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The influence of locomotor training on dynamic balance during steady-state walking post-stroke

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ABSTRACT

Slow walking speed and lack of balance control are common impairments post-stroke. While locomotor training often improves walking speed, its influence on dynamic balance is unclear. The goal of this study was to assess the influence of a locomotor training program on dynamic balance in individuals post-stroke during steady-state walking and determine if improvements in walking speed are associated with improved balance control. Kinematic and kinetic data were collected pre- and post-training from seventeen participants who completed a 12-week locomotor training program. Dynamic balance was quantified biomechanically (peak-to-peak range of frontal plane whole-body angular-momentum) and clinically (Berg-Balance-Scale and Dynamic-Gait-Index). To understand the underlying biomechanical mechanisms associated with changes in angular-momentum, foot placement and ground-reaction-forces were quantified. As a group, biomechanical assessments of dynamic balance did not reveal any improvements after locomotor training. However, improved dynamic balance post-training, observed in a sub-group of 10 participants (i.e., Responders), was associated with a narrowed paretic foot placement and higher paretic leg vertical ground-reaction-force impulse during late stance. Dynamic balance was not improved post-training in the remaining seven participants (i.e., Non-responders), who did not alter their foot placement and had an increased reliance on their nonparetic leg during weight-bearing. As a group, increased walking speed was not correlated with improved dynamic balance. However, a higher pre-training walking speed was associated with higher gains in dynamic balance post-training. These findings highlight the importance of the paretic leg weight bearing and mediolateral foot placement in improving frontal plane dynamic balance post-stroke.

1. Introduction

Post-stroke hemiparetic gait is characterized by slow walking speed, asymmetry (Olney and Richards, 1996) and balance disorders (Geurts et al., 2005). Thus, various methods of locomotor rehabilitation have been used to improve gait coordination (Hollands et al., 2012) and overall mobility. One intervention that has received much attention is locomotor training, which consists of walking on a treadmill with partial body-weight support (Hesse, 2008) and manual assistance from trainers, followed by overground training. Locomotor training is based on the task-specific repetitive treatment concept in post-stroke rehabilitation (Hesse et al., 1994) with a number of early studies reporting improved walking speed in individuals post-stroke (e.g., Hesse, 2008; Peurala et al., 2005). However, the effect of locomotor training on dynamic balance during walking remains unclear.

The efficacy of locomotor training has been investigated in a large clinical trial, Locomotor Experience Applied Post-stroke (LEAPS) (Duncan et al., 2011). This study demonstrated that locomotor training was not superior to a home-based strength and balance program. Locomotor training emphasized repetitive stepping practice on the treadmill and overground, but did not include progressive balance-specific training, yet both groups improved in Berg-Balance-Scale similarly one year post-stroke. However, no biomechanical data were available to elucidate the mechanisms underlying these findings. Later, Bowden et al. (2013) performed locomotor training on individuals with chronic post-stroke hemiparesis, following a nearly identical protocol as in the LEAPS trial

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and found similar walking speed increases despite their participants being in the chronic phase (>6 months post-stroke). In addition, they found a significant increase in the Berg-Balance-Scale post-training in all participants as a group as well as in those who increased their walking speed post-training. Further, they found no relationships between changes in the Berg-Balance-Scale and changes in the walking speed from pre- to post-training. However, dynamic balance was not quantified using objective biomechanical methods and the underlying mechanisms for any observed changes in balance were not identified. Given that at least 50% of stroke survivors experience falls within one year of their stroke (e.g., Ashburn et al., 2008), further investigation into the effectiveness of locomotor training on dynamic balance using quantitative methods is warranted.

To maintain dynamic balance, the sum of linear and angular momenta of all the body segments about the body center-of-mass (i.e., whole-body angular-momentum) needs to be regulated by the net external moment (e.g., Pijnappels et al., 2005). The net external moment is generated by the distance between the body center-of-mass and center-of-pressure (in each foot) along with the ground-reaction-forces (Fig. 1). Thus, both proper foot placement and generation of appropriate ground-reaction-forces are critical elements for maintaining dynamic balance (e.g., Silverman and Neptune, 2011).

Previous research has shown that frontal plane movements require greater active control than sagittal plane movements (Bauby and Kuo, 2000). In a recent study, we assessed dynamic balance using the analysis of whole-body angular-momentum in each of the three anatomical planes and during a variety of walking tasks in individuals post-stroke and in healthy adults (Vistamehr et al., 2018). Interestingly, we identified significant balance deficits in the frontal plane in individuals post-stroke, manifested in a higher peak-to-peak range of angular-momentum. Furthermore, the rate of change of frontal plane angular-momentum during steady-state walking has been negatively correlated with the clinical balance scores (e.g., Berg-Balance-Scale and Dynamic-Gait-Index) (Nott et al., 2014), demonstrating that a higher range of angular-momentum is associated with poorer balance control (Nott et al., 2014; Vistamehr et al., 2016). However, no study has assessed if locomotor training improves dynamic balance in the frontal plane through improved regulation of whole-body angular-momentum. Additionally, it is not clear if there are any relationships between pre-training walking speed and improved balance control after locomotor training.

Thus, the purpose of this study was to assess the influence of a 12-week locomotor training program (Bowden et al., 2013) on dynamic balance during steady-state walking in individuals post-stroke and identify the underlying biomechanical mechanisms associated with observed changes. In addition, we examined whether changes in the frontal plane dynamic balance were associated with either pre-training walking speed or subsequent changes in walking speed. We hypothesized that locomotor training would improve dynamic balance and that the observed improvements would be associated with modified paretic foot placement and increased paretic leg ground-reaction-force output. We also hypothesized that both pre-training and subsequent changes in walking speed would be associated with improved dynamic balance from pre- to post-training.

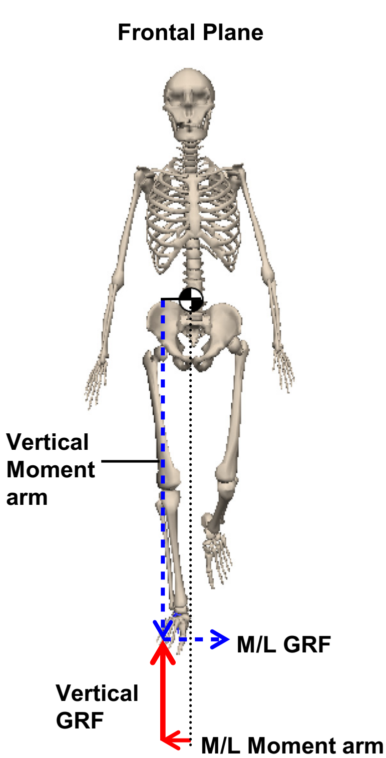

Fig. 1. The net external moment components in the frontal plane. Whole-body CoM is shown with '◐'. The GRF vectors and their corresponding moment arms appear in the same color. The higher magnitude of the vertical GRF compared to other components (after normalizing moment arms and GRFs by body height and weight, respectively) is highlighted by the line thickness.

2. Methods

From a previous study (Bowden et al., 2013), a subgroup of 17 individuals with post-stroke hemiparesis (11 left hemiparesis; age: 56.4 ± 12.4 years; 4 females) who had complete kinematic and ground-reaction-force (GRF) data sets pre- and post-training were selected. These individuals participated in a 12-week locomotor training program. In order to characterize the clinical profile of these participants, clinical assessments were conducted pre- and post-training. These assessments included self-selected comfortable (SS) and fastest-comfortable (FC) overground walking speeds, lower extremity Fugl-Meyer, Berg-Balance-Scale (BBS), Dynamic-Gait-Index (DGI), Activities-Specific Balance Confidence (ABC), and 6 min walk test (6MWT). One participant had a BBS < 45 and 16 participants had a DGI < 19, which indicate a higher risk of falling (e.g., Berg et al., 1995; Shumway-Cook et al., 1997). Information regarding the training procedure and subject inclusion criteria were previously described in detail (Bowden et al., 2013). The training sessions followed the protocol used in the LEAPS clinical trial (Duncan et al., 2007) and occurred 3 times per week, with each session including 20 min of walking on a treadmill with partial body-weight support while physical therapists provided manual step and postural training. The treadmill walking was followed by 10–20 min of overground training. The study protocol and consent form were approved by an Institutional Review Board and all participants provided informed, written consent prior to study participation.

Three-dimensional kinematics and GRFs were collected within one week of training initiation (pre) and completion (post). Participants walked on a split-belt instrumented treadmill (Techmachine, Andrezieux Boutheon, France) for multiple 30-second trials at their SS walking speed pre-training. During post-training data collection, each participant walked at a speed matched to their pre-training SS walking speed to control for changes in kinematics and kinetics related to changes in speed. Kinematic data were collected at 100 Hz using a 12-camera motion capture system (VICON, Los Angeles, USA) and GRFs were recorded at 2000 Hz. The kinematic and GRF data were low pass filtered using a fourth-order

Butterworth filter with cutoff frequencies of 6 Hz and 20 Hz, respectively. A 13-segment inverse dynamics model (C-Motion, Inc., Germantown, MD) was used to calculate center-of-pressure (CoP), center-of-mass (CoM) position and angular-momentum for each segment. At each time step of the gait cycle, whole-body angular-momentum (H) about the CoM was calculated as:

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

where $\vec{r}_i^{\,COM}$ and $\vec{v}_i^{\,COM}$ 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. $\vec{\omega_i}$, m_i and I_i are the angular velocity vector, and mass and moment of inertia of the i-th segment, respectively, and n is the number of segments. Whole-body angular-momentum was normalized by the product of subject mass, height and $\sqrt{g \cdot l}$, where $g = 9.81$ m/s^2 and l is the subject height. The term $\sqrt{g \cdot l}$ has units of m/s and provides a normalization technique similar to the concept of Froude number (e.g., Vaughan and O'Malley, 2005).

Dynamic balance in the frontal plane was assessed for each participant pre- and post-training using the range of whole-body angular-momentum (H_R) in the frontal plane, which was calculated as the difference between the minimum and maximum values of whole-body angular-momentum over the entire gait cycle. To assess the influence of locomotor training on dynamic balance, H_R for the entire group was compared between pre- and post-training using a paired t-test ($p < 0.05$). In addition, participants were stratified into two groups based on the changes in H_R between pre- and post-training. These groups consisted of Responders ($(H_R)_{Pre} - (H_R)_{Post} > 0$) and Non-responders ($(H_R)_{Pre} - (H_R)_{Post} < 0$). H_R between pre- and post-training was compared within each group using a paired t-test ($p < 0.05$). A decreased H_R post-training suggested improved dynamic balance control. To further understand the underlying biomechanical mechanisms used for maintaining dynamic balance in each group, GRFs and moment arm vectors were calculated and normalized by subject weight and height, respectively. At each time step, the net external moment (which is equal to the time rate of change of angular-momentum) is determined as:

$$\vec{M}_{ext} = \vec{r} \times \overrightarrow{GRF}$$

where $\vec{r}$ is the moment arm vector from the body CoM to CoP and $\overrightarrow{GRF}$ is the vector of GRFs (Fig. 1). Only the mediolateral and vertical GRFs and moment arms were analyzed as these are the only moment components regulating H in the frontal plane (Fig. 1). Both the GRF peak values and impulses were calculated in early (0–50%) and late (51–100%) stance to capture the instantaneous and average changes in the GRFs between pre- and post-training. GRF impulses were calculated using the time integral of GRFs during early and late stance. In order to analyze foot placement, the peak moment arm components as well as step width and step width variability were calculated during stance. Each moment component was compared between pre- and post-training using a paired t-test within each group. When significant differences ($p < 0.05$) were found, Pearson correlation analyses were performed between moment components and H_R post-training to identify the underlying mechanisms associated with maintaining dynamic balance within each group. To minimize the incidence of false negative findings (Perneger, 1998), multiple comparisons were corrected with a Benjamini-Hochberg post-hoc adjustment (Benjamini and Hochberg, 1995). In addition, to determine if improved walking speed was associated with improved dynamic balance, Pearson correlation analysis was conducted between changes in SS and FC walking speed and changes in H_R (from pre- to post-training). To determine whether the pre-training SS and FC walking speeds were indicators of improvement in dynamic balance post-training, Pearson correlation analysis was conducted between SS and FC walking speeds pre-training and the change in H_R (from pre- to post-training). To gain further insight into the clinical characteristics of the participants in each group, their clinical scores were compared between pre- and post-training using a paired t-test.

3. Results

3.1. The overall group

As a group ($n = 17$), there were no significant changes ($p = 0.313$) in H_R from pre- (0.0136 ± 0.0037) to post- (0.0137 ± 0.0038) training. However, both BBS (48.6 ± 2.9 pre; 51.1 ± 3.4 post; $p = 0.0043$) and DGI (13.5 ± 3.1 pre; 16.3 ± 2.9 post; $p = 0.0034$) increased significantly post-training.

3.2. Responders

Responders ($n = 10$, Table 1), decreased their H_R by 5% ($p = 0.007$) from pre- to post-training (Table 2). Further, they decreased their paretic leg mediolateral moment arm by 8.5% ($p = 0.002$), step width by 5% ($p = 0.03$), and step width variability by 29% ($p = 0.001$) post-training, while the vertical GRF impulses in the paretic leg increased by 10% ($p = 0.01$) during early and late stance (Table 2). H_R had a non-significant positive correlation with paretic leg moment arm ($r = 0.56$, $p = 0.09$) and a strong significant inverse correlation with the paretic leg vertical impulse during late stance ($r = -0.87$, $p = 0.001$). There were no changes in the vertical moment arm or mediolateral GRFs. In addition, SS and FC walking speeds as well as the BBS, DGI, ABC and 6MWT significantly improved post-training (Table 3).

3.3. Non-responders

Non-responders ($n = 7$, Table 1), increased their H_R by 9% ($p = 0.003$) from pre- to post-training (Table 2). Further, the non-

Table 1
Participant characteristics: gender, age, affected side, time since stroke, overground self-selected (SS) and fastest-comfortable (FC) walking speeds, and lower extremity Fugl-Meyer (FMA) corresponding to pre-training. Dynamic balance improved in Responders but did not improve in Non-responders post-training.

	Gender	Age	Side	Months since stroke	SS speed (m/s)	FC speed (m/s)	FMA
Responders							
1	M	48	L	59	0.71	1.11	25
2	F	45	L	11	0.51	0.90	27
3	F	74	R	8	0.79	1.15	31
4	M	56	L	12	0.76	0.99	24
5	M	54	L	26	0.76	1.06	21
6	M	74	R	22	0.48	0.85	21
7	M	57	R	32	0.50	1.06	20
8	M	68	L	17	0.63	0.90	23
9	M	62	L	56	0.60	0.82	30
10	M	43	R	11	0.70	1.01	23
Mean		58.1		25.4	0.64	0.98	24.5
SD		11.3		18.5	0.12	0.11	3.8
Non-responders							
1	M	46	L	9	0.54	0.74	21
2	M	64	L	12	0.43	0.53	19
3	M	44	L	10	0.50	0.73	27
4	F	31	R	10	0.33	0.73	18
5	M	57	L	27	0.44	0.57	17
6	F	62	L	17	0.43	0.77	25
7	M	73	R	7	0.37	0.56	21
Mean		53.9		13.1	0.43	0.66	21.1
SD		14.3		6.9	0.07	0.10	3.7

Table 2
Pre- and post-training biomechanical quantities mean (SD) for Responders (improved dynamic balance) and Non-responders (did not improve dynamic balance). Biomechanical quantities include: normalized range of whole-body angular-momentum (H_R) in the frontal plane, foot placement components normalized by body height (BH), ground reaction force (GRF) peaks and impulses normalized by body weight (BW). Significant differences (p < 0.05) between pre-and post-training are shown in **bold**.

	Responders Pre	Responders Post	*p*	Non-responders Pre	Non-responders Post	*p*
Whole-body angular-momentum (-)						
H_R - frontal plane	0.0135 (0.0036)	0.0129 (0.0033)	**0.0066**	0.0136 (0.0042)	0.0148 (0.0044)	**0.0035**
Foot placement (*m/BH*)						
Vertical moment arm – paretic leg	0.578 (0.008)	0.577 (0.012)	0.610	0.579 (0.008)	0.575 (0.009)	0.110
Vertical moment arm – nonparetic leg	0.585 (0.009)	0.583 (0.014)	0.454	0.587 (0.008)	0.585 (0.011)	0.416
M/L moment arm – paretic leg	0.091 (0.017)	0.083 (0.015)	**0.002**	0.108 (0.029)	0.099 (0.019)	0.401
M/L moment arm – nonparetic leg	0.106 (0.080)	0.119 (0.068)	0.538	0.087 (0.043)	0.065 (0.019)	0.378
Step width	0.136 (0.020)	0.129 (0.017)	**0.029**	0.143 (0.028)	0.138 (0.028)	0.474
Step width variability	0.0095 (0.0019)	0.0068 (0.0016)	**0.001**	0.0135 (0.0048)	0.0115 (0.0044)	0.143
Peak GRFs (*N/BW*)						
Vertical early stance – paretic leg	1.024 (0.083)	1.054 (0.105)	0.161	0.993 (0.085)	1.021 (0.056)	0.199
Vertical early stance – nonparetic leg	0.990 (0.041)	0.995 (0.066)	0.818	0.964 (0.020)	1.002 (0.026)	**0.024**
Vertical late stance – paretic leg	1.013 (0.039)	1.034 (0.074)	0.247	0.962 (0.087)	0.987 (0.052)	0.224
Vertical late stance – nonparetic leg	1.011 (0.037)	1.031 (0.057)	0.153	0.983 (0.034)	1.015 (0.027)	**0.012**
M/L early stance – paretic leg	0.087 (0.017)	0.080 (0.013)	0.129	0.085 (0.029)	0.085 (0.023)	0.924
M/L early stance – nonparetic leg	0.077 (0.015)	0.074 (0.006)	0.391	0.072 (0.019)	0.081 (0.022)	0.052
M/L late stance – paretic leg	0.089 (0.015)	0.083 (0.009)	0.069	0.085 (0.027)	0.083 (0.019)	0.644
M/L late stance – nonparetic leg	0.075 (0.011)	0.077 (0.011)	0.556	0.061 (0.014)	0.069 (0.017)	**0.046**
GRF impulses (*N.s/BW*)						
Vertical early stance – paretic leg	0.269 (0.072)	0.297 (0.079)	**0.013**	0.252 (0.032)	0.289 (0.071)	0.068
Vertical early stance – nonparetic leg	0.313 (0.081)	0.320 (0.092)	0.567	0.354 (0.106)	0.382 (0.124)	0.126
Vertical late stance – paretic leg	0.358 (0.108)	0.397 (0.105)	**0.010**	0.377 (0.134)	0.393 (0.142)	0.584
Vertical late stance – nonparetic leg	0.537 (0.114)	0.566 (0.111)	0.076	0.632 (0.071)	0.650 (0.087)	0.373
M/L early stance – paretic leg	0.020 (0.007)	0.021 (0.008)	0.352	0.019 (0.009)	0.022 (0.012)	0.117
M/L early stance – nonparetic leg	0.018 (0.006)	0.018 (0.006)	0.474	0.021 (0.010)	0.024 (0.012)	**0.022**
M/L late stance – paretic leg	0.029 (0.005)	0.029 (0.006)	0.811	0.032 (0.010)	0.033 (0.010)	0.423
M/L late stance – nonparetic leg	0.032 (0.009)	0.034 (0.010)	0.106	0.030 (0.008)	0.033 (0.008)	0.357

Table 3
Clinical characteristics pre- and post-training for Responders (improved dynamic balance) and Non-responders (did not improve dynamic balance). Overground self-selected (SS) and fastest-comfortable (FC) walking speeds as well as lower extremity Fugl-Meyer, Berg-Balance-Scale (BBS), Dynamic-Gait-Index (DGI), Activities-specific Balance Confidence scale (ABC), and 6-minute walk test (6MWT) were assessed pre- and post-training. Significant differences (p < 0.05) between pre-and post-training are shown in **bold**.

	Responders pre	Responders post	*p*	Non-responders pre	Non-responders post	*p*
SS walking speed	0.63 (0.14)	0.93 (0.12)	**0.0002**	0.43 (0.07)	0.61 (0.09)	**0.0003**
FC walking speed	0.98 (0.11)	1.24 (0.12)	**0.0002**	0.66 (0.10)	0.77 (0.16)	0.058
Fugl-Meyer	24.5 (3.78)	26.1 (3.67)	0.062	21.1 (3.67)	21.7 (4.72)	0.27
BBS	49.2 (2.70)	52.6 (1.84)	**0.003**	47.9 (3.13)	48.9 (3.98)	0.240
DGI	14.0 (3.27)	17.3 (2.50)	**0.015**	12.7 (2.81)	14.5 (2.95)	0.063
ABC	72.0 (11.92)	82.1 (12.19)	**0.009**	66.9 (15.81)	76.7 (14.38)	0.106
6MWT	772.8 (268.71)	1020.6 (149.72)	**0.005**	526.6 (165.48)	674.5 (110.48)	0.051

paretic leg vertical GRF peak increased during early stance by 4% (p = 0.02) and late stance by 3.2% (p = 0.01) (Table 2). Also, Non-responders increased their nonparetic leg mediolateral GRF peak during late stance by 13% (p = 0.046) and the mediolateral GRF impulse during early stance by 14% (p = 0.022) post-training. Although there were no significant differences in the foot placement, H_R had a positive correlation with paretic leg mediolateral moment arm (r = 0.83, p = 0.033) and an inverse correlation with the paretic leg vertical GRF impulse during late stance (r = −0.83, p = 0.019) post-training. In addition, H_R had a positive correlation with the nonparetic leg vertical GRF impulse during early stance (r = 0.82, p = 0.015). Further, aside from the SS walking speed, there were no other significant improvements in the clinical scores (Table 3).

3.4. Walking speed and dynamic balance

There were no significant correlations between changes in H_R and increased self-selected (r = 0.47, p = 0.067) and fastest-comfortable (r = 0.43, p = 0.08) walking speeds (Fig. 2). However, changes in H_R from pre- to post-training were positively correlated with the pre-training self-selected (r = 0.63, p = 0.012) and fastest-comfortable (r = 0.75, p = 0.0016) walking speeds (Fig. 3).

4. Discussion

The purpose of this study was to assess the influence of a 12-week locomotor training program on dynamic balance in individuals post-stroke and understand the underlying biomechanical mechanisms for achieving any observed improvements in dynamic balance. In addition, we investigated if both pre-training and improved walking speeds were correlated with changes in dynamic balance from pre- to post-training.

4.1. The influence of locomotor training on dynamic balance

Our hypothesis that locomotor training would improve dynamic balance was not supported as a group since the regulation of frontal plane angular momentum did not improve from pre- to post-training. Interestingly, both BBS and DGI improved significantly post-training. However, only 6 out of 17 individuals met

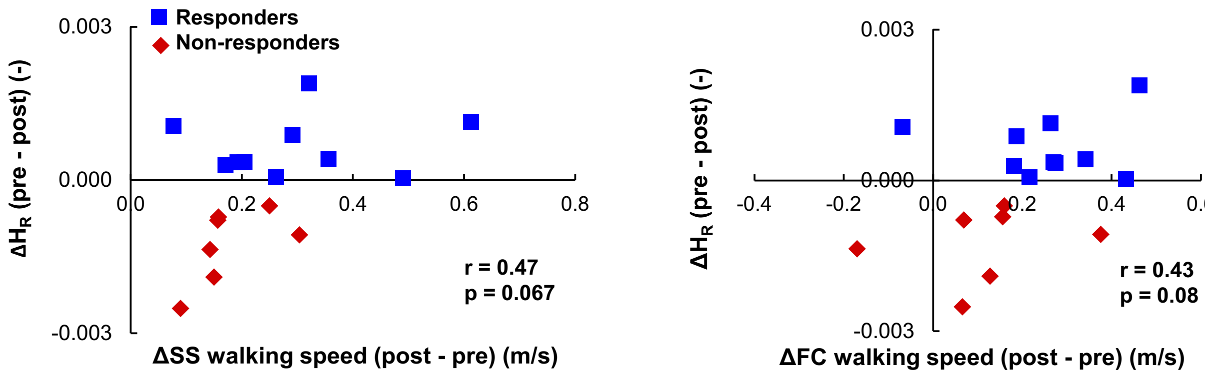

Fig. 2. Pearson correlation between improvements in the: self-selected (ΔSS); fastest-comfortable (ΔFC) walking speed and improvements in the frontal plane range of whole-body angular-momentum (ΔH_R) from pre- to post-training. Positive ΔSS, ΔFC, and ΔH_R values indicate improvement. Responders (improved dynamic balance) are shown with '■' and Non-responders (did not improve dynamic balance) are shown with '♦'. Significant correlations ($p < 0.05$) are shown with a '∗'.

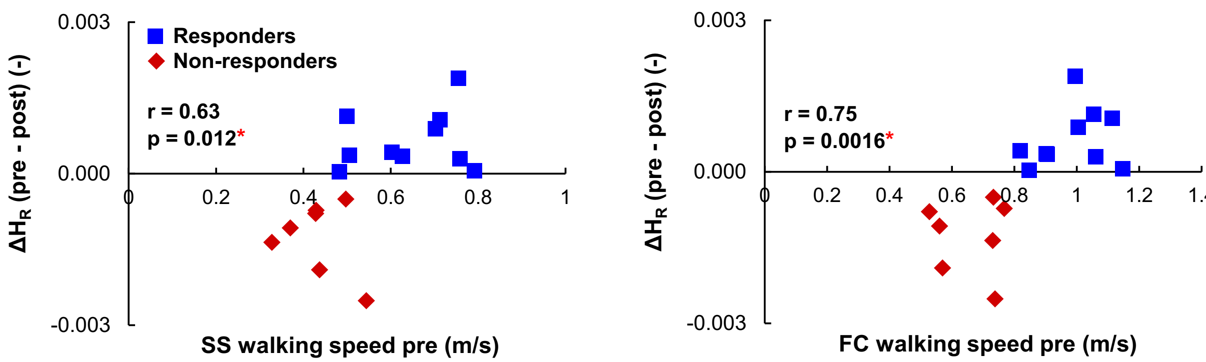

Fig. 3. Pearson correlation between the: self-selected (SS); fastest-comfortable (FC) walking speed pre-training and the change in the frontal plane range of whole-body angular-momentum (ΔH_R) from pre- to post-training. Responders (improved dynamic balance) are shown with '■' and Non-responders (did not improve dynamic balance) are shown with '♦'. Significant correlations ($p < 0.05$) are shown with a '∗'.

the BBS minimal detectable change of 2.5 points (Liston and Brouwer, 1996), and 9 out of 17 individuals met the DGI minimal detectable change of 2.6 points (Jonsdottir and Cattaneo, 2007). Thus, the observed statistically significant increases in the clinical balance scores may not necessarily reflect values outside measurement errors.

Only a sub-group (Responders) of the study participants improved their dynamic balance through improved regulation of H in the frontal plane (Table 1). By contrast, the regulation of H did not improve in Non-responders. Further, four out of seven Non-responders in this study were identified as those who had a clinically meaningful increase (0.16 m/s) in their walking speed (Bowden et al., 2013). That is, improved walking speed was not associated with improved dynamic balance (Fig. 2). However, changes in the regulation of H from pre- to post-training were associated with the pre-training SS and FC walking speeds (Fig. 3), which partially supports our second hypothesis that a faster pre-training walking speed would be associated with greater improvements in dynamic balance post-training. Although both SS and FC pre-training walking speeds were correlated with changes in dynamic balance, FC walking speed was a stronger and more significant indicator of changes in H_R (Fig. 3). In fact, all individuals with a pre-training FC walking speed greater than 0.8 m/s, intriguingly the same as the oft-used threshold for community walking ability (Perry et al., 1995), improved their balance (i.e., Responders). The effect of locomotor training on dynamic balance in individuals with severe walking impairment (identified by slow walking speed) is still unclear. Plummer et al. (2007) investigated the effect of stroke severity on locomotor recovery in a pilot study which was a precursor to the LEAPS clinical trial. Six participants completed 36 sessions of locomotor training followed by overground training. Three participants had initial moderate (>0.4 m/s and < 0.8 m/s) gait speed and the remaining three had severe (<0.4 m/s) walking impairment. After locomotor training, BBS improved in all the participants. However, paretic propulsion, an important biomechanical outcome in post-stroke recovery, increased (improved) only in two of the individuals with moderate gait speed. Moreover, paretic propulsion decreased in one individual with severe impairment due to more reliance on the nonparetic leg. Later, the LEAPS investigative team reported significant improvements in BBS after locomotor training (n = 139) regardless of the severity of initial walking impairment (Nadeau et al., 2013). Thus, based on the BBS assessments, individuals with slow walking speed would benefit from locomotor training in improving their balance. However, there is lack of biomechanical evidence to support this finding or to explain the mechanisms through which these individuals achieved any improvements in balance control during walking.

The observed improvements in dynamic balance assessed using whole-body angular-momentum were generally consistent with those assessed using clinical balance scores in the Responder group (Table 3). One of the advantages of using clinical balance scores is identifying fall risks. Previously, Nott et al. (2014) showed that poor regulation of angular-momentum during paretic single-leg stance was associated with poor BBS scores below the fall threshold. However, in this study, there were discrepancies between BBS and DGI assessments in identifying fall risks. Pre-training, only one participant had a BBS score below the fall threshold (<45), while 16 out of 17 participants had DGI scores below the fall threshold (<19). Further, post-training, there were two fallers based on BBS, while there were 12 fallers based on the DGI assessment. Although clinical balance measures provide a quick global assessment of overall balance performance, they may not be as sensitive as biomechanical measures to changes in the regulation of dynamic balance during walking. In addition, they cannot provide insight into the underlying biomechanical mechanisms of balance control. In this study, the analyses of angular momentum and the external moment components within each sub-group revealed how specific characteristics in foot placement and GRF generation affected dynamic balance pre- and post-training (Table 2). Thus, in addition to the clinical balance measures, the analysis of angular momentum can be a valuable assessment tool to gain insight into the effectiveness of novel training programs aimed at improving dynamic balance.

4.1.1. Foot placement

Prior studies have shown that mediolateral foot placement is an effective way to control frontal plane balance (e.g., Zissimopoulos et al., 2014). Others have used sensory electrical stimulation to improve paretic leg foot placement in individuals with chronic stroke (Walker et al., 2014). Previous research has shown that mediolateral foot placement requires active recruitment of the sensory-motor processes (Hof et al., 2010). Further, healthy individuals drive active muscular control of their mediolateral foot placement by sensing the mechanical state of their stance leg. However, this mechanism is shown to be disrupted in individuals post-stroke who are at higher fall risk (Dean and Kautz, 2015). In addition, individuals post-stroke have shown reduced ability to control step width and foot placement variability, particularly in targeted step tasks with decreased step width target size (Reissman and Dhaher, 2015). In this study, we observed that those who improved their dynamic balance post-training modified their paretic foot placement closer to the midline and reduced their step width and step width variability post-training (Table 2). On the contrary, there were no changes in the foot placement characteristics of those who did not improve their balance post-training. Also, placing the paretic foot farther away laterally from the body CoM was associated with a higher range of angular-momentum in the frontal plane. A prior study has shown that wider paretic foot placement is strongly related to the lower weight-bearing of the paretic leg (Balasubramanian et al., 2010). Further, a wider paretic foot placement along with the gravitational load creates a destabilizing moment about the body CoM, which acts to rotate the body towards the nonparetic leg. In order to maintain dynamic balance, the stance leg hip abductors generate a counteracting moment (MacKinnon and Winter, 1993; Neptune and McGowan, 2016). However, previous studies have reported impaired paretic leg hip abductor muscle activity post-stroke (e.g., Kirker et al., 2000) and identified inaccuracy in the paretic hip abduction as a predictor of wider paretic step (Dean et al., 2017). Others have suggested hip abductor strengthening to improve lateral stability in individuals post-stroke (Mercer et al., 2009). Thus, hip abductor weakness may also hinder counteracting the external moment generated by a wider lateral moment arm, which further highlights the importance of lateral foot placement in maintaining dynamic balance.

4.1.2. Ground reaction forces

Aside from foot placement, GRFs greatly influence the regulation of whole-body angular-momentum. A higher vertical GRF impulse during late stance from the paretic leg (i.e., higher weight bearing) was correlated with lower ranges of whole-body angular-momentum, suggesting better balance control (Responders). On the contrary, higher compensations from the nonparetic leg in generating vertical GRFs were associated with higher ranges of whole-body angular-momentum suggesting poor balance control (Non-responders). This finding is similar to that in unilateral below-knee amputees where higher range of angular-momentum in the frontal plane was related to the reduced peak vertical GRFs in both the intact and residual legs during late stance (Silverman and Neptune, 2011). This similarity can be attributed to the ankle plantarflexors being primary contributors to body support throughout the single-leg stance (e.g., Anderson and Pandy, 2003; Neptune et al., 2001) and that individuals with post-stroke hemiparesis and unilateral below-knee amputation have similarly diminished plantarflexor output. In addition, the ankle plantarflexors, particularly soleus, are the primary contributors to the generation of vertical GRF impulses (McGowan et al., 2009). Most importantly, the ankle plantarflexors are critical contributors to the regulation of whole-body angular momentum in the frontal plane (Neptune and McGowan, 2016). Thus an increased paretic leg vertical GRF impulse in Responders may be related to potential improvement in paretic leg ankle plantarflexor output. By contrast, the increased range of H and reliance on the nonparetic leg in Non-responders may be related to compensations due to paretic leg ankle plantarflexor weakness. Lastly, our prior study has shown that during walking adaptability tasks such as obstacle negotiation, soleus activity was significantly lower in the paretic leg than healthy individuals (Vistamehr et al., 2018). Further, lower soleus activity during obstacle negotiation was associated with a higher range of H in the frontal plane, suggesting poor balance control. Thus, incorporating walking adaptability tasks such as obstacle negotiation into rehabilitation intervention as a supplement to locomotor training may be an effective approach to improve dynamic balance post-stroke.

A limitation of this study was that H_R has not been benchmarked against fall rates, and the H_R threshold values predicting a high risk of falls are unknown. Thus, even though Responders improved their dynamic balance, they may still be at high fall risks as indicated by their DGI scores. This study provided a framework for assessing the influence of locomotor training on relative changes in dynamic balance from pre- to post-training and has identified the associated underlying mechanisms. Future studies can build upon this work and assess these changes in light of fall risk. In addition, future work is needed to expand this investigation in larger number of participants pertaining to lower level household and higher level community walkers.

5. Conclusions

As a group, locomotor training did not improve dynamic balance, quantified by the regulation of angular momentum. However, Responders improved their frontal plane dynamic balance, primarily by narrowing their paretic foot placement and increasing their weight bearing on the paretic side. Further, Non-responders did not improve their dynamic balance, and their nonparetic leg compensations increased post-training. In addition, improved walking speed post-training was not correlated with improved dynamic balance. However, the pre-training self-selected and fastest-comfortable walking speeds were found to be good indicators of those who would and would not improve their dynamic balance post-training.

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