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Mechanisms used to increase peak propulsive force following 12-weeks of gait training in individuals poststroke

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Abstract

Current rehabilitation efforts for individuals poststroke focus on increasing walking speed because it is a predictor of community ambulation and participation. Greater propulsive force is required to increase walking speed. Previous studies have identified that trailing limb angle (TLA) and ankle moment are key factors to increases in propulsive force during gait. However, no studies have determined the relative contribution of these two factors to increase propulsive force following intervention. The purpose of this study was to quantify the relative contribution of ankle moment and TLA to increases in propulsive force following 12-weeks of gait training for individuals poststroke. Forty-five participants were assigned to 1 of 3 training groups: training at self-selected speeds (SS), at fastest comfortable speeds (Fast), and Fast with functional electrical stimulation (FastFES). For participants who gained paretic propulsive force following training, a biomechanical-based model previously developed for individuals poststroke was used to calculate the relative contributions of ankle moment and TLA. A two-way, mixed-model design, analysis of covariance adjusted for baseline walking speed was performed to analyze changes in TLA and ankle moment across groups. The model showed that TLA was the major contributor to increases in propulsive force following training. Although the paretic TLA increased from pre-training to post-training, no differences were observed between groups. In contrast, increases in paretic ankle moment were observed only in the FastFES group. Our findings suggested that specific targeting may be needed to increase ankle moment.

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Introduction

Stroke is the leading cause of long term disability in the United States (Go et al., 2014). Current rehabilitation efforts for individuals poststroke focus on increasing walking speed because it is the best predictor of community ambulation and participation in home and community mobility (Perry et al., 1995). Unfortunately, more than 60% of individuals who achieve independent ambulation still walk at speeds that are insufficient to function effectively in the community (i.e., <0.8 m/s) (Duncan et al., 2007; Hill et al., 1997). Consequently, assisting stroke survivors to increase their walking speeds is a major focus for current rehabilitation efforts. For individuals poststroke, it has been shown that propulsive force, defined as the anterior component of the ground reaction force (AGRF), generated from the paretic limb is correlated to walking speed and increases with increased speed in individuals poststroke (Bowden et al., 2006; Kesar et al., 2011). In addition, the propulsive force from the paretic extremity is a direct measure of the paretic limb's output and, therefore, likely reflects actual neural recovery that restores the affected neural systems. Previous studies have shown that decreased propulsive force contributes to a greater energy cost of walking, and is correlated with hemiparetic severity, walking speed, and gait asymmetry in individuals poststroke. (Bowden et al., 2006) (Balasubramanian et al., 2007) Thus, recent research in poststroke gait rehabilitation have emphasized the importance of improving propulsive force (Awad et al., 2014).

It has been shown that propulsive force is associated with two critical factors: ankle plantarflexion moment and the position of the foot relative to the body center of mass (COM) (Hsiao et al., 2015a; Peterson et al., 2010). For individuals poststroke, the paretic ankle plantarflexor muscle's isokinetic moment has the greatest correlation to self-selected walking speed compared with all other lower extremity muscle groups (Kim and Janice, 2003). Ankle moment has also been correlated to paretic propulsion in a cross-sectional study in stroke survivors (Bowden et al., 2006). Thus, the ability to generate ankle plantarflexion moment is critical to forward propulsion. Another critical factor for propulsive force is the position of the foot relative to the body COM during late stance. This relative position affects the orientation of the ground reaction force (GRF) vector and, therefore, determines the proportion of the GRF being distributed anteriorly (Peterson et al., 2010). A previous study correlated propulsive force to leg extension, defined as the angle between the vertical and the vector joining the body COM to the foot segment COM (Peterson et al., 2010). Other studies measured trailing limb angle (TLA), defined as the angle between the vertical and the vector joining the 5th metatarsal joint and the greater trochanter, and found that this angle increases when walking speed increases during able-bodied (Hsiao et al., 2015a) as well as poststroke gait speed modulation (Hsiao et al., 2015b; Tyrell et al., 2011).

Using a biomechanical-based model, our lab has previously studied the relative contribution of TLA versus ankle moment to changes in propulsive force during speed modulation within a session in individuals poststroke (Hsiao et al., 2015b). A previous study reported changes in propulsive forces and positive joint work within the paretic and non-paretic legs following 8-weeks of body weight support training (Combs et al., 2012). Other studies reported

improvements in propulsion symmetry following a 12-week locomotor intervention incorporating treadmill training with body weight support and manual trainers during overground walking (Bowden et al., 2013; Routson et al., 2013). Using musculoskeletal simulation, Allen and colleagues found that a major source of increased propulsion following 12-weeks of gait training was the ankle plantarflexors (Allen et al., 2014). However, only 2 subjects were analyzed in their study and only 1 subject increased paretic propulsion following training. The changes of TLA were not reported in their study and warrants investigation. A better understanding of the mechanism used to gain propulsive force following gait training would inform future rehabilitation efforts. Thus, the primary purpose of this study was to determine the relative contribution of changes in TLA and ankle moment to increases in propulsive force following 12-weeks of gait training.

Because increased TLA was previously reported during faster walking speed, and increases in the activation of the plantarflexors is achievable through functional electrical stimulation, our laboratory has designed and completed a randomized controlled trial that incorporated fast speed treadmill gait training with functional electrical stimulation (FES) to the ankle musculatures (Kesar et al., 2011)(Awad et al., 2014)(Awad et al., 2015, in press). Specifically, three training groups were studied: (1) walking training at a self-selected speed (SS), (2) walking training at the fastest speed that subjects could maintain for at least 4 minutes (FAST), and (3) FAST with the addition of FES applied to the paretic limb dorsiflexors and plantarflexors (FastFES). Herein, we report the changes in TLA, ankle moment, and propulsive force in all three groups. Findings of this study would allow for the design of rehabilitation strategies targeting improved paretic propulsion.

Methods

Participants

Forty-five participants (age, 58.9 ± 12.2 (mean $\pm$ SD) years; 14 female; 16 right hemiparetic; self-selected walking speed, 0.69 ± 0.34 m/s; time since stroke, (median = 1.7, range = 0.4 – 30.5) years) with poststroke hemiparesis participated in this study (see Table 1 for inclusion/exclusion criteria). All participants signed informed consent forms that were approved by the Institutional Review Board of the University of Delaware.

Gait Evaluation

Gait analysis was conducted at baseline (pre) and after 12 weeks of gait training (post). An 8-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA, USA) was used to measure kinematic data as participants walked at their self-selected speeds on a split-belt treadmill (Bertec Corp., Columbus, OH, USA). The treadmill was instrumented with two, independent, 6 degree-of-freedom force plates that measured ground reaction forces at 1080 Hz. Self-selected walking speed was defined as the participant's comfortable overground walking speed during a 10-m walk test. Previous work has provided detailed description for the gait analysis setup (Awad et al., 2014; Kesar et al., 2014; Reisman et al., 2013). Participants wore an overhead harness system for safety and were allowed to use the handrails if they normally utilized an assistive device (i.e., a cane) or if they felt unsafe walking on a treadmill. Participants were only allowed to use handrails located at the side of

the treadmill, which mimicked walking with an assistive device. No bodyweight was supported by the harness and verbal instructions on using the handrails as little as possible were provided.

Training

Participants were randomly assigned to 1 of 3 training groups that were equivalent in structure: (1) walking training at a self-selected speed (SS), (2) walking training at the fastest speed that subjects could maintain for at least 4 minutes (FAST), and (3) FAST with the addition of functional electrical stimulation (FES) applied to the paretic limb dorsiflexors and plantarflexors (FastFES). Regardless of group, participants completed 3 sessions a week for a total of 12 weeks. Each training session comprised a total of 36 minutes (six 6-minute bouts). Bouts 1 to 5 consisted of treadmill walking and bout 6 consisted of overground walking. For the FastFES group, FES was delivered to ankle plantarflexor and dorsiflexor muscles during the first, third, and fifth minutes for bouts 1 to 5. Rest breaks of up to 5 minutes were provided between walking bouts. More detail on the FastFES intervention may be found in previous work from our laboratory (Awad et al., 2014; Reisman et al., 2013).

Biomechanical-Based Model

Detailed description of this biomechanical-based model can be found in (Hsiao et al., 2015b). Briefly, AGRF (F_a) is modeled as a function of ankle moment (M_a), TLA, and lever arm length (d) between the GRF and the ankle joint (Figure 1):

$$F_a = \frac{1}{d} M_a \sin(TLA). \quad (1)$$

For negative values in model predicted AGRF, a value of 0 was used when validating the model (no propulsive force). Applying Eq. 1 to the pre-training and post-training results, changes in propulsive force can be calculated as:

$$\Delta F_a = \frac{1}{d_1 + \Delta d} \left[M_{a1} \Delta \sin TLA + \Delta M_a \sin TLA_1 + \Delta M_a \Delta \sin TLA - \left(\frac{\Delta d}{d_1} \right) M_{a1} \sin TLA_1 \right], \quad (2)$$

where “ Δ ” denotes the change from pre to post-training, subscript “1” denotes the value at pre-training. As described in our previous study, four terms represent the contributions to changes in propulsive force: $M_{a1} \sin TLA$ represents the contribution of changes in TLA, $\Delta M_a \sin TLA_1$ represents the contribution from changes in ankle moment, $\Delta M_a \Delta \sin TLA$ represents the contribution from the interaction between changes in TLA and ankle moment, and $\left(\frac{\Delta d}{d_1} \right) M_{a1} \sin TLA_1$ represents the contribution from changes in lever arm length. Each of the above terms was calculated and negative values were set to 0 (no contribution). The relative contribution of each term was calculated by dividing each term by the sum of all terms. Note that $\frac{1}{d_1 + \Delta d}$ would have no impact on the relative contributions because it exists in all terms (Hsiao et al., 2015b).

Data processing

Cortex and Visual3D (C-Motion Inc., Bethesda, MD, USA) were used for data processing. Kinematic and kinetic data were filtered using a bi-directional Butterworth low-pass filter at 6 and 30 Hz, respectively. Both peak AGRF (Nilsson et al. 1989, Franz et al. 2014, Campanin et al. 2009, Moore et al. 2015, Hanley et al. 2014) and AGRF impulse have been used in several previous studies to characterize propulsion. Because investigation from our laboratory has identified much stronger correlations between walking speed and peak anterior ground reaction force (AGRF) compared with AGRF impulse (force-time integral), peak AGRF was used in the present study (unpublished observations). Peak AGRF was measured as the maximum AGRF normalized to body weight. Peak AGRF (F_a) was defined as the maximum AGRF during stance between the onset of the anterior ground reaction forces and toe-off. Ankle moment (M_a) was defined as the ankle plantarflexion moment resolved into the shank coordinate system and was normalized to body weight. TLA was defined as the angle between the laboratory's vertical axis and the vector joining the greater trochanter with the center of pressure (COP) (Figure 1). d was the perpendicular distance from the ankle joint to the vector joining the greater trochanter with the COP. The AGRF, ankle moment, TLA, and d at the instant of peak AGRF were used in our model. All data were averaged across strides within the 30-second trials.

Statistical Analysis and Model Validation

Correlations between changes in walking speed and changes in propulsive force in each limb following training for all subjects were analyzed using Pearson's correlation coefficients (r). Next, because the model was previously validated during speed modulation within a session, and not across sessions, the correlation between the predicted and measured peak AGRF after forcing the intercept to 0 was used to evaluate the ability of the model to predict the changes in propulsive forces following training. In addition, paired t-tests were used to detect whether differences existed between the predicted and the experimental changes in peak AGRF.

For participants who increased paretic propulsive force following training, Eq. 2 was used to determine the relative contribution of changes in ankle moment and TLA to increases in propulsive force. It was observed that baseline walking speed of the FastFES group was half a standard deviation greater than the SS group, and walking speed is strongly related to ankle moment and TLA, therefore we included baseline walking speed in our analysis as a covariate. A two-way mixed design analysis of covariance (ANCOVA) comparing time (pre-training vs. post-training) $\times$ group (SS vs. Fast vs. FastFES) adjusting for baseline walking speed was performed to analyze changes in TLA and ankle moment. If the interaction was significant, the simple main effects were examined using pairwise comparisons with Fisher's LSD. Alpha was 0.05. All statistics were run using SPSS (version 22.0, SPSS, Inc.).

Results

Forty-five participants completed the 12-week gait training program. Due to technical issues, GRF data for 3 participants and ankle moment data for an additional 3 participants

were not available. Thus, only 39 participants' data were analyzed. On average, walking speed increased from 0.69 to 0.82 m/s following training. Paretic propulsive force increased 23% from pre-training (7.21 %BW) to post-training (8.85 %BW) and non-paretic propulsive force increased 17% from pre-training (12.14 %BW) to post-training (14.24 %BW). Changes in propulsive force from the paretic and the non-paretic limbs explained 52% and 36% of the variance in changes in self-selected walking speed, respectively (Figure 2).

We validated the model in both the paretic and non-paretic leg at pre-training and post-training (Figure 3). The model explained more than 75% of the variance in propulsive force with the slopes close to 1 at pre and post-training (Figure 3 A–D). This model also explained 68% of the variance in changes in propulsive force as a result of training (Figure 3 E–F). No significant differences were found between the predicted and the measured changes in propulsive forces (mean: predicted paretic=16.15 N, measured paretic=15.69 N, predicted non-paretic=21.98 N, and measured non-paretic=20.26 N; $t=0.19$, $p=0.85$ for the paretic and $t=-0.52$, $p=0.66$ for the non-paretic).

Of the 39 participants, 26 increased their propulsive force in the paretic limb following intervention for participants who increased paretic propulsive force as a result of intervention ($N=9$ for Fast, $N=8$ for FastFES, and $N=9$ for SS). The relative contributions of the four components in Eq. 2 for propulsive force were quantified for each limb (see Table 2). In the paretic limb, 21 participants showed greater independent contribution from changes in TLA to propulsive force than from ankle moment. Ten participants showed no independent contribution from changes in ankle moment to the increased propulsive force and 4 participants showed no independent contribution from changes in TLA to increases in propulsive force.

Of the 26 participants who increased propulsive force in their paretic limbs, 3 did not increase propulsive force in their non-paretic limb (participants #2, 10, and 26). For the remaining 23 participants, 13 participants showed greater independent contribution from changes in TLA to propulsive force than from ankle moment in their non-paretic limbs. Six participants showed no independent contribution from changes in ankle moment and 6 participants showed no independent contribution from changes in the TLA to increases in non-paretic propulsive force.

For 26 individuals who increased paretic propulsive force, results of ANCOVA showed a main effect for time and no effect for group in paretic TLA (Figure 4). A significant interaction between time and group was found for paretic ankle moment. Post-hoc pairwise comparisons revealed that the significant increases in paretic ankle moment were observed only for the FastFES group (Fast: $p=0.14$, FastFES: $p=0.03$, SS: $p=0.38$).

In the non-paretic limb, results of ANCOVA showed a main effect for time and no effects for group in non-paretic TLA. No main effects were found for the non-paretic ankle moment. No interactions between time and group were detected for the TLA or ankle moment.

Discussion

In this study, we validated our previously developed biomechanical-based model to predict propulsive force across sessions. Changes in TLA and ankle moment in each group were calculated and the model was applied to quantify the relative contributions of changes in TLA and ankle moment to increases in propulsion force as a result of 12-weeks of gait training. The major findings of this investigation were that on average, changes in TLA is the major contributor to increases in propulsive force. Indeed, for participants who increased paretic propulsive force following training, increases in paretic TLA were observed in all 3 groups. In contrast, increases in paretic ankle moment were observed only in the FastFES group.

Our model was able to predict propulsive force in each limb at both pre and post-training (Figure 3). The validity of the model was similar to the results from our previous study during within-session speed modulation in individuals poststroke (Hsiao et al., 2015b). The ability of our model to predict changes in AGRF across sessions as a result of our intervention was slightly lower compared to the ability of our model to predict changes within a session as a result of speed modulation (paretic limb: current $r^2=0.68$ vs. previous $r^2=0.78$, non-paretic limb: current $r^2=0.68$ vs. previous $r^2=0.76$) (Hsiao et al., 2015b). This could be due to the variation in marker placement across sessions. Because our model incorporates ankle moment and TLA, marker placement at the ankle joint and the greater trochanter would affect the AGRF predicted by the model. However, the actual AGRF measured from the force plate is independent from marker placement. For example, a more anteriorly placed greater trochanter marker would induce an increase in the calculation of TLA without the actual changes in limb posture and propulsive force. Thus, it is expected that the error between the model predicted and measured AGRF will be greater when studying changes across sessions. However, the model was still able to explain the majority of variance in changes in propulsive force across sessions.

The observed primary contribution from TLA to increases in propulsive force is consistent with our previous finding during speed modulation within a session for individuals poststroke (Hsiao et al., 2015b). Results from the present study suggest that TLA is a more modifiable component that contributes to increased propulsive force compared with ankle moment for poststroke gait training. Because TLA captures hip and knee extension during terminal stance, improvements in strength or coordination of muscles around the knee and hip joint would increase TLA. Our results could therefore indicate that restoration of propulsive force following stroke is accomplished primarily by the proximal muscles (i.e. hip and knee extensors), especially in the paretic limbs. A previous study found that participants with lower distal muscle function improved lower limb mechanics by increasing activity of proximal muscles leading to better hip and knee extension (Mulroy et al., 2010). Similarly, a musculoskeletal modeling study demonstrated that an individual with severe plantarflexor weakness would exhibit increased contributions from the paretic hip and knee extensors (Higginson et al., 2006). Thus, the greater contribution from increase in TLA could be a strategy to substitute for poor control of the ankle plantarflexor muscles.

Our results showed that for those individuals who increased paretic propulsive force, only the FastFES group demonstrated significant increases in ankle moment (Figure 4). A recent systematic review showed that FES produced increases in muscle strength and voluntary range of motion when used during walking in individuals poststroke (Kafri and Laufer, 2014). Similarly, a previous musculoskeletal simulation study reported increased activation of ankle plantarflexor muscles following a 12-week FES intervention (Knarr et al., 2013). Thus, gait training with FES appears to provide increases in propulsive force via increasing ankle moment. Although the average change in ankle moment is only approximately 1 Nm %BW in the FastFES group (Figure 4), this amount of improvement is still relatively substantial (>10% increase) considering that the initial ankle moment was less than 10 Nm %BW.

Our biomechanical-based model provided the ability to quantify the independent contributions from changes in TLA versus ankle moment to increases in propulsive force, as well as the contribution from the interaction of changes in these two factors. The independent contribution from each factor represents the contribution solely due to changes in that factor with the other factor remains at baseline value. In contrast, the interaction term represents the contribution from changes in TLA and ankle moment without considering the baseline value of each factor (Eq. 2). This term is negligible during speed modulation in able-bodied individuals because the changes in each of these two variables are small compared to their baseline values (Hsiao et al., 2015a). However, in our stroke participants, baseline measurements of TLA for some individuals were very small and, therefore, considerable contributions from the interaction term were observed in these participants. For example, participants #7, 9, 14, and 25 all had very low baseline TLAs. In addition, they all showed considerable increases in both TLA and ankle moment. Interestingly, these same 4 participants also showed the greatest improvements in walking speed (Table 2). A previous study reported that for individuals poststroke, following 6-weeks of bodyweight support treadmill training, those in the high-response group displayed greater increases in both hip extension angle and soleus muscle EMG activity during terminal stance (Mulroy et al., 2010). Allen and colleagues also reported increased contribution from the ankle plantarflexors and hamstrings to forward propulsion for the subject who increased paretic propulsion following 12-weeks of gait training with body weight support (Allen et al., 2014). Results from the present study also highlight the importance of targeting and improving both TLA and ankle moment during gait training in individuals poststroke. However, it is important to note that we did not find a relationship between the contribution from the interaction term and increases in walking speed. This is because individuals can increase propulsion via increasing only TLA or only ankle moment, as indicated in our model (Hsiao et al., 2015a). Indeed, participant #11 increased only ankle moment and participant #2 increased only TLA in their paretic limbs, yet both participants increased their walking speeds following training. It is possible that the observed differences in mechanisms adopted by each individual is due to differences in baseline characteristics. For example, for an individual with small TLA but sufficient ankle moment at baseline, a substantial amount of propulsive force could be gained by increasing only TLA without the need for increasing ankle moment.

It is worth noting that although changes in TLA and ankle moment are mathematically separable variables, they may not be physically independent. For example, increases in TLA resulting from increased hip or knee extension could result in increases in the lever arm length of the ground reaction forces (d) and, therefore, increase ankle moment. Thus, TLA and ankle moment can potentially impact each other. However, we did not find a direct correlation between changes in TLA and changes in ankle moment following training in the present study. There are multiple factors other than TLA that could affect ankle moment. For example, with increased TLA, some participants showed increased ankle plantarflexion angle and therefore decreased lever arm length (d). Thus, individuals poststroke may not show increases in ankle moment with increased TLA.

Interestingly, the mechanism for an individual to increase propulsive force can be different between limbs. For example, participants #12 and 16 showed greater contribution from the TLA compared with ankle moment to increases in propulsive force in their paretic limb; however, for both of these individuals, the contribution from M_a was greater than TLA in the non-paretic limb. In fact, eight of the 23 participants who increased propulsive force from both limbs showed opposite strategies to gain propulsive force between limbs. Similar to our findings, a previous simulation study also reported that different muscle groups contributed to increases in paretic versus the non-paretic propulsion following training (Allen et al., 2014). Sousa and colleagues investigated interlimb coordination during poststroke gait and found that the propulsive impulse of the paretic trailing limb was negatively correlated with the braking impulse of the leading limb during double support (Sousa et al., 2013). Thus, the strategy for each limb to increase propulsive force during push off could be affected not only by the baseline characteristics of each limb, but also the braking force generated from the other limb. (Bowden et al., 2006)

Study limitations

The present study showed *how* individuals increased propulsive force, but we did not explore *who* would respond to the intervention. Future investigations are needed to determine whether baseline characteristics could help to identify individuals who would most likely to respond to the intervention. Another potential limitation of this study is that our participants were allowed to use the handrails if needed. The use of handrails could influence gait patterns and force distribution. However, the use of a handrail more closely replicated walking with an assistive device for those subjects who used a cane.

Conclusions

This is the first study to quantify the relative contribution of TLA and ankle moment to increases in propulsive force following 12-weeks of gait training. We found that TLA was the primary contributor to increases in propulsive force. This finding highlights the importance of increasing paretic limb extension for gait rehabilitation in individuals poststroke. In addition, increases in TLA were observed in all 3 groups. In contrast, our finding that increases in paretic ankle moment were observed only in the FastFES group suggested that specific targeting may be needed to increase ankle moment. Moreover, there

is an apparent need to study subject-specific characteristics as large variations were observed both between individuals and between limbs.

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Figure 1.

Diagram of variables of interest. F_a was the anterior component of the ground reaction force. M_a was the ankle plantarflexion moment. **COP** was the center of pressure. TLA was measured as the angle between the laboratory's vertical axis and the vector joining the greater trochanter with the COP. d was the perpendicular distance from the ankle joint to the vector joining the greater trochanter with the COP. (Adapted from Hsiao et al., 2015b, Figure 1,)

Figure 2. Relationship between changes in propulsive force (AGRF) and changes in self-selected walking speed as a result of intervention (N=39).

Figure 3.

Relationships between model predicted propulsive force (AGRF) and experimental measured propulsive force (N=39). (A) Paretic propulsive force at pre-training. (B) Non-paretic propulsive force at pre-training. (C) Paretic propulsive force at post-training. (D) Non-paretic propulsive force at post-training. (E) Changes in paretic propulsive force from pre to post-training. (F) Changes in non-paretic propulsive force from pre to post-training.

Figure 4.

Changes in trailing limb angle (TLA) and ankle moment (M_a) for participants who increased paretic propulsive force as a result of intervention (N=9 for Fast, N=8 for FastFES, and N=9 for SS). A two-way mixed design analysis of covariance (ANCOVA) comparing time (pre-training vs. post-training) $\times$ group (SS vs. Fast vs. FastFES) adjusting for baseline walking speed was performed. (A) For paretic limb TLA, a main effect for time was found. (B) For non-paretic limb TLA, a main effect for time was found. (C) For paretic limb ankle moment, a significant interaction between time and group was found. Post-hoc pairwise comparisons revealed that the significant increases in paretic ankle moment were observed only for the FastFES group. (D) For non-paretic limb ankle moment, no main effects and no interactions between time and group were found. * = $P < 0.05$ (see text for details).

Table 1

Participants' Inclusion/Exclusion criteria

Inclusion criteria	Exclusion criteria
A single cortical or subcortical stroke A poststroke duration of at least 6 months The ability to ambulate without the assistance of another individual Sufficient cognitive function to follow instruction and communicate with the investigators The ability to walk for 6 minutes without orthotic support Sufficient passive dorsiflexion range of motion to position the ankle in a Neutral position with the knee extended Sufficient passive hip extension to extend the hip 10°	Had a history of multiple strokes Cerebellar stroke Lower extremity joint replacement Bone or joint problems that limited their ability to walk A resting heart rate outside of the range of 40 to 100 beats per minute A resting blood pressure outside of the range of 90/60 to 170/90 mm Hg Neglect Hemianopia Unexplained dizziness during the past 6 months Chest pain or shortness of breath without exertion

Table 2

Participants' characteristics and relative contributions from changes in each variables to increases in propulsive force as a result of 12-weeks of intervention.

Subject #	LEEM	Walking Speed (m/s)	Changes in propulsive forces (N)		Relative Contribution to Paretic Propulsion				Relative Contribution to Non-Paretic Propulsion				Group	
			Paretic	Non-Paretic	TLA _p	M _{a,p}	Int _p	d _p	TLA _{np}	M _{a,np}	Int _{np}	d _{np}		
1	19	0.58	0.16	19.46	1.56	0.94	0.01	0.03	0.02	0.82	0.14	0.04	0	Fast
2*	26	0.88	0.15	5.69	-18.46	1	0	0	0	-	-	-	-	Fast
3	27	0.96	0.07	29.70	14.71	0	1	0	0	0	0.82	0	0.18	Fast
4	34	0.88	0.18	19.66	43.77	0.06	0.74	0.01	0.19	0.30	0.63	0.07	0	Fast
5	32	1.01	0.23	22.43	28.08	0.73	0	0	0.27	0.88	0	0	0.12	Fast
6	26	1.06	0.19	23.87	0.03	0.54	0.37	0.09	0	0	0.76	0	0.24	Fast
7	18	0.27	0.34	36.25	15.91	0.58	0.11	0.31	0	0.89	0	0	0.11	Fast
8	29	1.16	0.28	36.71	42.67	0.83	0.15	0.02	0	0.90	0.02	0.01	0.07	Fast
9	20	0.45	0.6	92.55	74.06	0.58	0.06	0.35	0	0.71	0.12	0.17	0	Fast
10*	23	0.92	0.12	2.75	-8.04	0.91	0	0	0.09	-	-	-	-	FES
11	28	0.61	0.14	5.13	46.01	0	1	0	0	0.67	0.23	0.10	0	FES
12	29	1.13	-0.04	23.82	1.70	0.65	0	0	0.35	0	0.88	0	0.12	FES
13	20	0.94	0.05	13.02	31.34	0.81	0.10	0.02	0.07	1	0	0	0	FES
14	12	0.30	0.35	30.52	82.93	0.68	0.08	0.24	0	0.35	0.35	0.09	0.21	FES
15	26	1.16	0.28	34.25	20.17	0.92	0	0	0.08	0.53	0.42	0.05	0	FES
16	21	0.61	0.33	62.00	46.68	0.72	0.14	0.14	0	0.23	0.71	0.06	0	FES
17	31	0.79	0.16	16.77	57.70	0.70	0.11	0.05	0.13	0.83	0.11	0.06	0	FES

Subject #	LEFM	Walking Speed (m/s)	Changes in propulsive forces (N)		Relative Contribution to Paretic Propulsion				Relative Contribution to Non-Paretic Propulsion				Group	
			Paretic	Non- Paretic	TLA _p	M _{a,p}	Int _p	d _p	TLA _{np}	M _{a,np}	Int _{np}	d _{np}		
18	27	0.75	0.13	3.17	8.17	0	0	0	1.00	0	0	0	1	SS
19	14	0.75	-0.01	13.25	6.41	0.69	0.08	0.04	0.20	0	0.45	0	0.55	SS
20	23	0.69	0.15	20.08	13.71	0	1	0	0	0.56	0.38	0.06	0	SS
21	28	0.72	0.24	21.80	36.34	0.88	0	0	0.12	0.07	0.92	0.01	0	SS
22	25	1.13	0.22	23.53	26.77	0.21	0	0	0.79	0	1	0	0	SS
23	26	0.80	0.26	28.06	49.63	0.89	0.07	0.05	0	0.81	0	0	0.19	SS
24	20	0.42	0.03	32.84	8.51	0.73	0.06	0.22	0	0.73	0	0	0.27	SS
25	16	0.18	0.37	49.88	77.51	0.21	0	0.79	0	0.92	0.02	0.06	0	SS
26*	16	0.55	0.01	28.83	-22.45	1	0	0	0	-	-	-	-	SS
Mean	24	0.75	0.2	28.64	31.93	0.54	0.22	0.10	0.14	0.49	0.35	0.03	0.13	
SD	6	0.3	0.14	19.12	25.05	0.34	0.35	0.18	0.26	0.38	0.35	0.05	0.23	

LEFM = Lower Extremity Fugl-Meyer score, TLA = trailing limb angle, M_d = ankle moment, d = lever arm length, change = changes in walking speed from pre to post-training, Int = interaction term. Subscript "p" denotes the paretic limb and "np" denotes the non-paretic limb.

* Subject#2, 10, and 26 did not increase non-paretic propulsive force following training. Thus, the relative contributions of the variables for these 3 subjects were not included in the overall averages in Table 1.