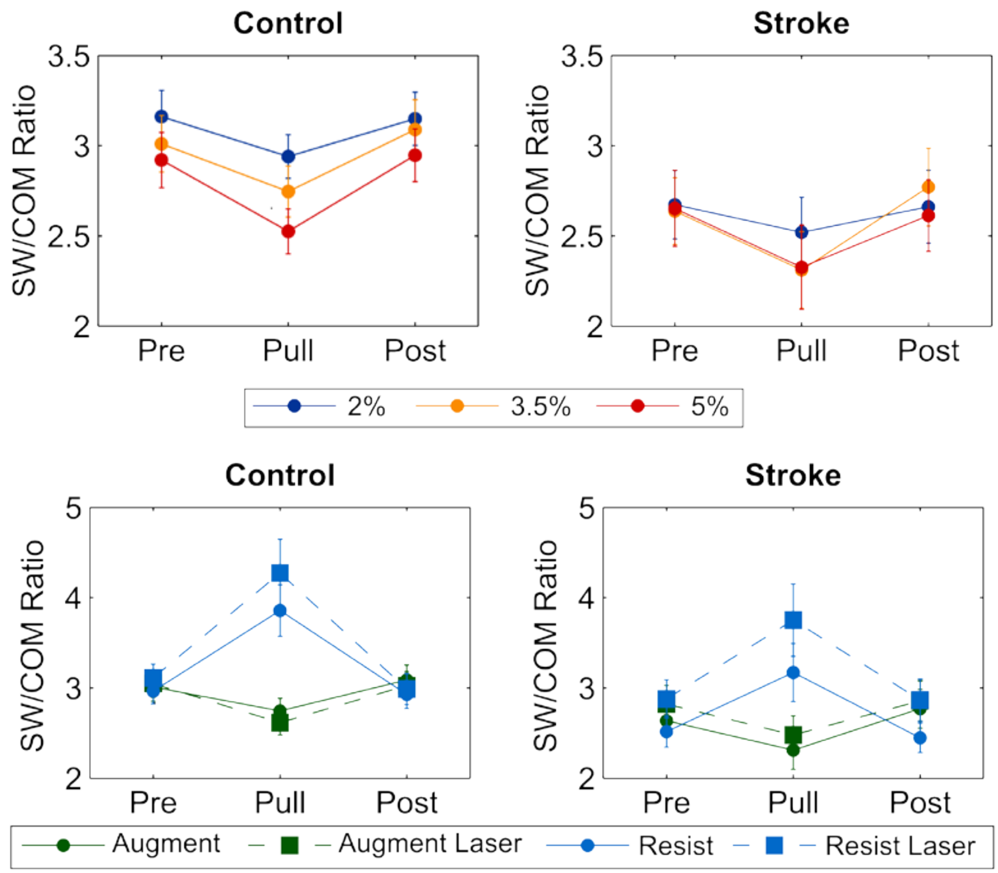

Figure 5-6: SW/COM Ratio. Ratio of step width to COM sway for different augment perturbation magnitudes (Upper) and perturbation types (Lower). On average, both groups had similar SW/COM ratios, but stroke participants demonstrated significantly lower modulation of the ratio in response to the perturbations.

5.3.4 Medial-Lateral Foot Placement

5.3.4.1 Paretic Limb

No significant baseline differences between the placement of the paretic limb and test limb in controls were observed (p=0.222). The two groups responded differently to the different perturbation types (Group*Type*Block, p=0.004), which are shown in Figure 5-7. Controls placed the foot more lateral of the COM during the augment perturbation block, and more medial during the resist perturbation block. However, stroke participants did not alter foot placement of the paretic limb in response to either perturbation type. As the augmenting perturbation force increased, the control group placed the limb more laterally during the perturbation block (Force*Block, $p<0.001$). In contrast the stroke group demonstrated small to no change in lateral foot placement at the 2% and 3.5% perturbation magnitudes, but did increase lateral foot placement for the 5% force (Group*Force*Block, $p=0.13$).

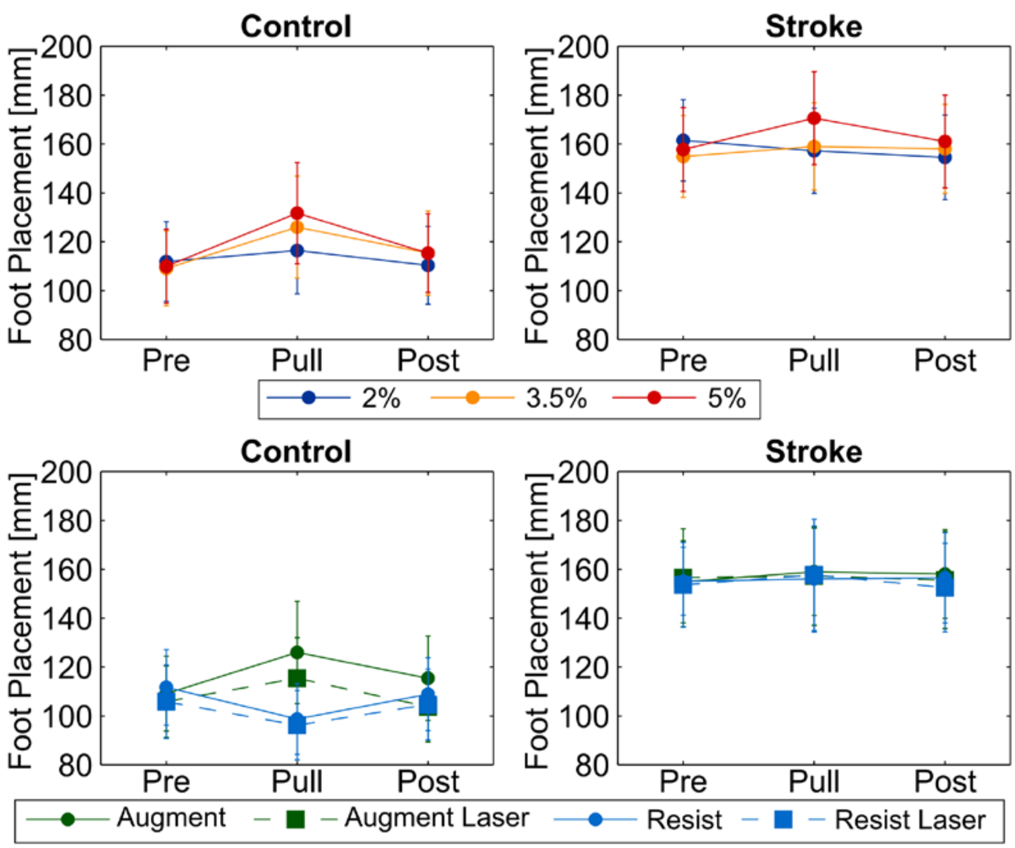

Figure 5-7: Paretic Limb Foot Placement. Placement of the paretic limb relative to COM position at heel strike in response to perturbations. Individuals with chronic stroke demonstrated less modulation of paretic limb foot placement location in response to different perturbation types (Lower), and different perturbation forces (Upper).

5.3.4.2 Non-Paretic

The effects of perturbation type and force magnitude are shown for both groups in Figure 5-8. Stroke survivors placed their non-paretic limb more lateral to the COM compared to the non-test limb of controls (Group, p=0.047). Foot placement was more medial with the resisting perturbations, and more lateral with the accentuating perturbations, but only during the perturbation block (Type*Block, p<0.001). The laser-targeting task had the general effect of reducing lateral foot placement across all blocks (Laser, p=0.001). There was also a trend towards a group interaction with the perturbation type (Group*Type, p=0.092), driven by the stroke group not increasing lateral foot placement during the augment perturbations. This trend was also observed when comparing the force levels for the augmenting perturbations. In general, lateral foot placement increased as the perturbation force increased (Force*Block, p=0.036), but the stroke group only slightly increased in lateral foot placement at the 5%BW force (Group*Block, p=0.09).

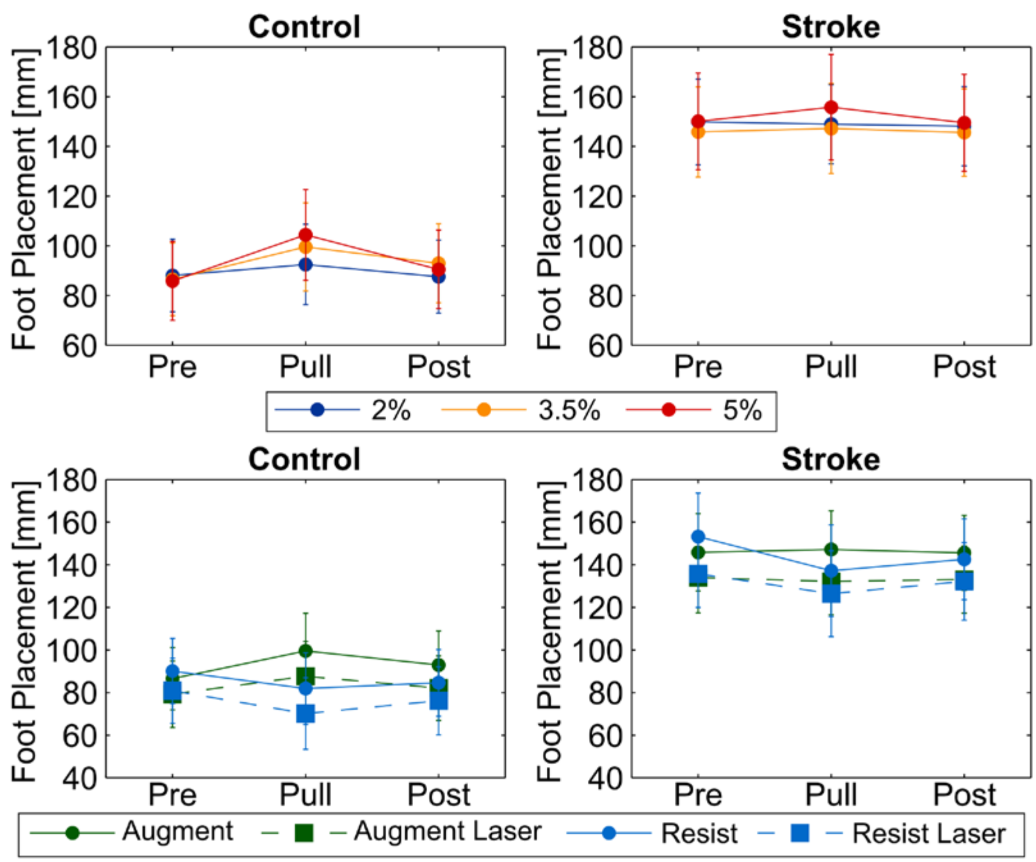

Figure 5-8: Non-paretic Foot Placement. Changes in non-paretic (non-test leg in controls) foot placement relative to COM location at heel strike for different perturbation amplitudes (Upper) and types (Lower). The stroke group did not alter non-paretic foot placement in response to the augmenting perturbations.

5.3.5 Cadence

The accentuating perturbations produced a significant increase in cadence during the perturbation block for both groups, with larger changes observed as the force magnitude increased (Block, $p<0.001$; Block*Force, $p=0.001$). Resisting perturbations tended to slightly reduced cadence when applied during walking (Type, $p=0.001$; Type*Block, $p<0.001$). Addition of the laser and stationary targeting task had the general effect of increasing cadence across all testing blocks (Laser, $p=0.003$), with the responses mainly driven by the stroke group (Laser*Group, $p=0.066$). Across all testing conditions, stroke survivors walked with significantly lower cadences compared to the control group (Group, p=0.021).

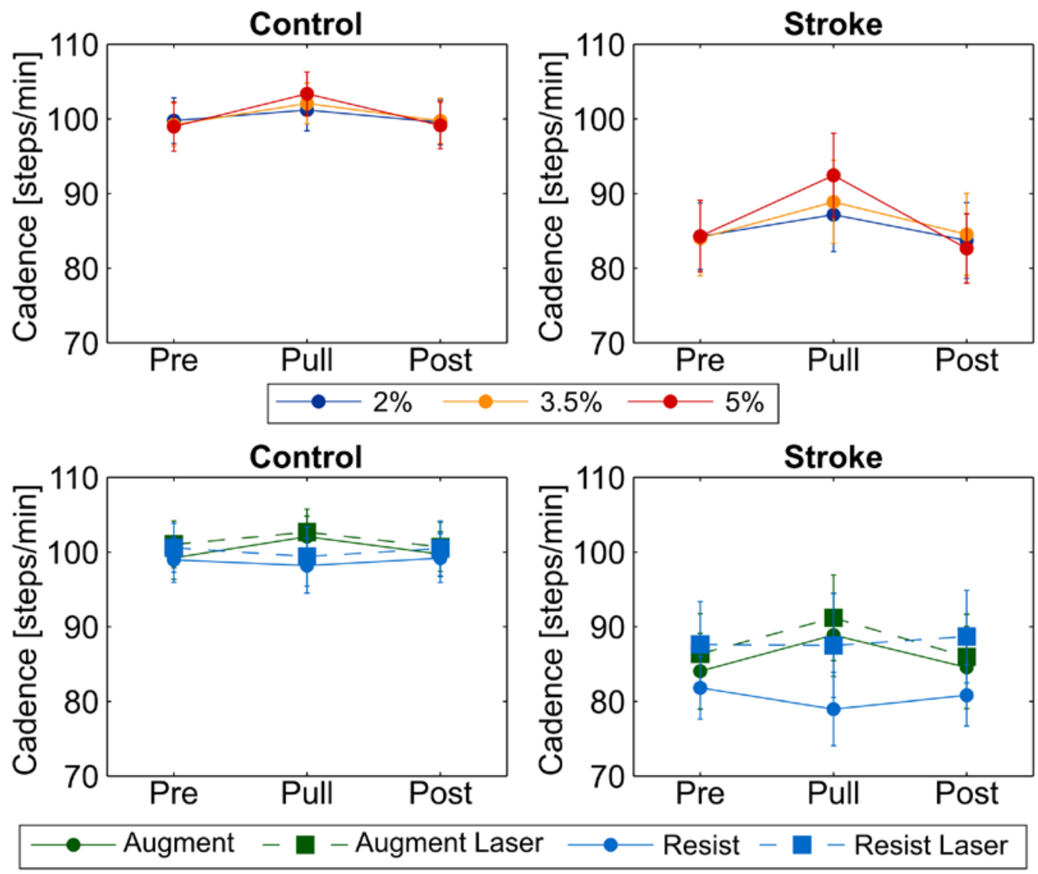

Figure 5-9: Cadence. Average cadence for different augment perturbation magnitudes (upper), and different perturbation types (lower). The augment perturbations increased cadence in both groups, while the resist perturbations reduced cadence.

5.3.6 Handrail Hold Forces

We observed a modulation of the handrail forces when the perturbations were applied. This modulation occurred mainly in the medial-lateral direction, while the non-paretic (non-test) leg was in swing, as shown in Figure 5-10. In general, the stroke group demonstrated larger lateral forces during swing, potentially to help with balance control during walking (Group, $p=0.039$). The medial lateral handle forces during the perturbation block were in the same direction as the perturbation. Accentuating perturbations caused the mean force during swing to become more medial, while the resisting perturbations increased the lateral forces (Type*Block, $p<0.001$). There was an interaction effect between the perturbation type and group ($p=0.046$) due to the mean force remaining lateral in the stroke group, but becoming medial for the control group when the accentuating perturbations were applied. As the force level of the accentuating perturbations increased, the mean handle force during swing became more medial (Force*Block, $p=0.001$).

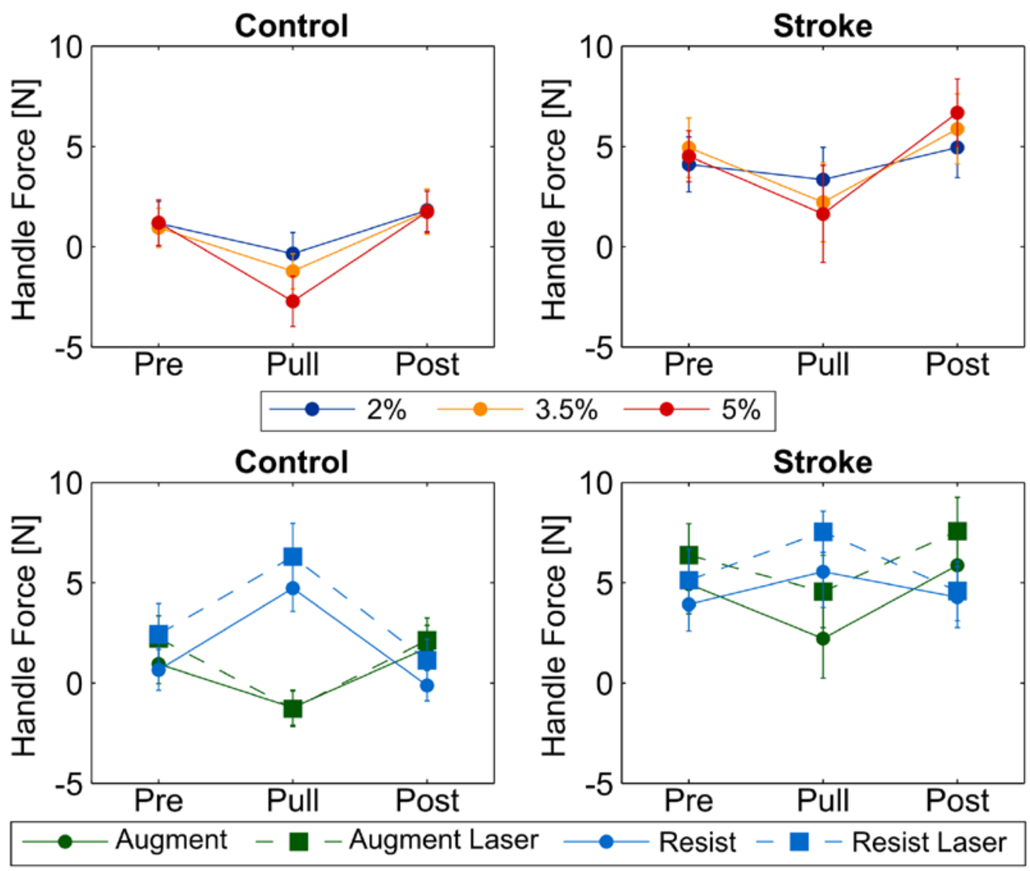

Figure 5-10: Mean Lateral Handle Force During Swing. Changes in average medial-lateral handle force during swing phase of non-paretic (non-test) limb, with lateral forces in positive direction. Overall, the stroke group had higher lateral forces compared to controls. Handle forces modulated during the perturbation block in the direction of the applied force.

5.4 DISCUSSION

The cable-drive device used in this study was able to provide consistent frontal plane perturbations of trunk motion across the stroke and control groups. Overall, the chronic stroke survivors in this study were able to adapt to external perturbations without falling. Locomotor adaptations made in response to the perturbations were focused on foot placement adjustments made to modify the base of support to the changing COM movement. However, foot placement adjustments made by the stroke group were smaller than the control group. This finding supports our hypothesis that deficits in foot placement control alter dynamic balance control strategy post-stroke.

Dynamic balance control strategy post-stroke was characterized by evaluating locomotor adaptations in response to external perturbations of COM movement. In this study, we constructed a cable-driven device, similar to that of Wu et al. (2011), to apply external perturbations of trunk movement. These applied perturbations were timed to the individual’s walking pattern, phasing with the fontal plane COM velocity. The augment perturbations increased COM sway, while the resist perturbations reduced COM sway in both groups. COM movement in the frontal plane was only significantly altered when the perturbations were applied, with no significant differences observed between the pre and post perturbation blocks. Since these effects are only present in the perturbation block, locomotor differences between the pre and post blocks would provide insight into any potential after-effects of the perturbation. Despite larger amounts of baseline COM movement in the stroke group, there was no significant interaction between group and either perturbation type or magnitude. The similarity of the perturbation magnitude between groups enables the characterization of changes in dynamic balance control strategy by directly comparing the locomotor adaptations between groups.

Step width was modified by both groups to adjust their base of support to the perturbed movement of the COM in the fontal plane, but the magnitude of these adaptations were smaller in the stroke group. Individuals increased step width in response to perturbations accentuating COM movement, and decreased step width to perturbations resisting COM movement. Similar results were observed when a short lateral perturbation was delivered to the trunk in young controls, with foot placement location kept at a constant distance outside the COM location even with the perturbations (Hof et al., 2010). Coupling between the extent of COM movement and step width were also observed when step width decreased in young and elderly participants in response to the application of forces to stabilize trunk movement during walking (Dean et al., 2007). Although the stroke group adjusted step width in a similar manner as controls, they made significantly smaller adjustments when the perturbations were applied. Since the change in COM movement was similar between groups, the smaller step width increases for the assisting perturbations could be due to reduced balance control, and may partially explain the increased fall risk post-stroke. Additionally, these reduced changes may also represent a change in dynamic balance control strategy post-stroke.

In addition to a lateral foot placement control strategy, dynamic balance can also be maintained by directly controlling COM movement during walking. If stroke survivors were attempting to directly control COM movement, we would anticipate smaller increases in COM sway for the assist perturbations. Smaller percent changes observed in the stroke group are likely attributed to larger amounts of baseline COM sway, because there was no significant interaction effect of group. However, it is possible that stoke subjects are making locomotor adaptations intended to control COM sway, but are not detected by our measures. Two potential sources of COM control observed in this study were changes in walking cadence, and increased handrail forces. The handrail could be used to generate forces to directly oppose the perturbation forces, while increasing cadence would act to reduce COM sway by reducing both duration and extent of COM movement. We observed both changes in cadence and lateral handrail forces in both groups when the perturbations were applied. Since no significant group interaction effects were observed for changes in COM sway during the perturbation block, we do not believe the stroke subjects were solely focused on controlling COM control movement. However, given larger percent changes in both handrail hold force and cadence in the stroke group, it is likely that stroke survivors are placing a greater emphasis on COM control to maintain balance during walking.

In addition to a reduced step width in responses to the external perturbations, stroke survivors demonstrated reduced foot placement modulation. Specifically, placement of the paretic limb relative to the COM did not change in response to the accentuating or resisting perturbations at 3.5%BW. The control group increased lateral foot placement of the test limb during the accentuating perturbations, and reduced lateral foot placement during the resisting perturbations. Additionally, control participants placed both feet more lateral as the magnitude of the perturbation forces increased. Stroke survivors showed a similar increase in lateral foot placement for the 5%BW perturbation, but the change in paretic foot placement was much smaller for the 3.5% and 2%BW force levels. A similar trend was also observed in non-paretic foot placement, with little modulation at the 2% and 3.5%BW force levels. The lack of paretic foot placement modulation presents a potential source of the increased fall risk post-stroke, since the base of support on the paretic side is not accommodating for the increased COM movement. However, increased lateral placement of the paretic limb for 5%BW accentuating perturbations indicates that stroke survivors retain the ability to make lateral foot placement corrections. Changes in their balance control strategy are likely related to both difficulties executing frontal plane step corrections (Nonnekes et al., 2010), and sensing the increased trunk movement (Ryerson et al., 2008). Augmented visual feedback of body movement had the general effect of reducing fontal plane COM movement and step width across the entire trial, but did not appear to increase locomotor adaptations to the perturbations in either group. These reductions resulted in a net increase in the SW/COM ratio with the targeting task. Additionally, SW/COM ratio increased more when the stationary targeting task was combined with the resisting trunk perturbations. The visual feedback signal had a larger impact in the stroke group, with larger increases in the SW/COM ratio compared to controls, and a trend towards larger reductions in COM sway. This group effect is likely due to an increased reliance on visual feedback for balance control in the stroke group (Marigold & Eng, 2006b). The observed improvements in frontal plane control are similar to those observed when feedback of trunk position was provided to young controls during walking (Verhoeff et al., 2009). This stationary targeting task may present a potential tool to improve dynamic balance control post-stroke, since it had the net effect of increasing the base of support relative to COM movement.

The trunk perturbations protocol used in this study also have a potential use as a training tool to improve dynamic balance control post-stroke. Both groups demonstrated an aftereffect in step width during the post perturbation block, but only for the augment perturbation delivered at 2%BW. Additionally, there was also a trend towards reduced COM sway at 2% force level, but the difference was not significant for either group. No significant aftereffects were observed for the 3.5%BW or 5% BW assist perturbations. Reisman et al. (2007) observed post-adaptation aftereffects when a split-belt speed perturbation was used to accentuate baseline asymmetries, resulting in the stroke survivors producing a more symmetric gait pattern when the speed perturbation was removed. The low level perturbations accentuating trunk movement have the potential to induce plastic changes that may be useful to reduce the larger step widths observed in chronic stroke survivors. Further research is necessary to characterize the duration of these aftereffects persist, as well as their impact on both balance control and walking function.

One potential limitation of our analysis of balance control strategy is the presence of the handrail hold throughout the walking trial. A light touch cue has been shown to stabilize motion of the pelvis in the frontal plane during walking post-stroke (Boonsinsukh et al., 2009). Furthermore, forces produced at the handrail can significantly contribute to the control of frontal plane COM movements (Tung *et al.*, 2011). During treadmill walking, holding onto the handrail is often necessary to ensure patient safety and comfort, especially for individuals with functional deficits. The effects of handrail hold were minimized in this study by limiting the handrail hold to the non-paretic (non-test) hand and placing the handle in front of the individual. Additionally, both stroke and control participants were to hold onto the handle throughout the duration of the experiment. The handle was instrumented with a six-axis load cell to quantify forces applied by the individual throughout the walking trial. In this experimental setup, we observed differences between groups and with the perturbations for the average medial-lateral force during swing of the non-paretic (non-test) leg. In general, stroke survivors used the handrail during walking more than controls, walking with larger lateral forces during swing. This lateral force would help provide stability when in single limb stance on the paretic leg. Both groups showed similar trends when perturbations were applied, with the assisting perturbations resulting in increased medial forces, while resisting forces increased the lateral forces. The modest change in forces during the perturbation block, demonstrates that the stroke group did not primarily generate forces at the handrail to counter the trunk perturbations.

The results of this study demonstrate that, similar to age-matched controls, stroke survivors were able to adjust their gait pattern in order to adapt to frontal plane trunk perturbations. However, smaller step width changes, and a lack of lateral foot placement modulation in the stroke group, demonstrate changes in dynamic balance control post-stroke. Providing additional feedback of body movement with the head mounted laser helped to improve foot placement and COM control during walking. Inclusion of the stationary targeting task into a rehabilitation protocol might help to further improve walking function by improving dynamic balance control. Additionally, post adaptation aftereffects demonstrate the potential use of low-level accentuating perturbations as a training tool to improve dynamic balance control post-stroke. Taken together the results of this study provide further insight into the changes in dynamic balance control strategy post-stroke.

CHAPTER 6: APPLICATIONS AND FUTURE DIRECTIONS

6.1 SUMMARY OF RESULTS

The results of this study provide further insight into stroke-related changes in dynamic balance control strategy during walking. Overall, stroke survivors walked with larger amounts of frontal plane COM movement, as well as larger step widths compared to age-matched control subjects. Despite these larger baseline differences, the ratio of step width to COM sway was consistent between groups. The similarity of the ratio between groups indicates that simply choosing a wider step width does not produce a safer walking pattern for the stroke group, since the movement of the COM also increases. Stroke survivors placed their paretic limb more lateral to the COM compared to the non-paretic limb, as well as both legs for the control group. This asymmetric foot placement would widen the base of support on the paretic side, helping to maintain balance during walking. However, we observed no changes in paretic foot placement relative to the COM when trunk movement was externally perturbed. This lack of paretic foot placement modulation would limit the ability of the individual to maintain balance when COM movement increases, which may partially explain increased fall prevalence post-stroke. In addition to characterizing dynamic balance control in chronic stroke survivors, we also evaluated the impact of augmented sensory feedback upon this control. Augmented sensory feedback improved paretic foot placement control during a targeted stepping task, and COM control during treadmill walking. These improvements were observed mainly in the frontal plane, and may help to improve dynamic balance control for chronic stroke survivors. Specifically, improved paretic foot placement control would enable stroke survivors to utilize a lateral foot placement control strategy, while improved COM control might reduce the need for wider step widths during walking. These results demonstrate that augmented sensory feedback signals could be used to improve balance control, and thus walking function for chronic stroke survivors.

6.2 IMPLICATIONS FOR REHABILITATION

The results of this dissertation provide information that can be used to direct rehabilitation techniques aimed at improve walking function in chronic stroke survivors by targeting specific deficits in dynamic balance control. Wider step widths have been observed in chronic stroke survivors compared to age matched controls walking at the same speeds (Chen *et al.*, 2005*b*), and is typically associated with stroke survivors selecting a more conservative walking pattern to maintain balance. However, increased COM movement during walking results in a similar step width/COM ratio between the stroke and control groups, which suggests potential underlying changes in dynamic balance control strategy. Rehabilitation techniques focused upon improving dynamic balance control by targeting medial-lateral control of paretic foot placement and/or frontal plane COM movement may increase walking function in chronic stroke survivors.

Imparied foot placement control post-stroke limits the effectiveness of a lateral foot placement control strategy in maintaining dynamic balance during walking. Deficits in medial adjustments of paretic foot placement are observed even when balance constraints are removed (Nonnekes et al., 2010). This deficit was observed in our studies as a lack of foot placement modulation in response to external perturbations of trunk motion. Not adjusting paretic foot placement to the task demands would lead to an increase in the relative fall risk. Therefore, improving paretic foot placement control might provide a means to reduced the incidence of falls during walking post-stroke. In the first aim of the dissertation, we demonstrated that somatosensory stimulation of the paretic foot/ankle improved paretic limb control and reduced medial-lateral targeting error during a stepping task. The inclusion of somatosensory stimulation into traditional rehabilitation techniques could help to improve paretic foot placement control, and thus walking function post-stroke. Further examination into the effects of somatosensory stimulation of the paretic foot/ankle during continuous walking needs to be completed, before incorporating this technique into rehabilitation protocols.

In addition to reduced foot placement control, the results of this dissertation also suggest deficits in the control of COM movement during walking post-stroke, with stroke survivors walking with larger amounts of COM sway compared to neurologically intact individuals. Feedback of body movement from the head mounted laser had the general effect of reducing COM sway in both groups, with larger reductions observed in the stroke group. This reduced sway could help to reduce energy expenditure associated with larger amounts of COM sway and larger step widths during walking (Donelan et al., 2001; 2004). Providing this additional sensory feedback source may have a larger effect during overground walking, when the lateral motion of the individual is not constrained by the size of the treadmill surface.

6.3 FUTURE STUDIES

One main limitation of the work presented in this dissertation is a lack of a direct metric of overall walking function in chronic stroke survivors. Typically the individual's comfortable or maximum overground walking speed is used to characterize walking function post-stroke (Lord et al., 2004), but walking speed remained constant throughout the experiment due to testing walking function on the treadmill. Both neurologically intact individuals and chronic stroke survivors were able to successfully complete the walking tasks without falling during either challenging treadmill walking conditions (Chapter 3) or perturbations of trunk motion (Chapter 5). However, stroke survivors utilized different walking patterns to maintain balance, mainly larger step widths and greater COM sway compared to controls. These differences may contribute to increased energy expenditure during walking post-stroke (Waters & Mulroy, 1999; Donelan *et al.*, 2001), which has been linked to an increased fall risk (Carver et al., 2011) and reduced walking function (Michael et al., 2005). Additionally, visual feedback of body movement reduced COM movement, decreasing the metabolic cost of walking (Donelan et al., 2004). Incorporating the measurement of the metabolic rate during the walking trials would provide additional insight into the overall impact of locomotor changes on walking function. It is likely that stroke survivors were expending more energy during the perturbation trials compared to controls, which would provide further insight into the consequences of altered dynamic balance control strategy post-stroke.

Further investigation is needed into how augmented sensory feedback impacts dynamic balance control and walking function post-stroke. In Chapter 2, we demonstrated that somatosensory stimulation of the paretic foot could be used to improve paretic foot placement control during a targeted stepping task. Providing somatosensory stimulation during external perturbations of trunk motion might facilitate modulation of paretic foot placement, which was not observed in the stroke group (Chapter 5). It is important to note that we did not observe changes in paretic foot placement control when somatosensory stimulation was applied during continuous treadmill walking. However, only a small number of stroke survivors were tested (n=6) with and without the stimulation during normal and reduced visual feedback treadmill walking. It is possible that the increased demands on lateral foot placement control due to the cable-driven perturbations of trunk motion may facilitate a greater impact of the stimulation.

The custom cable-driven device used in Chapters 4 and 5 might be useful as a training tool to improve walking function in chronic stroke survivors. Adaptation to differences in belt speed between the legs has been used to produce a more symmetric walking pattern in stroke survivors during the de-adaptation phase (Reisman et al., 2007; 2009). It is possible that perturbations assisting trunk movement may result in reduced amounts of COM movement and smaller step widths when the perturbations are removed. In the present study (Chapter 5) we observed an aftereffect of reduced step widths between the pre and post perturbation blocks. This aftereffect was observed in both stroke survivors and neurologically intact participants, but only for assisting perturbations delivered at 2% of the participant’s body weight. However, we did not observe this aftereffect of reduced step width when trunk perturbations were applied to healthy young participants (Chapter 4), but there were differences in the experimental design of the two studies. The lowest perturbation force was larger for the young participants (2.5% BW vs. 2%BW), the perturbations were provided over more steps

(100 vs. 35), and the application of catch perturbation catch trials in the post perturbation block in the young control study. Further study is necessary to characterize the factors contributing these aftereffects, to optimize the step width reductions in chronic stroke survivors. Additionally, it is necessary to evaluate whether step width aftereffects have the potential to transfer to overground walking in chronic stroke survivors, similar to the improved symmetry observed after split-belt adaptations (Reisman et al., 2009).

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Appendix A: DESCRIPTION OF CABLE DRIVEN DEVICE

A.1 SERVOMOTOR SYSTEM

Design and construction of the cable driven device was based upon the device description provided by Wu *et al.* (2011), and consists of a servomotor connected to a cable spindle. The commercial servomotor and drive system enables the user to set a desired torque output, which is then maintained by the drive electronics at a fast rate (< 1μs for current loop). A flexible coupling joins the servomotor to a custom cable spindle (r = 0.045m), translating the motor torque set point into a desired cable tension. The motor and spindle are mounted to a custom base plate, which can be positioned at any point along the 80/20 support frame. Cable spindle, motor mounting plates, and the motor base plate were machined out of aluminum (6061 alloy).

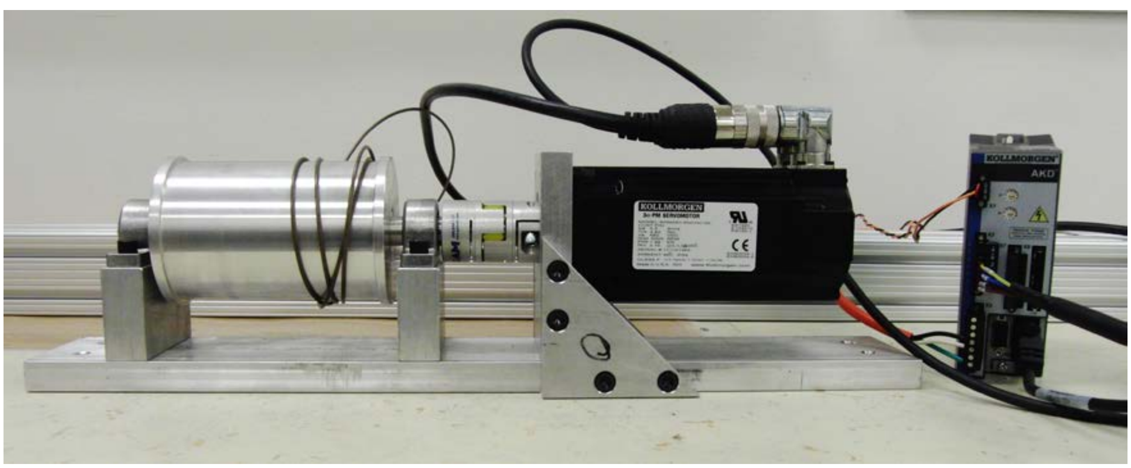


Figure A-1: Assembled Servomotor System. Commercial motor was mounted on custom aluminum base, and connected to cable spool by a flexible coupling.

Although only two motors were used to provide perturbations for the experiments described in this dissertation, the cable-driven system was designed around a total of four servomotors. The rotational inertia of the motor and cable spindle limited both the radius of the cable spindle and the size of the motor that could be used for this project. As the rotational inertia of the system increases, a larger baseline torque must be chosen to enable the system to overcome the inertia and spin freely. This baseline torque would be applied to the participant when connected to the cable, and it was limited to reduce the impact of the device during the no perturbation walking conditions. Different maximum force outputs of the system were provided by the choice of two different servomotors, AKM33H and AKM43H. The AKM33H motor can produce a peak cable tension of 190N, while the AKM43H produces a peak cable tension of 310N. We chose to use the 33H series motors for the trunk perturbations, since the lower motor inertia reduced the baseline cable tension to approximately 6N.

Part	Manufacturer	Part Number
33H Servomotor	Kollmorgen	AKM33H-ANCNC-00
43H Servomotor	Kollmorgen	AKM43H-ANCNC-00
Servomotor Drives	Kollmorgen	AKD-B00606-NAAN-0000
33H Coupling	GAM Enterprises Inc.	EKC-25Y-14N-14N
43H Coupling	GAM Enterprises Inc.	EKC-25Y-19N-14N
Spindle Bearings	McMaser Carr	5912K7
Nylon-Coated Wire Cable	McMaser Carr	3459T42
Cable Stop	McMaser Carr	3936T35

Table A-1: Servomotor System Components.

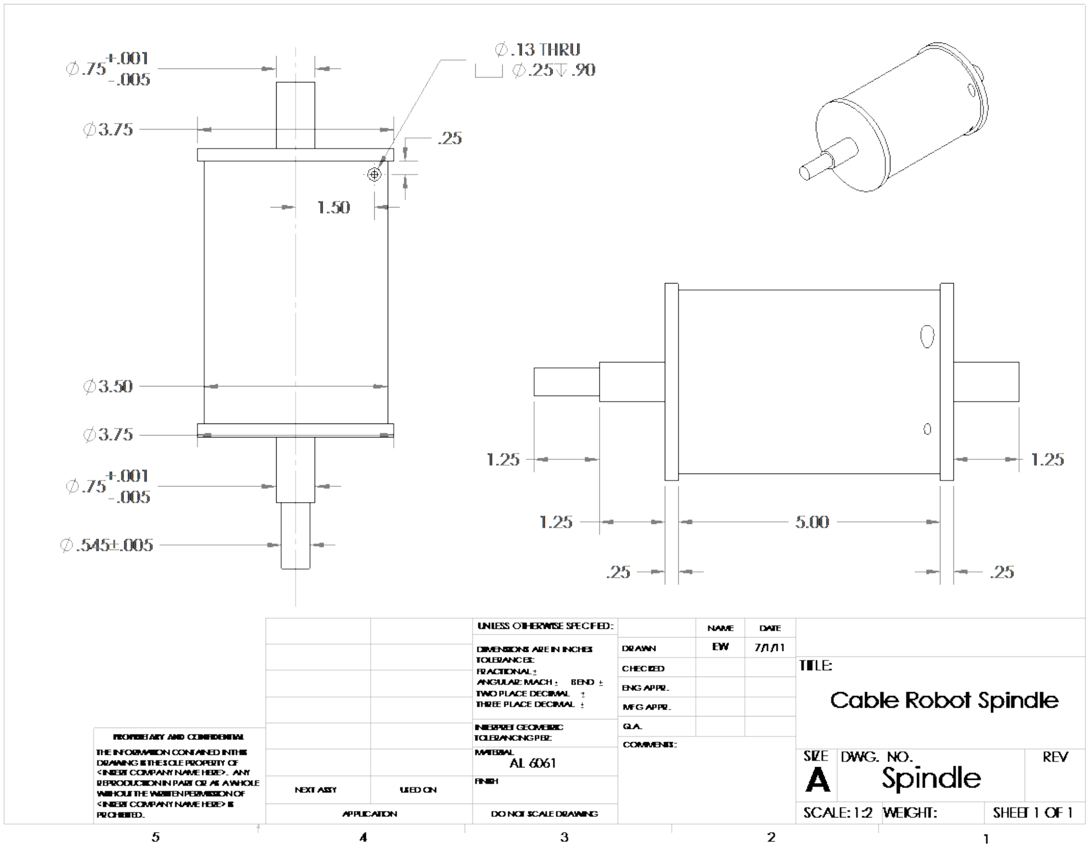


Figure A-2: Cable Spindle.

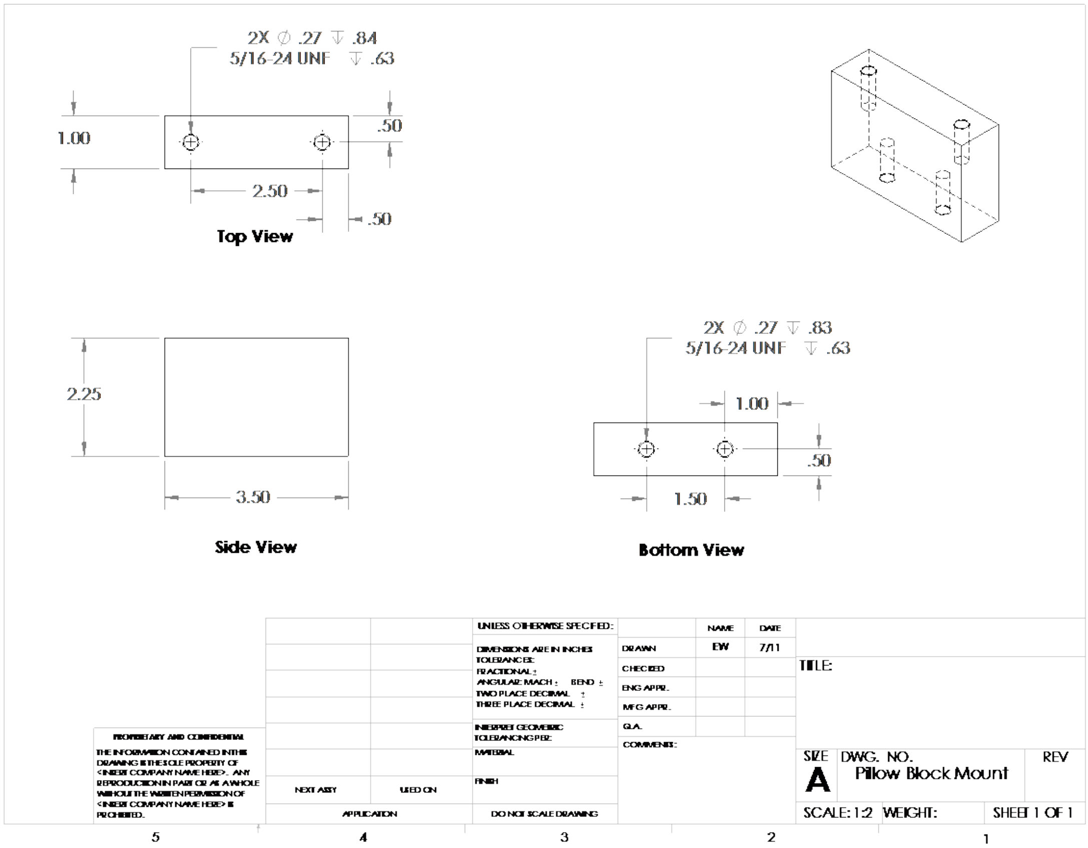
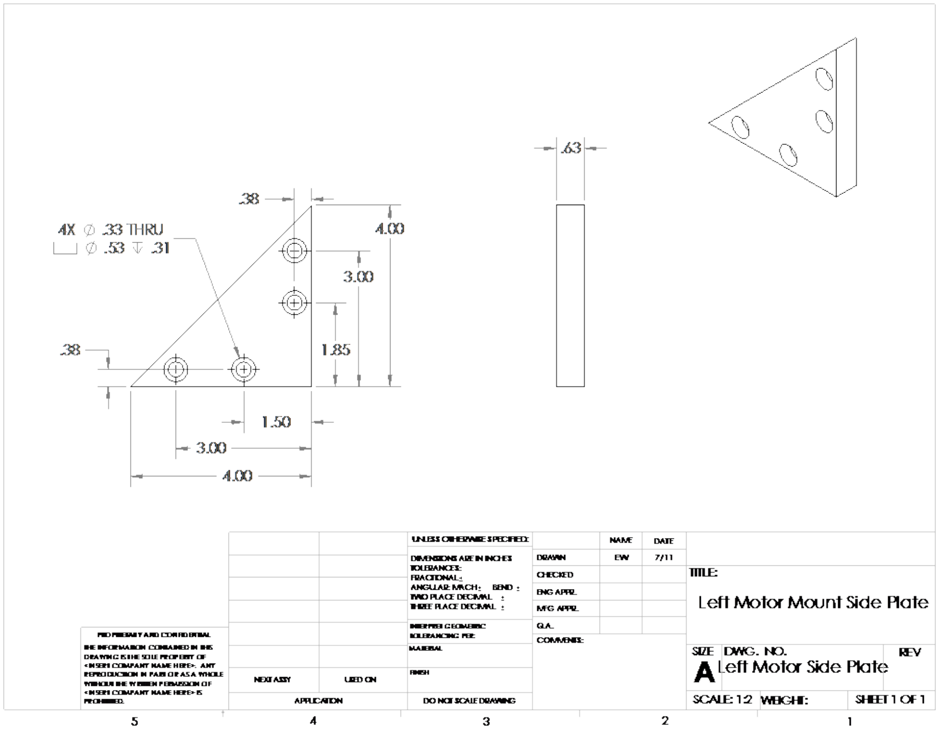


Figure A-3: Mounting Block for Spindle Bearings.



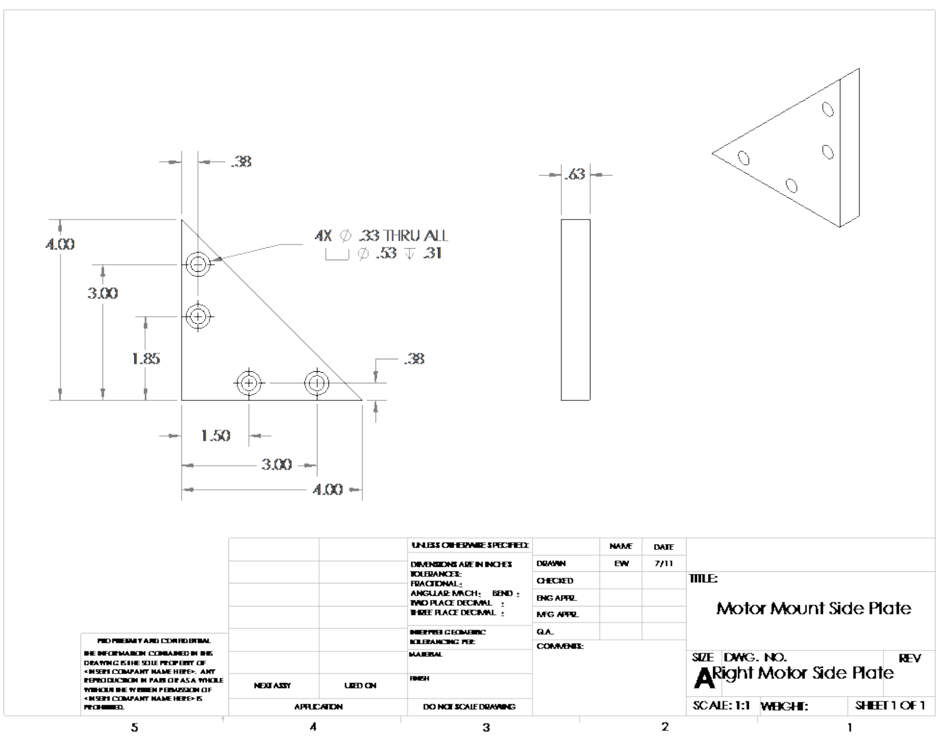
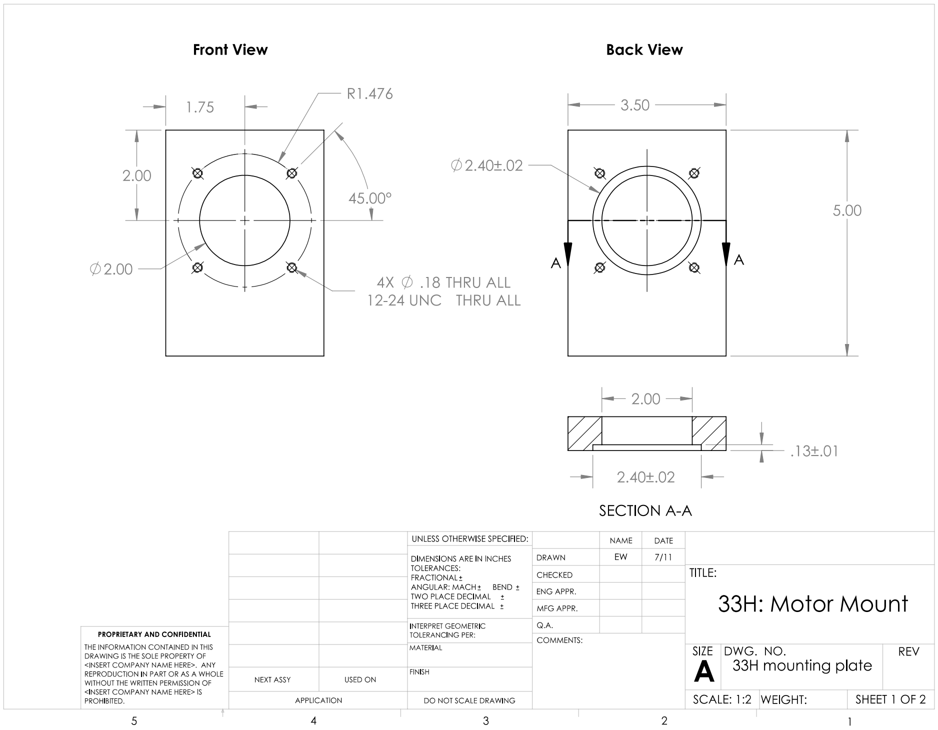


Figure A-4: Motor Mount Side Brackets.



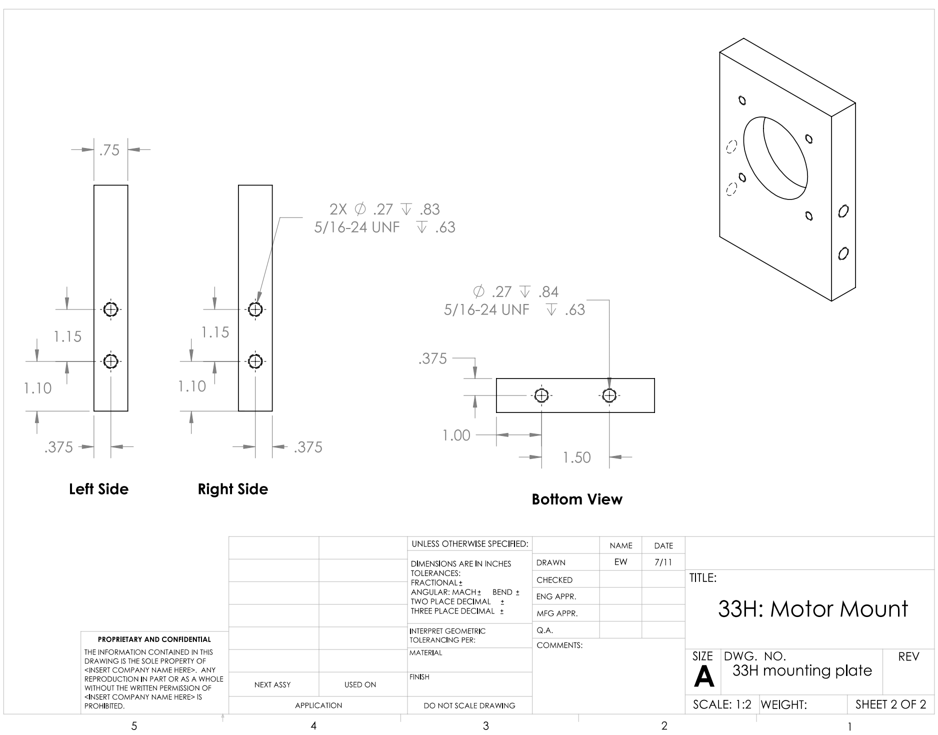


Figure A-5: Motor mounting plate for 33H series servomotor.

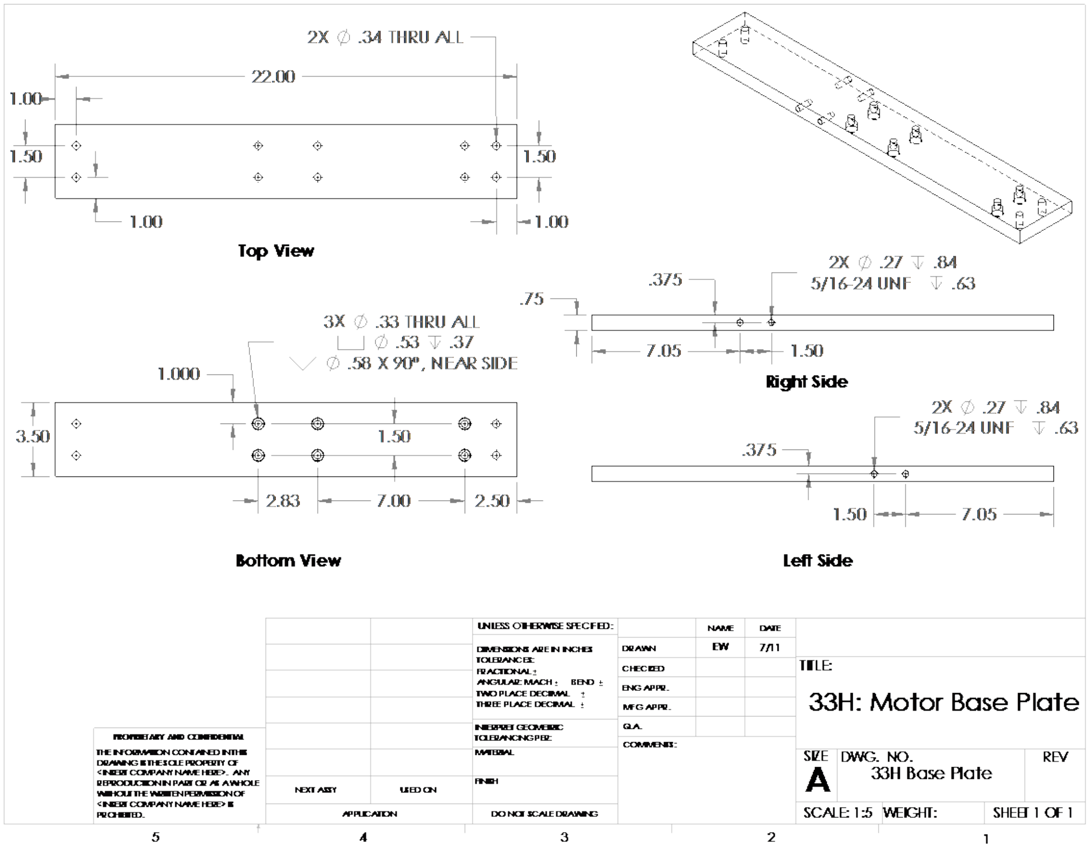
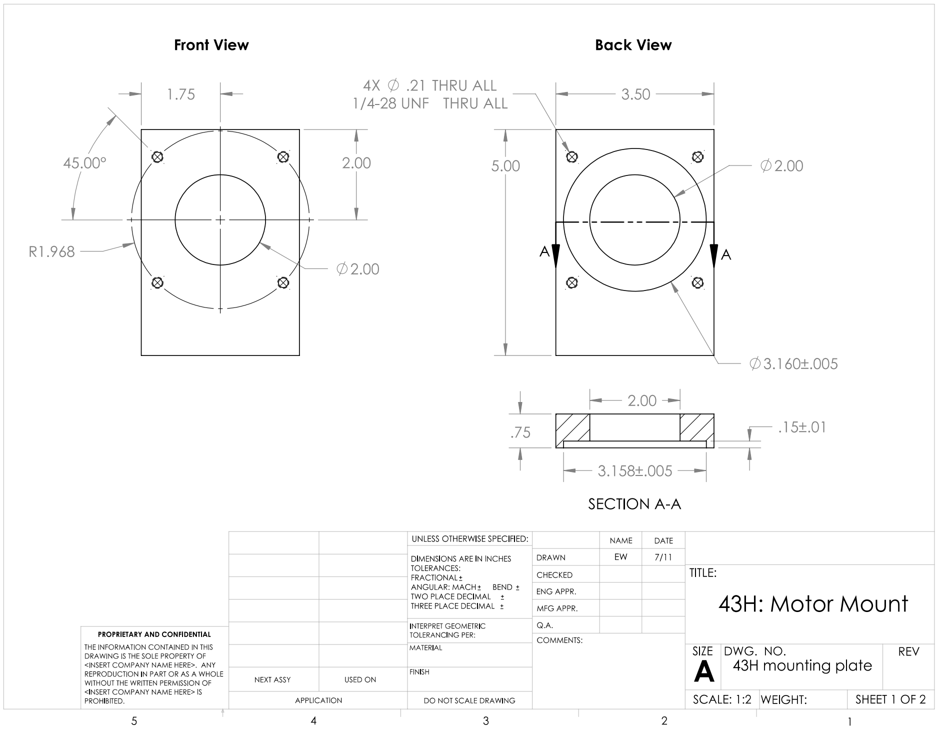


Figure A-6: Base Plate for 33H Series Servomotor.



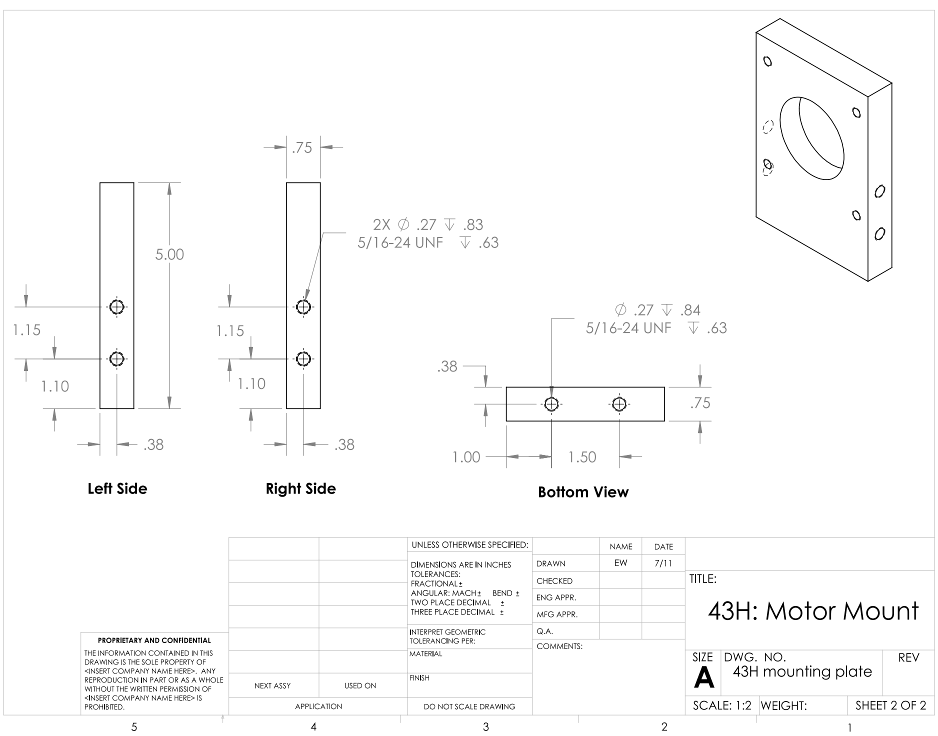


Figure A-7: Mounting Plate for 43H Series Servomotor.

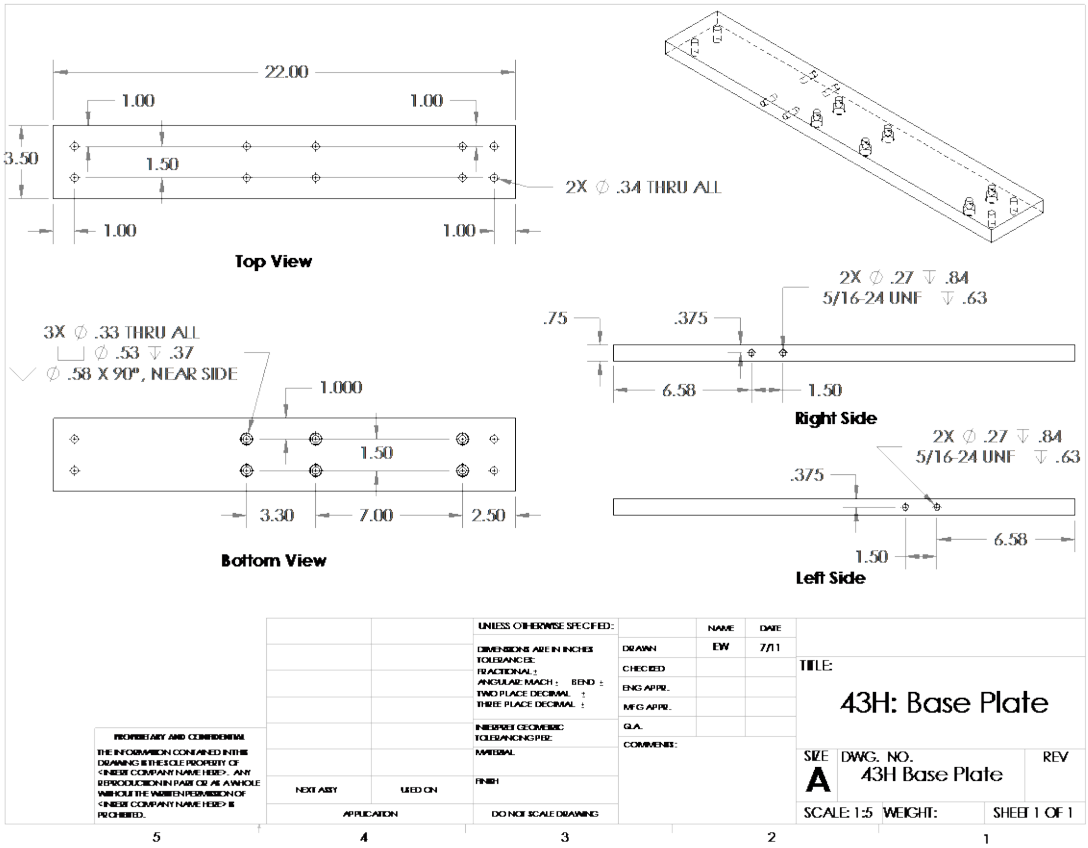


Figure A-8: Base Plate for 43H Series Servomotor.

A.2 MOUNTING STRUCTURE

The servomotor system, containing both the cable spindle and servomotor, was designed to be attached to an 80/20 frame placed around the instrumented treadmill. This frame was sized to enable two servomotor systems to be placed at the front and back of the treadmill, which allowed both the left and right legs to be perturbed during the same trial. The 80/20 frame consisted of rectangular box constructed out of the 3030 t-slot framing, and was 1.57m wide by 2.34m long by 2.54m high. A piece 1530 t-slot framing was placed between each vertical support pillar, providing a height adjustable mounting surface for the cable pulleys.

Part	Manufacturer	Part Number
Outer Frame	80/20 Inc.	3030
Adjustable Inner Frame	80/20 Inc.	1530
Corner Gussets	80/20 Inc.	4338
Motor Mount T-Studs	80/20 Inc.	3126
Slide in Nuts	80/20 Inc.	3201
Pulley Mount	Shaefer Marine, Inc.	78-05
Pulleys	Harken Inc.	183

Table A-2: Support Frame Parts List.

A.3 SOFTWARE CONTROL SYSTEM

The cable-driven device was controlled using a custom LabVIEW program running on a separate laptop computer. This program was used to start and stop Vicon data collections, count the number of gait cycles completed, and to create the cable force profile based upon the individual's walking pattern. In general, the force profile timing was based upon the time between consecutive heel strike events, with the force magnitude determined by the participant's body weight.

Timing between successive heel strike events was determined in real-time using a novel algorithm based upon medial-lateral weight shifts in the whole body COP. This algorithm was developed to reduce potential inaccuracies in the control algorithm if participants simultaneously stepped on both treadmill belts, or stepped with both feet only on one of the two belts. Medial-lateral COP location was initially calculated for each treadmill belt, using the medial-lateral (F_x) and vertical (F_z) reaction forces, and the anterior-posterior (M_y) ground reaction moment (Equation A-1). These forces and moments were sampled at 1000Hz, with the median value every 20ms used to calculate the COP location. A weighted sum based upon the vertical ground reaction forces was used to calculate the whole body COP from the individual treadmill belt locations. Medial-lateral weight shifts in the whole body COP occurred closely to heel strike events, and these weight shifts were easily detected using threshold detection algorithm based upon derivative of the whole body COP. The COP derivative was calculated in real time by taking the mean of the single point difference based upon the previous ten COP points. A ten point sliding window was used to reduce spikes in the derivative due to small COP changes, while also keeping the time delay low since only previous values could be used.

Weight shift events were detected when the derivative crossed a threshold value of ±125 mm/s, with a minimum of 400ms between successive threshold crossings.

$$COP_{ML} = \frac{(-0.015) * F_x - M_y}{F_z}$$

Equation A-1: Calculation of medial-lateral COP location based upon ground reaction forces from each treadmill belt.

Threshold crossings were used to calculate the elapsed time between each successive (left to right, and right to left) event. Separate calculation of the time from right to left, and left to right weight shift allowed for differences in limb stance times, which are observed as temporal asymmetries post-stroke. A running average over the past ten steps was used to construct the timing of the force profile. This ten step average enabled the timing of the force profile to adjust to global changes in the temporal parameters during the perturbations, while not being overly sensitive to single-step modifications. These average times were used to construct a ramp and hold profile for the motor, with the desired cable tension linearly increasing from the baseline tension (6N) to the peak force over one fourth of the average time between events, and remaining at the peak value till the next detected event. At the start of the next event, the motor which was currently at peak force output, ramps the cable tension from peak to baseline over one eighth the duration between events, as the other motor begins to ramp up to peak cable tension. These increasing and decreasing ramps were used to produce a gradual change in the perceived force, as the perturbation changed direction. The resulting force profile phased with the COM velocity in the frontal plane (Figure 4-3), with the perturbations either augmenting or resisting COM movement during walking.

Control of the force perturbation magnitude was obtained by setting the desired motor output torque to be maintained by the servormotor drive electronics. The servormotor drive was configured to operate in torque or current control mode, in which the drive electronics would attempt to maintain a desired current (torque) set point. This current set point was determined by the voltage value on external analog input channel of the servomotor drive, with the voltage magnitude corresponding to the desired current output of the drive. The relationship between the drive current and motor torque was 0.511 Nm/A for the 33H servomotor used in these experiments. Additionally, the cable force experienced by the participant was determined by dividing the motor torque output by the radius of the cable spool, 0.0445m. Calculation of the analog voltage corresponding to the servomotor drive output current is described in Equation A-2.

$$Current\ (A) = \frac{Force\ (N) * (0.0445m)}{0.511\ {}^{Nm}/_{A}}$$

Equation A-2: Calculation of current command for 33H series servomotor and drive based upon desired perturbation force to participant.

A.4 INSTRUMENTED HANDRAIL

A six-axis load cell was attached to the treadmill handrail in order to quantify handrail hold forces. The load cell was mounted to an aluminum bracket that was attached to the treadmill handrail using two U-bolts with a vibration-damping insert. A piece of rubber was placed between the insert and the handrail to further reduce any potential rotation of the handle system. A PVC spacer was used to connect the vise base to the load cell, while also providing a degree of electrical isolation. The vise base and adjusting knuckle allowed for the position of the handle to be slightly adjusted to a comfortable position for the participant.

Figure A-9: Side View of Instrumented Handle.

Part	Manufacturer	Part Number
Load Cell	AMTI	MC3A-250
Load Cell Amplifiers	AMTI	Gen 5
Servomotor Drives	Kollmorgen	AKD-B00606-NAAN-0000
33H Coupling	GAM Enterprises Inc.	EKC-25Y-14N-14N
43H Coupling	GAM Enterprises Inc.	EKC-25Y-19N-14N
Handle Standoff	McMaster Carr	92511A354
Threaded Rod	McMaster Carr	98750A013
U-Bolts	McMaster Carr	3176T34
Vise Base	PanaVise Inc.	336-V75
Adjusting Knuckle	PanaVise Inc.	851-00
Tapered Handle	McMaster Carr	62385K32

Table A-3: Instrumented Handle Parts List.

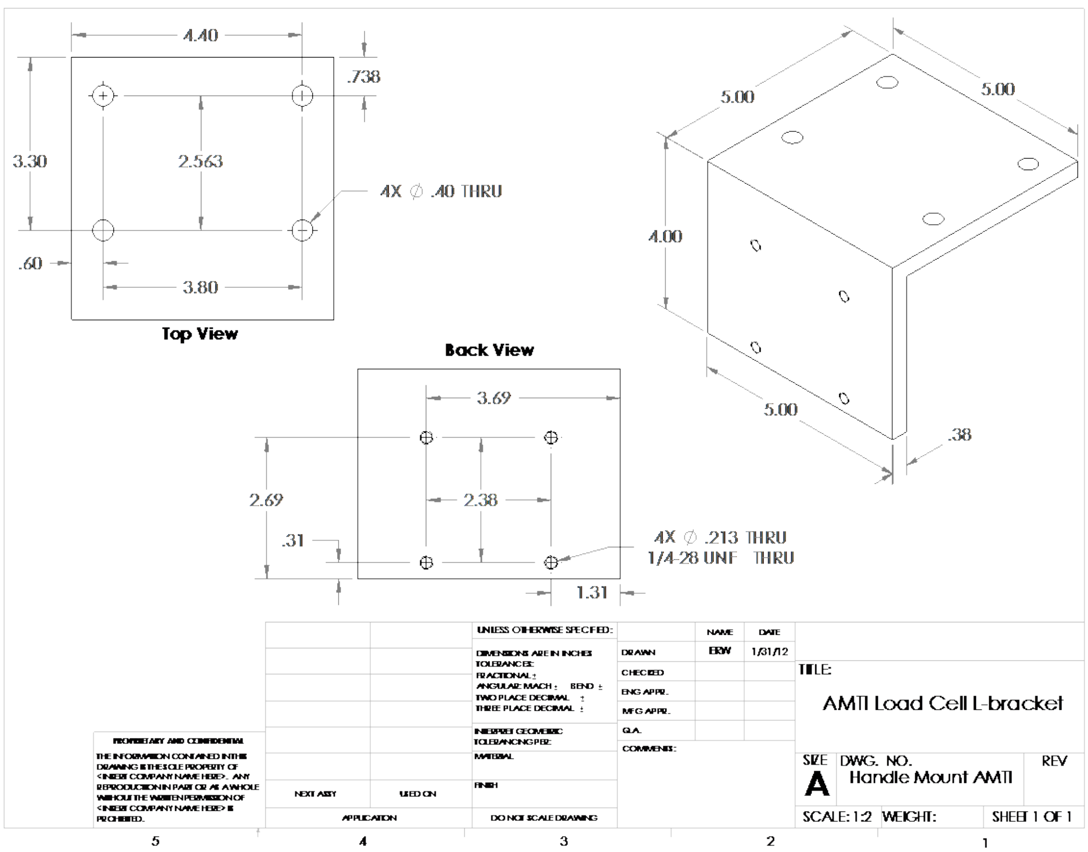


Figure A-10: Load Cell Mounting Plate.

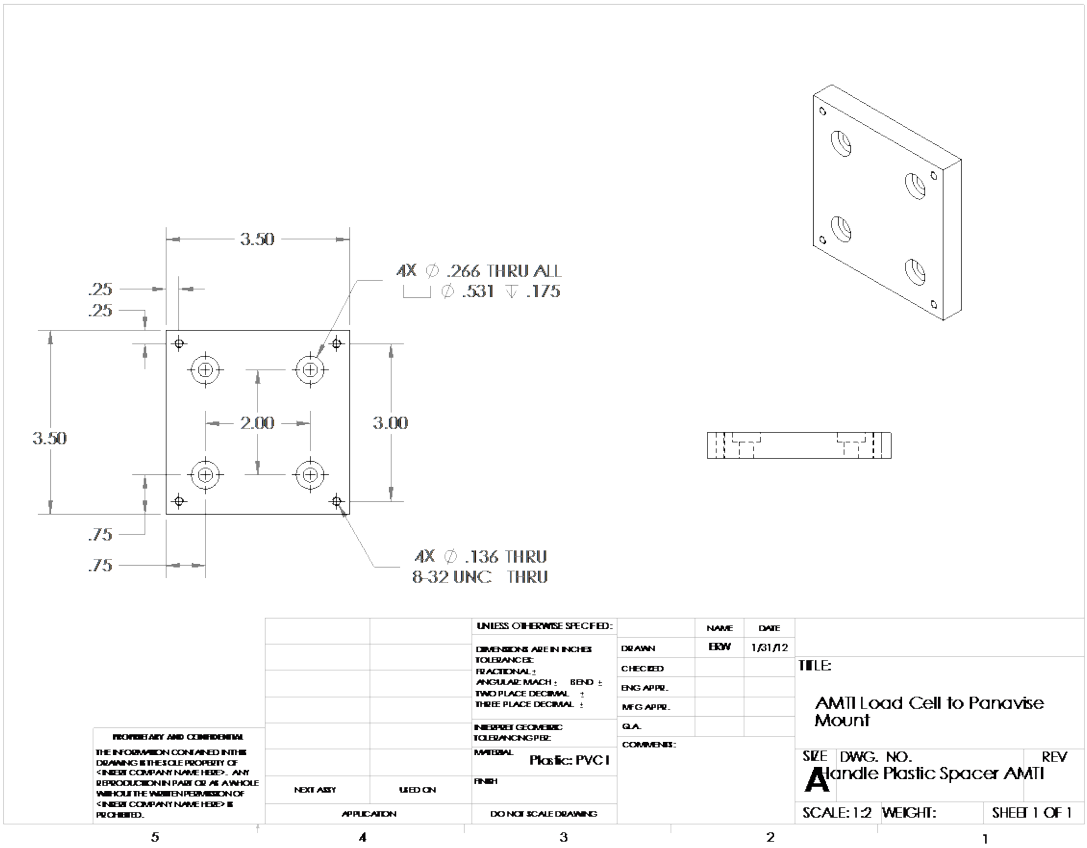


Figure A-11: Load Cell PVC Spacer. Placed between load cell and vise base.

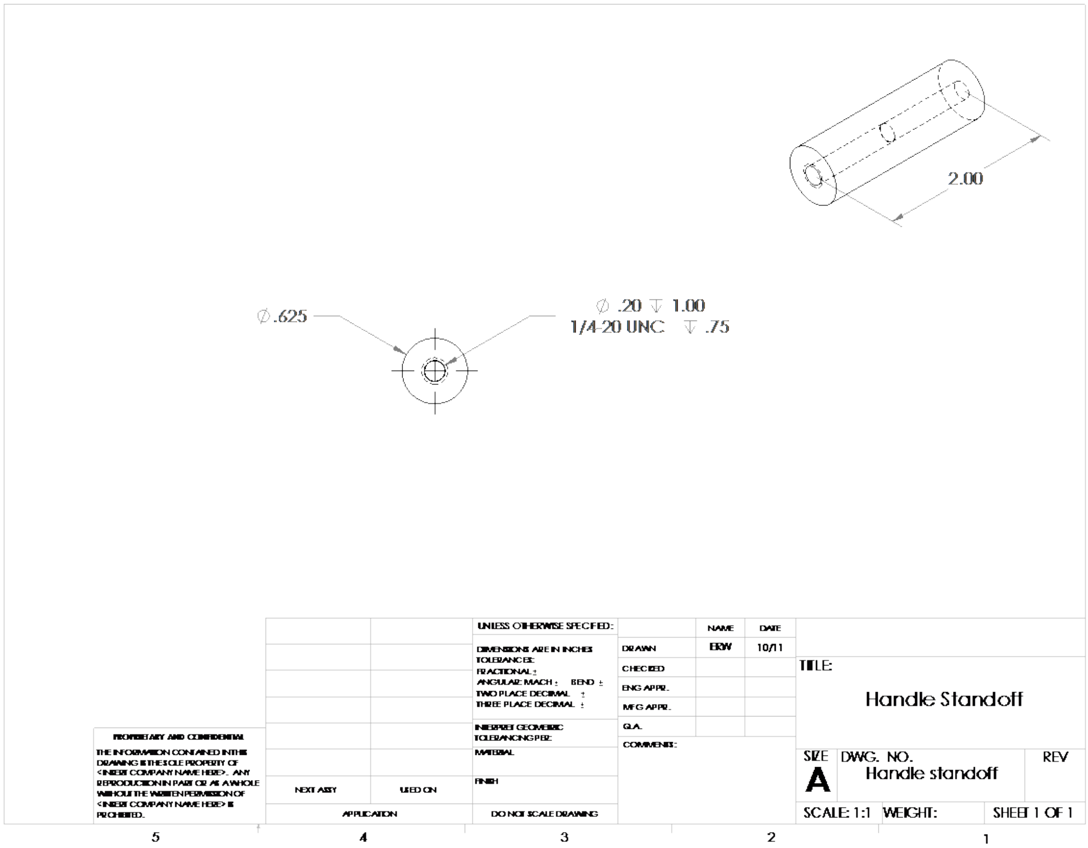


Figure A-12: Threaded Standoff. Connects vise base and handle assembly.