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The influence of lateral stabilization on walking performance and balance control in neurologically-intact and post-stroke individuals

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ABSTRACT

Background: Individuals post-stroke have an increased risk of falling, which can lead to injuries and reduced quality of life. This increased fall risk can be partially attributed to poorer balance control, which has been linked to altered post-stroke gait kinematics (e.g. an increased step width). The application of lateral stabilization to the pelvis reduces step width among neurologically-intact young and older adults, suggesting that lateral stabilization reduces the need for active frontal plane balance control. This study sought to determine if lateral stabilization is effective at improving common measures of gait performance and dynamic balance in neurologically-intact and post-stoke individuals who responded to the stabilization by reducing their step width.

Methods: Gait performance was assessed by foot placement and propulsion symmetry while dynamic balance was assessed by peak-to-peak range of frontal plane whole body angular momentum *(H_R)* and pelvis and trunk sway.

Findings: Controls and post-stroke Responders who reduced their step width in response to stabilization also reduced their mediolateral pelvis sway, but did not exhibit changes in gait performance. Contrary to expectations, both groups exhibited an increased H_R, possibly indicative of decreased balance control. This increase was the result of increased relative velocity between the pelvis and head, arms and trunk segment.

Interpretation: These results suggest that a reduction in pelvis motion alone, as opposed to relative motion between the pelvis and upper body, may increase H_R, decrease balance control and diminish gait performance. This finding has important implications for locomotor therapies that may seek to reduce pelvis motion.

1. Introduction

Individuals post-stroke are highly prone to falling, which can lead to long-term injuries, disabilities and reduced quality of life. Chronic (> 6 months) stroke survivors are more than twice as likely to fall when compared to neurologically-intact elderly (Jorgensen and Jacosen, 2002). For community-dwelling individuals post-stroke, falls most often occur during gait, likely due in part to reduced balance control (Weerdesteyn et al., 2008). Multiple factors may contribute to this reduced control, such as cognitive impairment (Tatemichi et al., 1994), reduced muscle strength (Bohannon, 2007) and asymmetric gait (Lewek et al., 2014). To help reduce these deficits, physical therapists can employ different strategies such as music therapy for cognition deficits (Särkämö et al., 2008), resistance training for muscle weakness (Bohannon, 2007) and weight supported treadmill walking in an effort to improve walking symmetry (Chen et al., 2005). A review of post-stroke therapies showed that receiving physical therapy was better for functional recovery than no therapy (Pollock et al., 2014), but others have shown that rehabilitation techniques targeting gait improvement had little influence on fall frequency (Dean et al., 2012). Thus, there still exists the need for more evidence-based therapies that specifically target balance control during rehabilitation to reduce fall risk post-stroke.

Maintaining balance during gait requires active control, but this control may be disrupted post-stroke. Frontal plane balance requires more active stabilization than sagittal plane balance, and this stabilization is largely achieved through lateral foot placement (Kuo et al., 2000) and appropriate muscle control (Neptune and McGowan, 2016). Over 50% of individuals post-stroke report impaired tactile and proprioceptive feedback (Connell, 2008), which can negatively affect the ability to sense foot position. Individuals post-stroke also exhibit disrupted motor control such as improper muscle coactivation

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(Lamontagne et al., 2000) or altered muscle excitation timing (Garland et al., 2009). Possibly as a result of these somatosensory and motor deficits, individuals with chronic stroke exhibit a reduced link between mediolateral motion of the pelvis and paretic step width (Stimpson et al., 2019). Rather than adjusting mediolateral foot placement on a step-by-step basis, individuals post-stroke tend to consistently place their paretic leg farther laterally from the pelvis, irrespective of the pelvis's mechanical state (Stimpson et al., 2019). This asymmetry in mediolateral foot placement has been linked to an altered pattern of muscle control in stroke survivors with balance deficits (Dean and Kautz, 2015). Thus, improved balance control may be accompanied by reduced foot placement asymmetry.

While foot placement location is generally acknowledged as important for ensuring bipedal walking balance (Bruijn and van Dieen, 2018), recent work has suggested that push-off with the trailing leg can also contribute to gait stabilization (Reimann et al., 2018). Following a stroke, the pattern of push-off forces (or propulsion) often changes, with paretic propulsion magnitude linked to kinematic asymmetries and muscle activation patterns (Roelker et al., 2019). Thus, quantifying the propulsion and braking produced by each leg may be useful in assessing balance control and overall walking performance post-stroke.

In addition to foot placement and propulsion asymmetries, previous work has shown that some individuals post-stroke walk with wider steps when compared to neurologically-intact, age-matched controls (Chen et al., 2005). Wider steps improve static balance by increasing the base of support, but also increase the peak-to-peak range of frontal plane whole body angular momentum (H_R) (Nott et al., 2014). Angular momentum is a promising measure of dynamic balance, as an increased H_R correlates with lower Berg Balance Scale and Dynamic Gait Index scores in post-stroke individuals (Nott et al., 2014; Vistamehr et al., 2016), indicating poor balance and increased fall risk (Blum and Korner-Bitensky, 2008; Jonsdottir and Cattaneo, 2007). Thus, wider steps could be a maladaptive compensatory mechanism that ultimately decreases balance control.

When lateral stabilization is applied to the pelvis, young and older adults walk with reduced step width (Dean et al., 2007; Donelan et al., 2004; Ortega et al., 2008). In some cases, step width variability and energetic cost have also been reported to decrease during stabilized walking, suggesting that lateral stabilization reduces the need for active frontal plane balance control. However, the effect of lateral stabilization on step width, balance control and gait performance measures such as appropriate foot placement and ground reaction force generation has yet to be examined in individuals post-stroke. Removing the need to actively control frontal plane balance with an impaired motor control system may result in a more typical walking pattern for individuals post-stroke and lead to improved rehabilitation outcomes.

The purpose of this study was to investigate whether the reductions in step width typically caused by external lateral stabilization are accompanied by improvements in walking performance (i.e., foot placement and propulsion symmetry metrics) and dynamic balance (i.e., H_R and pelvis and trunk sway). To address this question, we focused our analyses on participants who reduced their step width when laterally stabilized and hypothesized that such neurologically-intact and post-stroke subjects (i.e., responders in both groups) would show improved walking performance and increased balance control.

2. Methods

2.1. Experimental data

Twenty hemiparetic, post-stroke individuals (10 right hemiparesis, 9 male, age: 50 (SD 16) years) and twenty age- and gender-matched neurologically-intact controls (9 male, age: 49 (SD 16) years) were analyzed. All participants provided informed, written consent in accordance with the Medical University of South Carolina's institutional review board.

Kinematic data were collected at 120 Hz using a 16-camera motion capture system (PhaseSpace Inc., San Leandro, USA) using a modified Helen Hayes marker set. Kinetic data were collected using a dual-belt instrumented treadmill (Bertec Corp., Columbus, USA) at 2000 Hz. Subjects walked at their self-selected speed during two conditions: normal and stabilized. For the normal condition, subjects completed three 30-s walking trials and were allowed a 1-min rest or longer if needed after the completion of each trial. For the stabilized condition, lateral stabilization was delivered through a custom cable-based robotic system (see Appendix for more detail). Subjects became familiar walking with the lateral stabilization over a 6-min period, after which their step width was checked to see if it had reached a consistent value (determined using a repeated-measures ANOVA comparing step widths across the 6-min period of walking during the adaptation trial), confirming that they had completed their adaptation to the stabilization. If their step width had not reached steady-state at the end of the 6-min, participants repeated the adaptation trials until their step width reached a consistent value. Subjects then completed three 30-s trials while stabilized.

2.2. Data analysis

GRF data collected from the instrumented treadmill were smoothed with a 4th order Savitzky-Golay filter acting on 21 samples. Kinematic and kinetic data were then resampled at 100 Hz. Due to the frequency of cross-over steps, a consequence of reduced step width, heel-strike events were determined using a velocity-based algorithm (Zeni et al., 2008). All data were processed and analyzed using custom LabVIEW (National Instruments, Austin, USA) and MATLAB (MathWorks, Natick, USA) programs.

Gait performance was assessed using foot placement and propulsion symmetry metrics. Step width (SW) was defined as the mediolateral (ML) distance between paretic and non-paretic heel markers at consecutive heel strikes (HS) and normalized by subject leg length. SW variability was defined as the standard deviation of SW over a 30-s trial. ML foot placement was calculated as the lateral distance between the body center of mass (*CoM*) and heel marker at HS normalized by subject leg length, and ML foot placement variability was calculated as the standard deviation of these values. Individual leg contributions to propulsion (braking) were quantified as the time integral of the positive (negative) anterior GRF over the paretic leg stride. ML foot placement ratio and propulsion ratio were used as measures of gait symmetry, and were defined as the paretic side (left leg for Controls) variable divided by the sum of the paretic and non-paretic side (left and right legs for Controls) variables with a value of 0.5 indicating perfect symmetry (i.e. both legs were placed at equal distances from the *CoM* and both legs produced equal propulsion/braking).

Balance control was quantified using H_R and pelvis and trunk sway, where H was calculated as the sum of the angular momentum of thirteen segments about the whole-body *CoM* and normalized to subject leg length, mass and walking speed (Shell et al., 2017; Vistamehr et al., 2018) as follows:

$$\vec{H} = \sum_{i=1}^{n} [\, (\vec{r}_i^{\,COM} - \vec{r}_{body}^{\,COM}) \times m_i (\vec{v}_i^{\,COM} - \vec{v}_{body}^{\,COM}) + I_i \vec{\omega}_i \,]$$

where $\vec{r}_i^{\,COM}$, $\vec{v}_i^{\,COM}$ are the position and velocity vectors of the i^{th} segment's *CoM*, respectively. $\vec{r}_{body}^{\,COM}$ and $\vec{v}_{body}^{\,COM}$ are the position and velocity vectors of the whole-body *CoM*, m_i, I_i and $\vec{\omega_i}$ are the mass, moment of inertia and angular velocity of each segment and n is the number of body segments. H_R was computed as the peak-to-peak range of frontal plane H over the gait cycle. Pelvis and trunk sway were defined as the peak-to-peak range of pelvis and trunk *CoM* movement in the mediolateral direction during the paretic leg stride. The *CoM* of each segment was calculated as the centroid of the rigid body associated with each segment and tracked by an inverse kinematics algorithm.

2.3. Statistical analysis

Twenty subjects in each group provided at least 80% power for detecting a difference of one standard deviation. Thus, this sample size provides sufficient power even with anticipated participant dropouts or non-responders that aren't included in the analyses. A subject was included in the presented analyses if their step width significantly decreased ($p < 0.05$) between the normal and stabilized conditions, as determined using a one-tailed *t*-test. These subjects are hereby referred to as Controls and post-stroke Responders. The Shapiro-Wilk normality test was used to test for normal distribution of all outcome variables (step width; step width variability; bilateral ML foot placement variability; ML foot placement ratio; bilateral propulsive impulse; bilateral braking impulse; propulsion ratio; H_R; pelvis sway; trunk sway) within each group. For normally distributed variables, paired two tailed *t*-tests were used to test for significant differences between normal and stabilized conditions within groups for the outcome measures. The Wilcoxon signed-rank test was used to test for differences between conditions within groups for the non-normally distributed outcome measures. A difference was considered significant if $p < 0.05$.

3. Results

Thirteen post-stroke Responders and 18 Controls responded to the lateral stabilization by significantly reducing their step width ($p \leq 0.0002$; Fig. 1). On average, post-stroke Responders decreased their normalized step width by 19% (SD 12%) while Controls decreased their step width by 36% (SD 18%) between normal and stabilized conditions. Post-stroke Responders walked at a self-selected speed of 0.50 m/s (SD 0.20 m/s) while Controls walked at 0.81 m/s (SD 0.28 m/s). The remaining 7 stroke subjects (Non-Responders) and 2 control subjects (Non-Responder Controls) either did not change or increased their step width while stabilized. Because this study was focused on the effects of the stabilization on individuals that responded to it by reducing their step width, their results were not included in the main study, but can be found in the Appendix.

3.1. Gait performance

In general, stabilization did not significantly improve or alter gait performance in the post-stroke Responders (Table 1). Even though post-stroke Responders significantly decreased their step width, there was no significant change in ML foot placement ratio ($p = 0.82$). Thus, the post-stroke Responders continued to place their paretic leg closer to their *CoM* than their non-paretic leg even though both legs were placed closer to the *CoM* while stabilized. Stabilization significantly decreased ML foot placement variability in both the paretic ($p = 0.002$, average decrease of 0.01 LL) and non-paretic legs ($p \leq 0.002$, average decrease of 0.02 LL), but did not decrease step width variability in the post-stroke Responders ($p = 0.38$).

The Control subjects showed significant changes in all gait performance measures except step width variability (Table 1). The addition of stabilization significantly changed their ML foot placement ratio ($p = 0.01$), bringing it closer to symmetric (0.5) than while walking normally. ML foot placement variability in Controls also decreased significantly for both legs ($p < 0.01$, average decrease of 0.01 LL).

Stabilization significantly decreased the post-stroke Responders' ($p \leq 0.04$; Fig. 2) and Controls' propulsive impulses ($p = 0.04$; Fig. 2), but did not affect either group's braking impulses (Fig. 2). For both conditions, post-stroke Responders produced less propulsion with their paretic leg compared to their non-paretic leg, and there was no change in post-stroke Responders' paretic propulsion ratio (Fig. 3). Conversely, stabilization significantly altered propulsion symmetry in Controls ($p = 0.02$; Fig. 3). Controls were nearly symmetrical in the normal walking condition, but their average propulsive impulse over a stride significantly reduced for both legs when stabilized ($p < 0.0001$) with a greater reduction in the right leg impulse (Fig. 2).

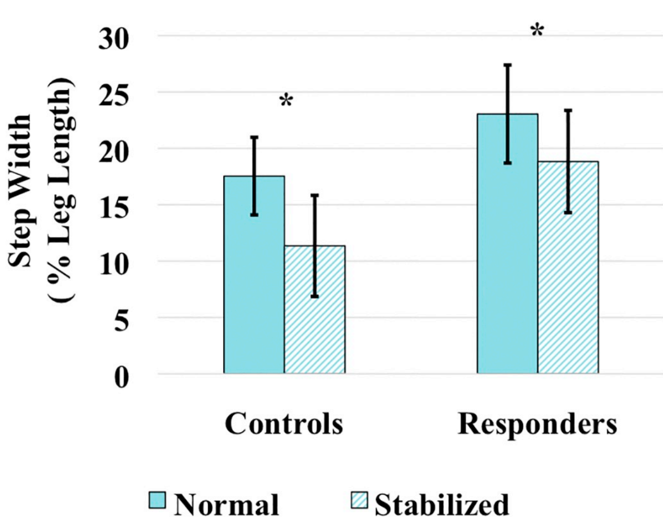

Fig. 1. Step width (normalized to subject leg length) decreased for both Controls and post-stroke Responders while walking with stabilization. * indicates a significant ($p < 0.05$) difference between normal and stabilized conditions.

3.2. Dynamic balance control

Contrary to our expectations, both post-stroke Responders and Controls showed a significant increase in H_R with stabilization ($p < 0.0001$; Fig. 4). On average, the stabilization increased post-stroke Responders' H_R by 27% (SD 20%) and Controls by 43% (SD 32%). Stabilization significantly decreased pelvis sway in both post-stroke Responders ($p = 0.03$; Fig. 5a) and Controls ($p < 0.0001$; Fig. 5a), while trunk sway only significantly decreased for Controls ($p = 0.0002$; Fig. 5b).

4. Discussion

The goal of this study was to examine the effects of lateral stabilization on gait performance and dynamic balance control. Our hypothesis that those who responded to lateral stabilization would walk with improved gait performance and balance control was not supported. Despite the stabilization reducing the need for post-stroke Responders to rely on their impaired motor system for balance control, neither post-stroke Responders nor Controls exhibited significant improvements in gait performance or balance control metrics.

4.1. Gait performance

Despite the lack of improvement in gait performance or balance control, the response to the stabilization was largely consistent with previous research in neurologically-intact adults (Dean et al., 2007; Donelan et al., 2004; Ortega et al., 2008; Wu et al., 2017) and individuals with incomplete spinal cord injury (Wu et al., 2017). The application of lateral stabilization appeared to reduce the need to actively control frontal plane balance as evidenced by a reduction in step width and ML foot placement variability for both Controls and post-stroke Responders. We also found that stabilization led to a decrease in the sway of the pelvis *CoM* in Controls and post-stroke Responders. While lateral stabilization has yet to be studied in the post-stoke population, our results were consistent with those showing a reduction in body *CoM* sway in neurologically-intact and incomplete spinal cord injury individuals when walking in a stabilizing force field (Wu et al., 2017).

Pelvis stabilization did not cause the same changes in gait

Table 1
Gait performance metrics for Controls and post-stroke Responders. All values are presented as mean (one standard deviation). **Bolded** values indicate a significant ($p < 0.05$) change between normal and stabilized conditions. (LL): normalized to subject leg length.

		Normal	Stabilized	p-Value
Controls	*M/L foot placement ratio*	**0.46 (0.04)**	**0.52 (0.11)**	**0.009**
	Left M/L foot placement variability (LL)	**0.03 (0.01)**	**0.02 (0.01)**	**< 0.0001**
	Right M/L foot placement variability (LL)	**0.03 (0.01)**	**0.02 (0.01)**	**< 0.0001**
	Step width variability (LL)	0.015 (0.005)	0.016 (0.004)	0.80
Post-stroke Responders	*M/L foot placement ratio*	0.46 (0.04)	0.47 (0.11)	0.823
	Paretic M/L foot placement variability (LL)	**0.05 (0.01)**	**0.04 (0.01)**	**0.002**
	Non-paretic M/L foot placement variability (LL)	**0.05 (0.01)**	**0.03 (0.01)**	**0.001**
	Step width variability (LL)	0.018 (0.005)	0.017 (0.005)	0.38

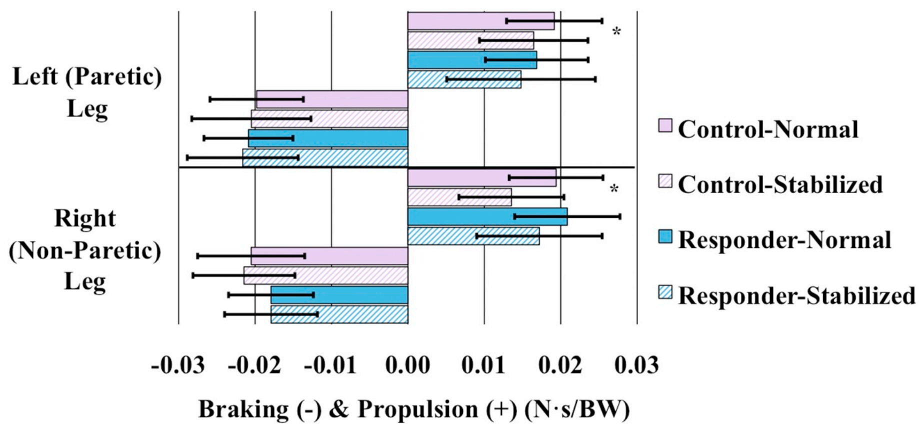

Fig. 2. For both Controls and post-stroke Responders, walking with stabilization significantly decreased average propulsion produced by both legs, but had no effect on braking. * indicates a significant ($p < 0.05$) difference between normal and stabilized conditions.

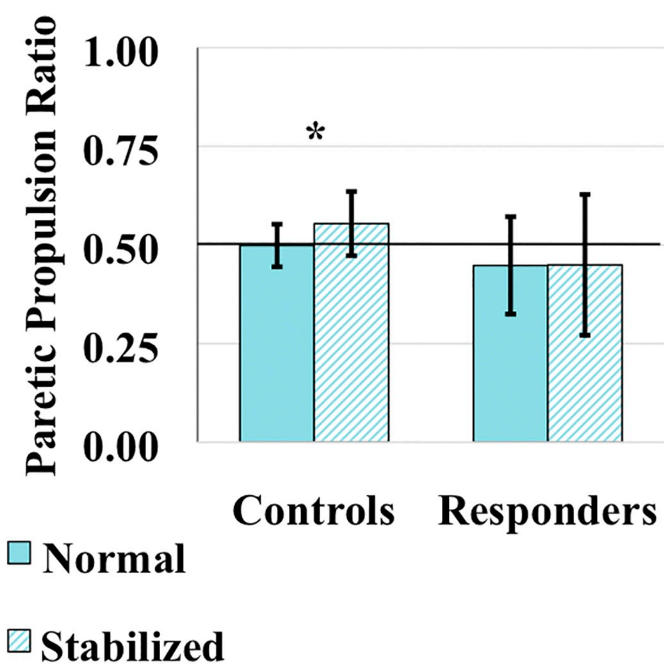

Fig. 3. Average paretic propulsion ratio for Controls and post-stroke Responders. Black line at 0.5 indicates perfect symmetry and * indicates a significant ($p < 0.05$) difference between normal and stabilized conditions.

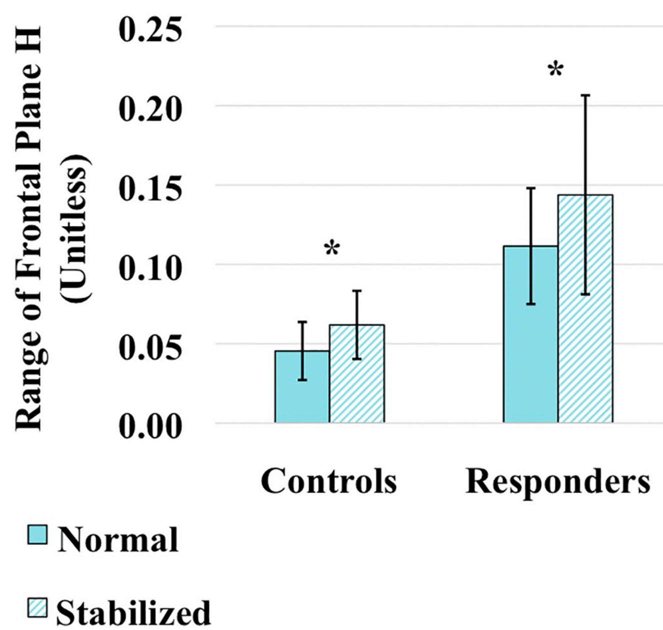

Fig. 4. Range of frontal plane whole body angular momentum (H_R) increased for both Controls and post-stroke Responders. Normalized to subject leg length, mass and walking speed. * indicates a significant ($p < 0.05$) difference between normal and stabilized conditions.

performance for each subject, as seen in the increased inter-subject variability for both Controls and post-stroke Responders with stabilization (Figs. 6a-b). This result indicates that despite the similar global adaptation to the stabilization (i.e., step width reduction) there was not a consistent direction of change in gait performance metrics within the groups. This increased variability was accompanied by a majority of subjects exhibiting poorer, less symmetric walking performance. While the average paretic propulsion and ML foot placement ratios for Controls and post-stroke Responders were within 0.05 of perfect symmetry (0.50), this result was not because all subjects walked symmetrically. There was a large range of propulsion (Fig. 6a) and ML foot placement (Fig. 6b) ratios within both groups and this range increased with

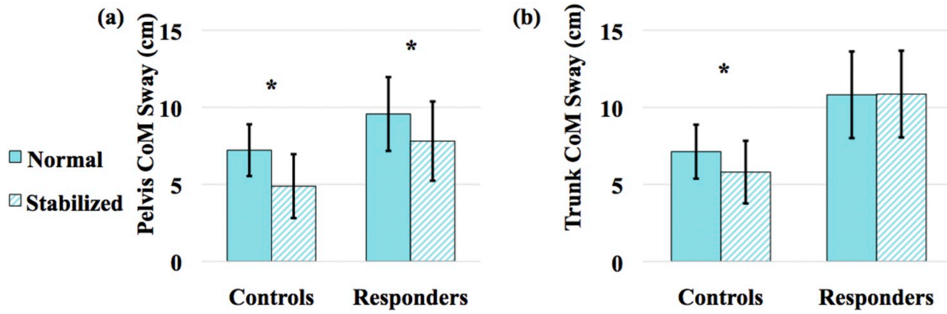

Fig. 5. Range of (a) pelvis center of mass (*CoM*) sway and (b) trunk *CoM* sway in the frontal plane. * indicates a significant ($p < 0.05$) difference between normal and stabilized conditions.

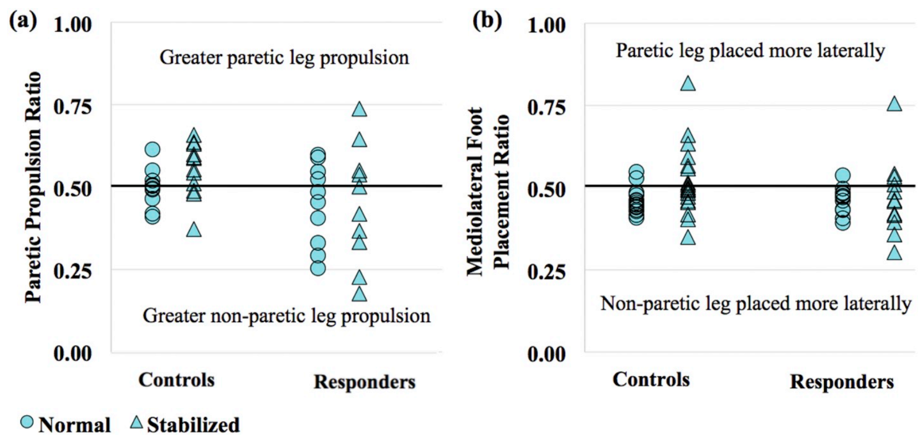

Fig. 6. (a) Paretic propulsion ratios (b) mediolateral foot placement ratios for all Controls and post-stroke Responders. Symmetry ratio is the paretic (left) leg value divided by the sum of the paretic (left) and non-paretic (right) leg values. Black line at 0.5 indicates perfect symmetry.

stabilization such that a majority of the individuals in both groups did not improve propulsion or ML foot placement symmetry while walking with stabilization.

Despite the increased inter-subject variability with stabilization, it is interesting to note that in both the normal and stabilized walking conditions post-stroke Responders tended to place their non-paretic leg farther from their body *CoM* than their paretic leg. This was in contrast to previous work that found individuals post-stroke often walk with their paretic leg placed more laterally than their non-paretic leg (Balasubramanian et al., 2010; Dean and Kautz, 2015). Perhaps individuals who walk with excessively lateral paretic foot placement are particularly resistant to the “assistance” provided by lateral stabilization, and thus were classified as Non-Responders and not included in these analyses.

In addition to propulsion symmetry not improving with stabilization, both Controls and post-stroke Responders produced significantly less propulsion when walking with stabilization (Fig. 3). There was not a corresponding decrease in braking, thus both groups had an average net negative impulse while walking with stabilization. Since both groups averaged a net impulse of approximately zero when walking normally, the net negative impulse was most likely due to the addition of the stabilization. While the robotic stabilization system was on rollers that allowed the subject to drift anteriorly and posteriorly on the treadmill, the tracks that the system runs on were not perfectly frictionless. It is possible that the friction coupled with the hoop harness attachments in the front and back of the subject may have provided a small propulsive force that decreased the net propulsive force generated by the subjects. However, we do not believe this affected our results as it is unlikely that the small propulsive force generated by the stabilization would cause decreased gait performance and balance control, particularly since a majority of the measures were in the lateral direction.

4.2. Balance control

In addition to the lack of effect or, in some cases, the negative effect of stabilization on gait performance, H_R significantly increased for both post-stroke Responders and Controls. This increased H_R suggests that walking with pelvis stabilization reduced balance control, contrary to our hypothesis, and may help explain why subjects' gait performance did not improve with stabilization. H in the frontal plane is influenced by the ML and vertical moment arms between the body *CoM* and individual segments, the relative ML and vertical velocities, and the segment's angular momentum about its own *CoM* in the frontal plane (Eq. (1)). We initially assumed that the reduction in step width and pelvis sway would lead to a corresponding reduction in the ML moment arms and relative velocities between the body *CoM* and the individual segment's *CoM*, thus decreasing H_R.

To investigate potential causes of the unexpected increase in H_R, we performed a post hoc analysis to quantify the contribution of individual body segments to overall frontal plane H. The advantage of using H over other more global measures of balance control such as margin of stability (Hof, 2008) is that we can analyze the dynamic interaction among body segments to understand the observed differences between conditions. We compared the contributions of the HAT (head, arms and trunk) and lower body (pelvis and legs) segments to the peak value of frontal plane H between the normal and stabilized conditions. When the pelvis was stabilized, the HAT segment's contribution to the peak value of frontal plane H significantly increased from 26% (SD 8%) to 38% (SD 7%) for Controls ($p < 0.0001$; Fig. 7a) and from 34% (SD 4%) to 42% (SD 5%) for post-stroke Responders ($p < 0.0001$; Fig. 7b). This increase was accompanied by a significant decrease in the lower body contribution to peak H for Controls ($p < 0.0001$; Fig. 7a) and post-stroke Responders ($p < 0.0001$; Fig. 7b).

Separating the HAT segment into individual segments revealed that

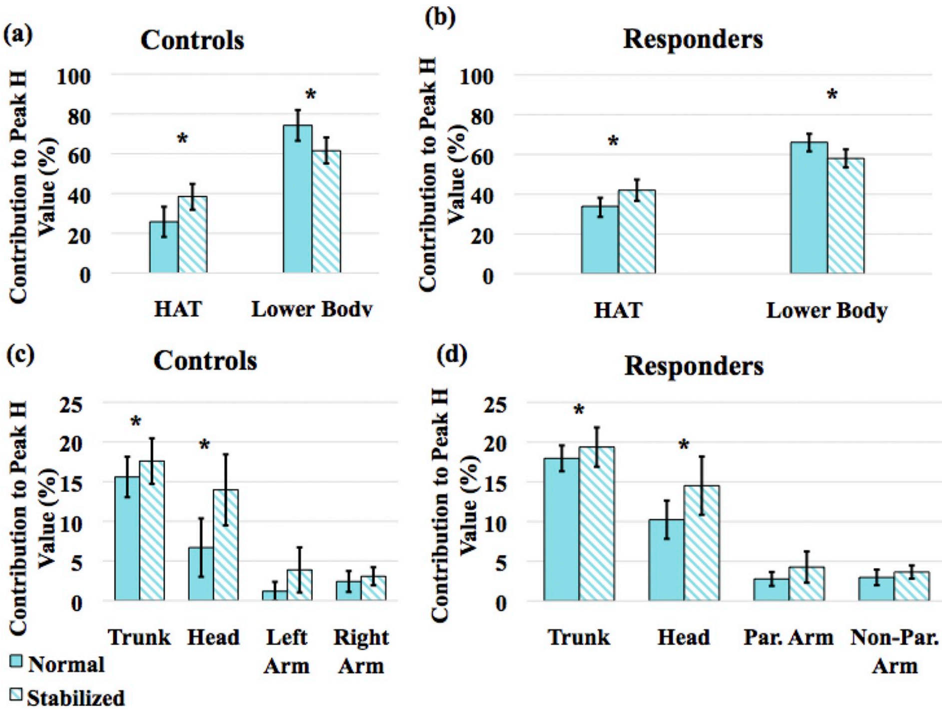

Fig. 7. Contribution of HAT (head, arms and trunk) and Lower Body (pelvis and legs) to the peak value of frontal plane H for (a) Controls and (b) post-stroke Responders. Individual segment contributions to HAT for (c) Controls and (d) post-stroke Responders. * indicates a significant ($p < 0.05$) difference between normal and stabilized conditions.

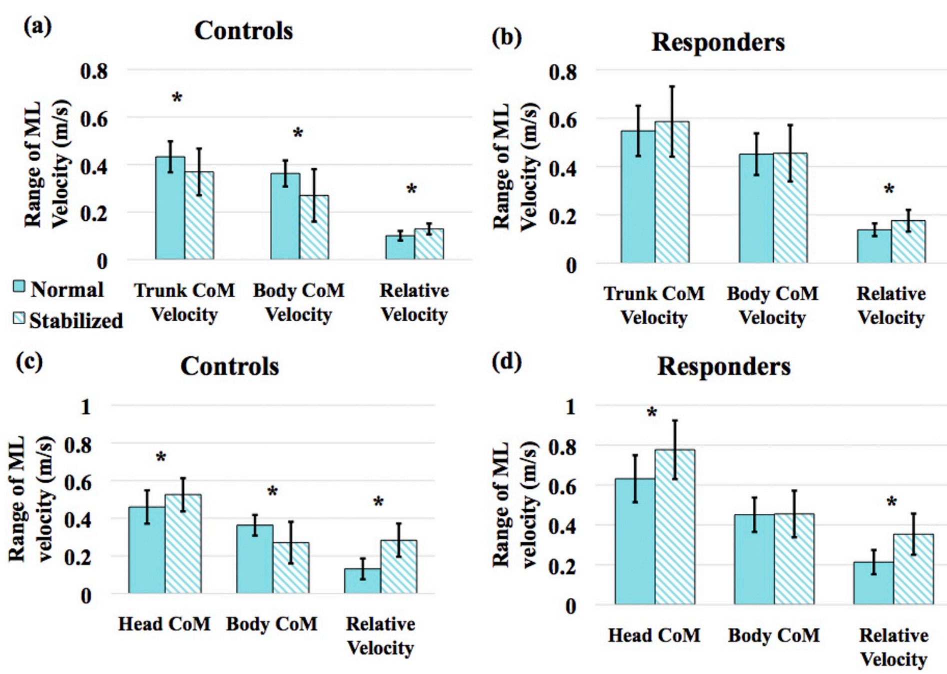

Fig. 8. Range of mediolateral (ML) velocity of the trunk center of mass (*CoM*), body *CoM* and the relative velocity between the two for (a) Controls and (b) post-stroke Responders. Range of ML velocity of the head *CoM*, body *CoM* and the relative velocity between the two for (c) Controls and (d) post-stroke Responders. * indicates a significant ($p < 0.05$) difference between normal and stabilized conditions.

the greatest increases in the peak value of *H* were due to increased angular momentum of the trunk and head for both Controls (Fig. 7c) and post-stroke Responders (Fig. 7d). The analysis indicated there was a significant decrease in the range of linear velocities of the body ($p = 0.0007$) and trunk ($p = 0.005$) *CoM* in Controls (Fig. 8a). Despite the overall decrease in the range of both velocities, the body *CoM* velocity decreased more than the trunk *CoM* velocity leading to an increased relative velocity. The direction of change of the trunk and body *CoM* velocities was not as consistent in post-stroke Responders (Fig. 8b), but it did ultimately lead to a significant ($p = 0.002$) increase in relative velocity. The trunk angular momentum also significantly increased in Controls and post-stroke Responders (Fig. 9a), which corresponds to an increase in the trunk's angular velocity since the moment of inertia is constant. This trend of increased relative velocity was also present in the head, but the change was due to a significant increase in the ML velocity of the head *CoM* for Controls ($p = 0.002$; Fig. 8c) and post-stroke Responders ($p = 0.001$; Fig. 8d). There was no significant change in the head's angular velocity for Controls (Fig. 9b), but there was a small, but significant ($p = 0.008$), decrease for post-stroke Responders (Fig. 9b). For Controls, there were significant, but small, increases in range of the ML moment arms and relative vertical velocities of the trunk and head and the range of the trunk's vertical moment arm (Table 2). For post-stroke Responders there were significant and small increases in the ML moment arms of the head and trunk and relative velocity of the head (Table 3). The magnitude of the changes was much smaller for these components than the increases observed in the ranges

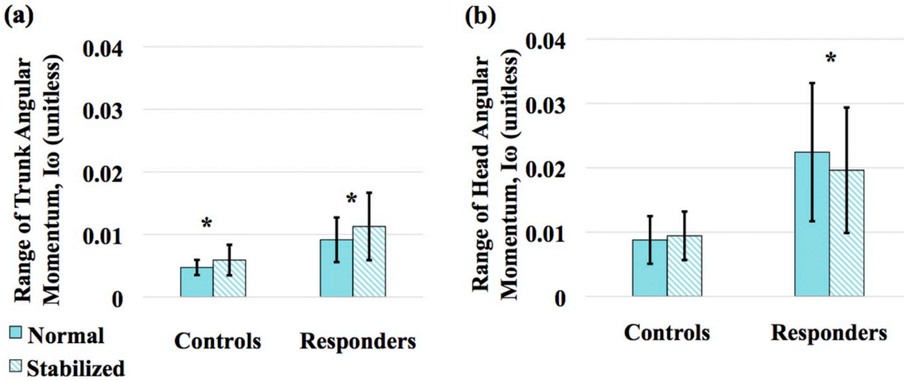

Fig. 9. Range of (a) trunk and (b) head angular momentum about its own Center of Mass (Iω) normalized by subject leg length, mass and walking speed. * indicates a significant ($p < 0.05$) change between normal and stabilized conditions.

Table 2
Range of mediolateral (ML) moment arm, vertical (vert.) moment arm, segmental vert. velocity and relative vert. velocity of the trunk and head for Controls. All values are presented as mean (one standard deviation). **Bolded** values indicate a significant ($p < 0.05$) difference between normal and stabilized.

	Normal	Stabilized	p-Value
Trunk			
Range ML moment arm (cm)	**1.4 (0.4)**	**1.8 (0.5)**	**0.004**
Range vert. moment arm (cm)	**0.5 (0.1)**	**0.7 (0.3)**	**0.002**
Range vert. velocity (m/s)	0.29 (0.1)	0.33 (0.2)	0.12
Range relative vert. velocity (m/s)	**0.06 (0.01)**	**0.1 (0.07)**	**0.003**
Head			
Range ML moment arm (cm)	**2.0 (0.9)**	**4.1 (1.4)**	**< 0.0001**
Range vert. moment arm (cm)	0.7 (0.2)	0.7 (0.2)	0.45
Range vert. velocity (m/s)	0.3 (0.1)	0.3 (0.1)	0.94
Range relative vert. velocity (m/s)	**0.07 (0.02)**	**0.09 (0.04)**	**0.03**

Table 3
Range of mediolateral (ML) moment arm, vertical (vert.) moment arm, segmental vert. velocity and relative vert. velocity of the trunk and head for post-stroke Responders. All values are presented as mean (one standard deviation). **Bolded** values indicate a significant ($p < 0.05$) difference between normal and stabilized conditions.

	Normal	Stabilized	p-Value
Trunk			
Range ML moment arm (cm)	**2.4 (0.8)**	**2.8 (0.9)**	**< 0.001**
Range vert. moment arm (cm)	0.7 (0.2)	0.7 (0.2)	0.22
Range vert. velocity (m/s)	0.25 (0.09)	0.24 (0.09)	0.89
Range relative vert. velocity (m/s)	0.08 (0.02)	0.08 (0.02)	0.14
Head			
Range ML moment arm (cm)	**4.0 (1.2)**	**6.7 (3.0)**	**< 0.0001**
Range vert. moment arm (cm)	1.0 (0.3)	1.1 (0.5)	0.50
Range vert. velocity (m/s)	0.24 (0.09)	0.23 (0.08)	0.08
Range relative vert. velocity (m/s)	**0.09 (0.03)**	**0.10 (0.04)**	**0.003**

of ML and angular velocity. Therefore, the increase in the relative velocity between the HAT segments and body *CoM* coupled with the increase in the angular velocity is the most likely cause of the increased H_R in the Controls and post-stroke Responders. While lateral stabilization may have decreased the absolute motion of the pelvis, it created larger relative motion between the body *CoM* and HAT *CoM*, thus increasing the overall H_R. This increase in relative motion suggests that stabilizing the pelvis will not improve balance control unless care is taken to ensure that the relative motion between the pelvis and HAT segment does not increase as a result.

One limitation of this study is that we assumed that the lateral stabilization forces would act through the body *CoM* and thus not create an additional external moment. While care was taken in placing the harness that applied the stabilization, there was no way of knowing if the applied forces were perfectly in line with the body *CoM*. However, since there was an overall decrease in pelvis sway, which was the purpose of the stabilization, it is reasonable to assume that any additional frontal plane moment created by the stabilization was relatively small. Another important point was that the subjects were not instructed on how to adapt to the stabilization (i.e., reduce step width). Thus, improvements in gait performance and dynamic balance may be possible if subjects have the opportunity to train with the stabilization and are provided feedback on how to improve their walking performance. This is an important area for future work.

5. Conclusions

The study results indicate that reduction in absolute motion of the pelvis, as opposed to relative motion between the pelvis and upper body, may lead to decreased balance control by increasing H_R. It is possible that the reduced pelvis motion provided by the stabilization led to insufficient foot clearance that necessitated altered motion of other segments. This finding has important implications for locomotor therapies such as body weight supported walking that constrain the pelvis during locomotion. Future work should continue to explore how reducing the lateral motion of the pelvis affects the contributions of the other body segments to frontal plane H and dynamic balance control and whether feedback during stabilization training may improve walking performance.

Declaration of competing interest

The authors have no conflicts of interest to declare.

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Appendix A

Table A1
Gait performance and dynamic balance metrics for Control Non-Responders. All values are presented as the mean (one standard deviation). LL: normalized to subject leg length. BW: subject body weight.

	Normal	Stabilized
Step width (LL)	0.16 (0.06)	0.17 (0.09)
Step width variability (LL)	0.013 (0.00)	0.016 (0.003)
M/L foot placement ratio	0.47 (0.009)	0.46 (0.03)
Left M/L foot placement variability (LL)	0.02 (0.02)	0.03 (0.01)
Right M/L foot placement variability (LL)	0.03 (0.02)	0.03 (0.01)
Pelvis CoM sway (cm)	6.4 (1.2)	6.6 (2.0)
Trunk CoM sway (cm)	6.0 (2.3)	7.3 (3.5)
Left leg propulsion (N•s/BW)	0.03 (0.005)	0.03 (0.00)
Left leg braking (N•s/BW)	−0.03 (0.00)	−0.03 (0.00)
Right leg propulsion (N•s/BW)	0.03 (0.01)	0.03 (0.00)
Right leg braking (N•s/BW)	−0.03 (0.01)	−0.03 (0.01)
Propulsion symmetry ratio	0.49 (0.04)	0.60 (0.14)
Range of frontal plane H	0.024 (0.01)	0.043 (0.02)
Self-selected speed (m/s)	1.1 (0.3)	

Table A2
Gait performance and dynamic balance metrics for post-stroke Non-Responders. All values are presented as the mean (one standard deviation). (LL): normalized to subject leg length. BW: subject body weight.

	Normal	Stabilized
Step width (LL)	0.22 (0.09)	0.23 (0.09)
Step width variability (LL)	0.018 (0.003)	0.019 (0.006)
M/L foot placement ratio	0.48 (0.13)	0.46 (0.19)
Paretic M/L foot placement variability (LL)	0.05 (0.02)	0.04 (0.02)
Non-paretic M/L foot placement variability (LL)	0.04 (0.01)	0.04 (0.01)
Pelvis CoM sway (cm)	10.4 (3.6)	9.6 (1.4)
Trunk CoM sway (cm)	11.6 (3.6)	12.6 (3.3)
Paretic leg propulsion (N•s/BW)	0.01 (0.01)	0.01 (0.01)
Paretic leg braking (N•s/BW)	−0.02 (0.01)	−0.02 (0.01)
Non-Paretic leg propulsion (N•s/BW)	0.02 (0.01)	0.02 (0.01)
Non-Paretic leg braking (N•s/BW)	−0.02 (0.01)	−0.02 (0.01)
Propulsion symmetry ratio	0.33 (0.18)	0.26 (0.14)
Range of frontal plane H	0.14 (0.07)	0.19 (0.09)
Self-selected speed (m/s)	0.49 (0.23)	

Appendix B. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.clinbiomech.2020.01.005.

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