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Targeted pelvic constraint force induces enhanced use of the paretic leg during walking in persons post-stroke

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Abstract

The purpose of this study was to determine whether activation of muscles in the paretic leg, particularly contributing to propulsion, and gait symmetry can be improved by applying a targeted resistance force to the pelvis in the backward direction during stance phase while walking in individuals post-stroke. Thirteen individuals post-stroke participated in two experimental sessions, which consisted of treadmill walking, with either targeted or constant resistances, together with overground walking. For the targeted condition, a resistance force was applied to the pelvis during the stance phase of the paretic leg. For the constant condition, the resistance force was applied throughout the whole gait cycle. Participants showed greater increase in medial hamstring muscle activity in the paretic leg and improved step length symmetry after the removal of targeted resistance force, compared to effects of a constant resistance force ($P < 0.03$). In addition, treadmill walking with the targeted resistance induced more symmetrical step length during overground walking 10 min after the treadmill walking, compared to the result of the constant resistance force ($P = 0.01$). Applying a targeted resistance force to the pelvis during the stance phase of the paretic leg may induce an enhanced use of the paretic leg and an improvement in gait symmetry in individuals post-stroke. These results provide evidence showing that applying a targeted resistance to the pelvis may induce a forced use of the paretic leg during walking.

Keywords

stroke; locomotion; forced use; resistance force; constraint-induced movement therapy

I. INTRODUCTION

LOCOMOTOR training using a treadmill is a commonly used approach in clinics for improving walking function in stroke survivors [1]. While some stroke survivors show improvements in walking speed [2], and in endurance after treadmill training [3], the functional gains reported have been relatively small. Robotic-assisted treadmill training has been tested in stroke survivors, but results indicate that it is not superior to physical

therapist-assisted treadmill training for improving walking function [4], particularly for these patients in the chronic phase (i.e., > 6 months) [5]. One possible reason for the reduced effectiveness of current manually or robotically assisted treadmill training may be that it does not effectively target the specific impairment of the paretic leg in hemispheric stroke survivors during training.

Recent studies indicate that the impairments in propulsion of the paretic leg are key factors that contribute to the gait deficits [6], [7], such as slow walking speed and step length asymmetry. Individuals post-stroke reserve a robust propulsion on the paretic leg and exploitation of this reserve may potentially enhance walking recovery [8]. Thus, targeting improvement in propulsion of the paretic leg might potentially be a promising approach towards improving gait symmetry and walking function in stroke survivors.

Constraint induced movement therapy (CIMT) is a promising paradigm designed to restore motor function of the affected arm of individuals post-stroke [9], [10]. Previous randomized controlled studies indicated that CIMT is more effective than conventional interventions in improving motor function of the affected arm in individuals post-stroke [9], [11]. It is still a challenge to effectively transfer this paradigm to lower limb training. Experimental assessments have shown that it can be quite challenging to specifically target enhanced propulsion of the paretic leg during treadmill training, because many stroke survivors employ a compensatory strategy consisting of relying more on the non-paretic leg to compensate for the weakness of the paretic leg during walking [12]. The reduced use of the paretic leg during walking might be partially due to a “learned non-use” [13], which is a problem that could potentially be reversed by a repetitive “forced use” protocol, implemented by the application of a mechanical constraint to the sound limb [14]. However, it has proved difficult to immobilize the non-paretic leg using a mechanical constraint and to induce a forced use of the paretic leg during walking, because both the paretic and non-paretic legs need to be involved during locomotion [15].

Results from previous studies indicated that the application of a resistance force to the leg during swing phase increases muscle activity of the leg flexor muscles and improves the step length symmetry in stroke survivors [16], [17]. However, in these studies, the resistance force was delivered to the paretic leg during the swing phase of gait, which may have no or limited impact on the propulsive capacity of the paretic leg. Results from a previous study in unimpaired controls indicated that muscle activity of the medial gastrocnemius decreased with the application of a constant forward assistance to the waist and increased with the application of a constant backward resistance to the pelvis during treadmill walking [18], suggesting the important role of medial gastrocnemius in generating forward propulsion. Thus, it is possible that the application of a resistance force to the pelvis during the stance phase of gait may enhance muscle activity of leg extensors. In particular, because stroke patients mostly rely on their non-paretic leg for producing force to move the body forward [19], the application of a constant backward resistance to the pelvis may actually reinforce this compensatory strategy, resulting in limited impact on the muscle activation of the paretic leg during stance. Thus, in order to increase muscle activation of the paretic leg during stance and reduce the compensatory movement from the non-paretic leg, the backward resistance force might need to be controlled for timing, i.e., only applied to the pelvis during

the stance phase of the paretic leg. In addition, during the early to mid-stance phase of the paretic leg, the non-paretic leg is mostly in the swing phase, applying backward resistance during this period of time may effectively target the use of the paretic leg, particularly leg extensors, which contribute to propulsion.

In light of these observations, the aim of this study was to determine whether introducing a targeted resistance force delivered during the stance phase of the paretic leg would induce enhanced muscle activity, particularly the hip and ankle extensors, of the paretic leg in hemispheric stroke survivors. Further, we propose to determine whether repeated exposure to the targeted resistance force during walking could induce a learning effect of promoting improved use of the paretic leg, particularly during the period after the resistance was removed, and whether this improvement could transfer to overground walking. We hypothesized that 1) the repeatedly applying the targeted resistance force would result in greater retention of enhanced muscle activity of extensors from the paretic leg, and improved symmetry of gait pattern after the removal of the resistance, compared with the results of constant resistance force; and 2) the improved symmetrical gait pattern obtained from treadmill walking with targeted resistance force would transfer to overground walking.

II. MATERIALS AND METHODS

A. Participants

Thirteen individuals (52.2 ± 13.0 years, 6 females) with post stroke hemiparesis participated in this study (Table 1). All participants were chronic stroke (>6 months) patients with the average time post injury was 8 ± 4.8 years. The inclusion criteria were the following: 1) age 21 to 75 years old; 2) single unilateral, supratentorial, ischemic or hemorrhagic stroke, confirmed with magnetic resonance imaging or computed tomography scan; 3) showing paresis/weakness of the paretic leg; and 4) ability to stand and walk (> 10 m) without assistance from another person (the use of assistive devices or orthoses was allowed but the orthoses need to be below knees).

Exclusion criteria were the following: 1) evidence of brainstem or cerebellar stroke; 2) a score of < 24 on the Mini Mental State Examination [20]; 3) other neurological conditions, such as cardiorespiratory/metabolic disorders and orthopedic conditions, which affect their ambulatory ability; 4) uncontrolled hypertension (systolic > 200 mm Hg, diastolic > 110 mm Hg); 5) botulinum toxin injection within the prior 6 months; and 6) inability to tolerate 30 min of treadmill walking. The Northwestern University Medical School Institute Review Board approved all procedures and all participants signed informed consent before data collection.

B. Apparatus

A custom designed cable-driven robotic system [21], which was mounted over a treadmill, was used to deliver a controlled backward resistance force at the pelvis while participants walked on a treadmill (Fig. 1A). The cable-driven system consisted of two servomotor systems (AKM 33H, Kollmorgen, Radford, VA) that were used to drive 2 aluminum cable spools, which were located at both sides of the treadmill. Two nylon-coated stainless-steel

cables (1.6 mm in diameter), which were driven by 2 cable spools, were attached to the participant's pelvis through a waist belt and a pulley system, through which the resistance force was delivered. A compression/tension load cell (MLP-25, Transducer Techniques, Temecula, CA), which was attached in series with the cables, was used to record the applied force. The cable-driven robotic system was controlled by a PC using a custom-written program in LabVIEW (National Instrument, Austin, TX, USA). In order to control the timing when the pelvic resistance force was delivered, the ankle position signals were measured using two custom designed 3-dimensional position sensors, which were attached at the participant's ankles through Velcro straps (Fig. 1A) [22]. The controller used the ankle position signals to trigger the resistance force at targeted phase of gait.

C. Experimental Protocol

All participants were tested under two conditions (i.e., a Targeted Resistance Force and a Constant Resistance Force) in two experimental sessions with a 1-week interval in between. The order of the two test sessions was randomized across participants (Fig. 1B). For the Targeted Force condition, a backward resistance force was applied to the pelvis only during the time when the paretic leg was in contact with the treadmill belt (i.e., starting from ~100ms after the initial contact of the paretic leg for 500ms, and no force was applied during other phases of the gait cycle). We selected this early to mid-stance phase for the targeted force condition because during this period of time the non-paretic leg was mostly in the swing phase and therefore the resistance force applied could force participants to use more on their paretic leg. The magnitude of the force was set at 8% of body weight, which was determined based on a previous study [23], and was adjusted based on the tolerance of each participant.

For the Constant Force condition, a backward resistance force was applied to the pelvis for the whole gait cycles. The treadmill speed was set at a self-selected comfortable speed for each participant, and this choice remained the same for all testing sessions. An overhead harness, which was attached to a body weight support system, was used for protection only; no body weight support was applied for all participants. For safety, participants were also allowed to hold on a sidebar using their non-paretic arm during walking.

Each session included the following procedures: participants walked on a treadmill without an applied force for 1 min (baseline), and then a resistance force (Targeted or Constant Force) was applied while participants walked for 15 min (adaptation). Afterwards, the resistance force was removed, and participants continued walking on the treadmill for additional 2 min (post-adaptation). The motor skills obtained during the adaptation period may be partially washed out during the post-adaptation period. Thus, in order to test for transfer of the motor skills from treadmill to overground walking, participants walked on the treadmill with the applied resistance force for another 15 min (re-adaptation) after they took a sitting break for 5 minutes.

In addition, participants walked overground on the GAITRite® mat (CIR Systems, Inc., Franklin, NJ, USA) before treadmill walking, immediately, 10 min, and 20 min after treadmill walking (Fig. 1C). Three trials were tested for each condition. Participants were transported in a wheelchair between the treadmill and overground walking to reduce the

potential washout of motor skills during the transition period. Before data collection, participants also walked on the treadmill at their maximum speed for 30 strides. EMG activity during this period was used to normalize the integrated EMGs.

D. Data Collection

Muscle activities of tibialis anterior (TA), medial gastrocnemius (MG), rectus femoris (RF), and medial hamstring (MH) from both legs were recorded using surface EMG electrodes (Bagnoli™, Delsys, Boston, MA, USA). Specifically, we measured MG and MH muscle activities as contributors to propulsion of the leg during the stance phase, and measured TA and RF muscle activities to see whether any co-contractions of agonist (i.e., MG and MH) and antagonist (i.e., TA and RF) occurred during the stance phase of walking while the resistance force was applied. EMG signals were amplified ($\times 1000$) and band-pass filtered (20–450 Hz) using Bagnoli-16 Amplifier (Delsys, MA, USA), and then sampled with an A/D board (National Instruments, Austin, TX, USA) at 500 Hz using a custom-written program in LabVIEW. Ankle position data were recorded using two customized 3-dimensional position sensors using the same LabVIEW program at 500 Hz (Fig. 1A). Gait speed and other spatiotemporal gait parameters were calculated using the software from the GAITRite®.

E. Data Analysis

All EMG and kinematic data were analyzed using custom-written programs in MATLAB. Specifically, ankle position data were low-pass filtered using 4th-order Butterworth filter with the cutoff frequency at 10 Hz. Step length was defined as the anteroposterior distance between the two legs' ankle position at initial contact of each leg [24]. The asymmetry index of step length was quantified as following [25], [26]:

$$\text{Asymmetry index} = \left| 1 - \left(\frac{\text{paretic}}{\text{non-paretic}} \right) \right| \times 100 \quad (1)$$

EMG data were high-pass filtered using 4th-order Butterworth filter with the cutoff frequency at 10 Hz, and notch filtered (with the frequency ranged from 58–62 Hz and 118–122 Hz), rectified, and smoothed (with the frequency at 20 Hz). The EMG data were segmented into different gait cycles based on ankle position data. The integrals of the muscle activity during stance phase of the gait were calculated for both legs. The integrated EMG was further normalized to the integral of the muscle activity of each muscle while subjects walked at their maximum walking speed. Asymmetry of step length and EMG integrals during the following 8 subintervals were calculated and compared across two testing conditions and different time periods. Specifically, the asymmetry of step length and integrals of EMG during the following intervals were calculated: the last 20 steps before loading (baseline; B); the first 5 steps (R1), 5 steps of the first quarter (R2), the middle 5 steps (R3), 5 steps of the third quarter (R4), and the last 5 steps (R5) during the adaptation period; the first (P1) and the last 5 steps (P2) during the post-adaptation period.

In order to compare the change in integrals of EMG and asymmetry of step length between the two testing conditions, the integrals of EMG and the asymmetry of step length during the

adaptation and post-adaptation periods were subtracted from the mean of the integrated EMG and mean of asymmetry of step length during baseline.

Step length asymmetry and gait speed during overground walking were automatically calculated as described previously at the following periods of time: before treadmill walking (Baseline); immediately after treadmill walking (T1); 10 min after treadmill walking (T2); 20 min after treadmill walking (T3).

F. Statistical Analysis

Paired *t* tests were conducted to compare the muscle activity and asymmetry of step length between the two testing conditions for each time of periods separately (i.e., from B to P2; Fig. 3A, D, & G), and to compare changes in these variables between two conditions (Fig. 3C, F, & I). Repeated-measures ANOVAs were used to compare the integrated EMG and asymmetry of step length at different time of periods (3 timepoints; B, R1, & P2) during treadmill walking within each condition (Fig. 3B, E, & H). For these within-condition comparisons, we chose the timepoint R1 (early adaptation) to examine the immediate effect of the resistance force on the EMG and asymmetry of step length, and the timepoint P2 (late post-adaptation) to examine the retention of motor adaptation induced by the resistance force, compared to the baseline. In particular, we focused on the retention of motor adaptation during the late post-adaptation because it has more potential clinical significance. Post-hoc Tukey-Kramer tests were used to identify specific differences (3 comparisons were conducted) with Bonferroni corrections when appropriate. For overground walking, repeated-measures ANOVAs were used to compare the asymmetry of step length and walking velocity at different time periods (4 time points; Baseline, T1, T2, & T3) for each testing condition (Fig. 4). Post-hoc tests were also used to identify specific differences (6 comparisons were conducted) with Bonferroni correction. In addition, a 2-way ANOVA, with factors of condition (Targeted vs. Constant forces) and time (4 assessment points), were used to test interactions between condition and time on the asymmetry of step length and walking velocity. The IBM SPSS Statistics 20.0 statistical package (IBM Corp., Armonk, NY, USA) were used for analysis. The alpha level for all statistical tests was 0.05.

III. RESULTS

Application of the Targeted Resistance Force during the stance phase of walking induced enhanced muscle activity in MH of the paretic leg. For instance, as shown from one typical participant (Fig. 2A), integrated EMG of MH from the paretic leg appeared to increase with the application of the targeted force during the adaptation period. In addition, enhanced EMG of MH was seemingly retained following the removal of the force during the post-adaptation period for this participant. The integrated EMG of MH from the non-paretic leg showed modest changes during the adaptation and post-adaptation periods, Fig. 2B. In addition, the application of the targeted force during walking seemed to induce an improvement in step length symmetry (i.e., decreased step length asymmetry), which was seemingly retained following the removal of the force during the post-adaptation period, see Fig. 2C.

In contrast, the application of Constant Resistance Force appeared to induce slight enhanced muscle activity of MH from the paretic leg during the early adaption period for the first several steps, then, quickly returned to a level that was comparable to baseline during the middle to late adaptation period, and the post-adaptation period Fig. 2A. The integrated muscle activity of MH from the non-paretic leg appeared to show modest changes during the adaptation period, but seemingly increased during the late post-adaptation period following the removal of the force, see Fig. 2B. In addition, the application of the constant force induced slightly increased and varied step length asymmetry during the adaptation period, but returned to a level that was comparable to baseline during the post-adaptation period.

Group averages of integrated EMG of MH are shown in Fig. 3. The application of Targeted Force had a significant impact on the muscle activity of MH from the paretic leg ($F = 9.02$, $P < 0.01$, ANOVA; Fig. 3B) but had no significant impact on the non-paretic leg ($F = 2.08$, $P = 0.15$; Fig. 3E). Post-hoc analysis indicated that muscle activity of MH from the *paretic leg* during late post-adaptation (P2: 118.81 ± 14.83) was significantly greater than that during baseline (B: 98.26 ± 14.22 ; $P < 0.01$), Fig. 3B.

In contrast, the application of the Constant Force had significant impact on the muscle activity of the MH in the *non-paretic* leg ($F = 6.58$, $P < 0.01$; Fig. 3E). Post-hoc analysis indicated that the integrated MH muscle activity of the non-paretic leg during the early adaptation (R1: 102.80 ± 20.62 ; $P < 0.01$) and late post-adaptation (P2: 96.05 ± 25.25 , $P = 0.02$) periods was significantly greater than that during baseline (B: 81.88 ± 16.06). In addition, the application of the constant force also had significant impact on the muscle activity of MH in the paretic leg ($F = 4.35$, $P = 0.03$; Fig. 3B). However, post-hoc analysis indicated that the integrated MH muscle activity of the paretic leg showed no significant difference between the different periods ($P > 0.15$).

Application of the Targeted Force (B: 98.26 ± 14.22 ; R1: 109.80 ± 11.62 ; R2: 112.63 ± 14.07 ; R3: 116.64 ± 14.83 ; R4: 112.70 ± 15.07 ; R5: 114.22 ± 14.59 ; P1: 118.02 ± 15.37 ; P2: 118.81 ± 14.93) induced greater muscle activity of MH in the paretic leg only during the late post-adaption period (P2), compared with the constant force condition (B: 92.93 ± 7.03 ; R1: 108.01 ± 7.45 ; R2: 101.77 ± 7.23 ; R3: 99.54 ± 8.67 ; R4: 95.33 ± 9.93 ; R5: 94.14 ± 9.48 ; P1: 104.19 ± 11.94 ; P2: 86.56 ± 9.82) ($P = 0.02$; Fig. 3A). In addition, application of the Targeted Force (B: 87.22 ± 11.24 ; R1: 95.33 ± 11.06 ; R2: 95.13 ± 11.72 ; R3: 90.83 ± 10.69 ; R4: 88.37 ± 11.07 ; R5: 90.69 ± 11.60 ; P1: 88.34 ± 11.09 ; P2: 83.18 ± 10.79) induced comparable muscle activity of MH in the non-paretic leg during all subintervals of treadmill walking, compared to the constant force condition (B: 81.88 ± 4.63 ; R1: 102.80 ± 5.95 ; R2: 94.91 ± 7.04 ; R3: 89.26 ± 5.24 ; R4: 87.07 ± 6.30 ; R5: 89.87 ± 7.73 ; P1: 88.94 ± 4.86 ; P2: 96.05 ± 7.29) ($P > 0.40$; Fig. 3D).

The changes in MH muscle activity of the paretic leg from baseline to late post-adaptation were significantly greater for the Targeted Force than that for the Constant Force (i.e., 20.54 ± 11.20 vs. -6.37 ± 17.49 ; $t = -6.01$, $P < 0.01$; Fig. 3C), although the changes in MH muscle activity of the *paretic leg* from baseline to early adaptation were not significantly different between two conditions ($P > 0.50$; Fig. 3C). In addition, the changes in MH muscle activity of the *non-paretic* leg from baseline to late post-adaptation were significantly greater for the

constant force condition than that for the targeted force condition (14.17 ± 14.92 vs. -4.04 ± 10.1 ; $t = 5.00$, $P < 0.01$; Fig. 3F), although the changes in MH muscle activity of the non-paretic leg from baseline to early adaptation were not significantly different between two conditions (i.e., 8.10 ± 24.75 vs. 20.92 ± 18.52 ; $t = 1.57$, $P = 0.15$; Fig. 3F). The application of the Targeted and Constant Forces induced no significant changes in integrated EMGs of TA, MG and RF from both legs ($p > 0.08$).

Application of the Targeted Resistance Force had a significant impact on the step length asymmetry during treadmill walking ($F = 10.313$, $P < 0.01$; Fig. 3H). Further, post-hoc analysis indicated that step length asymmetry during late post-adaptation was significantly less than at baseline (B: $30.20 \pm 24.08\%$; P2: $13.19 \pm 10.30\%$; $P = 0.03$) and that during early adaptation (R1: $35.38 \pm 22.04\%$; $P = 0.01$). The application of the Constant Force had no significant impact on the step length asymmetry ($F = 2.61$, $P = 0.10$; Fig. 3H).

Application of the Targeted Force (B: 30.20 ± 24.08 ; R1: 35.38 ± 22.04 ; R2: 25.59 ± 22.43 ; R3: 19.73 ± 12.16 ; R4: 19.35 ± 10.27 ; R5: 21.75 ± 11.85 ; P1: 23.89 ± 23.50 ; P2: 13.19 ± 10.30) induced less step length asymmetry during the mid and late adaptation periods (R3–5) and the late post-adaptation period (P2), compared to the constant force condition (B: 33.66 ± 26.87 ; R1: 36.66 ± 27.08 ; R2: 38.55 ± 25.70 ; R3: 35.53 ± 20.46 ; R4: 41.68 ± 20.89 ; R5: 38.70 ± 21.78 ; P1: 33.47 ± 10.51 ; P2: 41.04 ± 26.60) ($P < 0.03$; Fig. 3G). The changes in step length asymmetry from baseline to late post-adaptation were significantly greater for the targeted force condition than that for the constant force condition (i.e., decreased; $-17.01 \pm 19.26\%$ vs. $7.38 \pm 14.11\%$; $t = 3.31$, $P < 0.01$; Fig. 3I).

For the step length asymmetry during overground walking (Fig. 4A), the main effect of condition ($F = 2.06$; $P = 0.13$) and time ($F = 1.82$; $P = 0.21$) was not significant. However, the condition $\times$ time interaction was significant ($F = 3.25$, $P = 0.03$). Specifically, the targeted resistance training exhibited less step length asymmetry at the time of 10 min after the training (T2), compared with the Constant Resistance training (18.22 ± 14.20 vs. 28.30 ± 20.61 ; $t = -3.01$, $P = 0.01$). In addition, for the Targeted Force condition, the main effect of time was not significant ($F = 2.21$, $P = 0.11$). The step length asymmetry at the time of 10 min after the training (T2) tended to decrease from baseline, although this was not significant ($P = 0.13$). For the constant force condition, the main effect of time was significant ($F = 3.50$, $P = 0.03$). The step length asymmetries at the time of immediately (T1) and 10 min after treadmill training (T2) tended to increase from baseline, but these were not significant ($P > 0.14$).

For the overground walking velocity (Fig. 4B), there was significant main effect of condition ($F = 10.40$, $P < 0.01$) and time ($F = 7.98$, $P < 0.01$). The condition $\times$ time interaction was not significant ($F = 1.24$, $P = 0.09$). The main effect of time was significant for both the Targeted ($F = 5.75$, $P < 0.01$) and Constant Force conditions ($F = 4.24$, $P = 0.01$). The gait velocities at the time of 10 min and 20 min after the end of treadmill training tended to increase from baseline for both conditions, but these were not significant (i.e., for the Targeted Force condition, $P > 0.09$, and for the Constant Force condition, $P > 0.13$).

IV. DISCUSSION

We aimed to determine whether the application of a targeted backward resistance force to the pelvis during the stance phase of the paretic leg during walking would result in an enhanced use of the paretic leg, particularly muscles contributing to propulsion, and thus improve gait symmetry in hemispheric stroke survivors. We found that participants showed greater muscle activity of the MH in the paretic leg and improvement in symmetry of step length during late post-adaptation period after the removal of the Targeted Resistance Force, compared to the effect of the Constant Resistance Force. Furthermore, participants showed more symmetrical step length during overground walking 10-min after treadmill training with Targeted Resistance, compared to the effect of the Constant Resistance.

The application of a Targeted Resistance Force may promote enhanced use of the paretic leg of such individual stroke survivors. Many individuals with stroke often use a compensatory strategy during walking by relying more on their non-paretic leg and generating less force from their paretic leg. Results from this study indicated that the application of a Targeted Resistance Force during the stance phase of the paretic leg could increase muscle activity (e.g., MH) of the paretic leg during treadmill walking. Specifically, for the Targeted Force condition, the force was applied to the pelvis during the early to mid-stance phase of the paretic leg, and during this period of time, the non-paretic leg was mostly in the swing phase. Thus, individual stroke survivors could not rely on the non-paretic leg at this time point but were required to rely on the paretic leg for generating additional torque to counteract this resistance force and move the pelvis and upper body forward.

Repetitive exposure to the Targeted resistance force may result in an increased muscle activity of the paretic leg even when the resistance force was removed during the post-adaptation period. For instance, we observed increased muscle activity in MH of the paretic leg during the post-adaptation period when the force was removed for the Targeted Force condition. The retention of the improved muscle activity could be achieved through use-dependent motor learning, referring to behavioral or neural changes induced by movement repetition [27], [28]. Specifically, the repetition of a certain movement or muscle activity induced by an external perturbation may influence a movement or muscle activity pattern when that perturbation is eliminated [27]. This is in line with a previous finding demonstrating that a movement curvature induced by repetitive hand-reaching movements with an obstacle persists even when the obstacle is eliminated [29]. There is also evidence that increased muscle activity in stroke survivors induced by repetitive exposure to a resistance force is retained after the removal of the force [30]. Interestingly, the targeted force application was less effective to enhance the muscle activation of MG, known to generate propulsive force during the late stance phase [31], likely because the targeted timing (i.e., early to mid-stance phase) of the force application might be too early to induce significant increase in EMG of MG.

The enhanced use of the paretic leg induced by the application of the Targeted Resistance Force might also result in a more symmetrical gait pattern during walking. In this study, participants showed different asymmetrical gait patterns during baseline, with 7 participants walking with a shorter step length on the paretic leg and 5 participants walking with a

shorter step length on the non-paretic leg for the Targeted Force condition. Thus, different strategies might have been used for improving step length symmetry. For instance, for those 7 participants who showed a shorter step length on the paretic leg at baseline, the enhanced muscle activation of the hip extensors (MH) might facilitate the paretic leg to generate additional propulsion force, which might result in an improved step length of the paretic leg during the post-adaptation period, as shown in Table 2.

On the other hand, for those 5 participants who showed a shorter step length on the non-paretic leg during baseline, the enhanced muscle activation of the hip extensors might facilitate forward movement of the pelvis, which might facilitate leg swing of the non-paretic leg, resulting in an increased step length of the non-paretic leg. As a consequence, these stroke survivors showed a more symmetrical gait pattern with the application of the targeted resistance force, although different strategies might have been used, depending on their gait pattern during baseline.

The application of a Constant Resistance Force might reinforce the compensatory strategy by forcing the subject to rely more on the non-paretic leg during walking. Specifically, for the Constant Force condition, the resistance force was applied during the whole gait cycle, including the stance phase of both the paretic and the non-paretic legs. Thus, participants might need to generate additional force from both legs to counteract the resistance force. Furthermore, because many stroke survivors have weaker muscles in the paretic leg [32], they might feel more challenged to generate enough additional force to counteract the resistance force during the stance phase of the paretic leg as compared with the non-paretic leg. In other words, the effort level required might be greater during the stance phase of the paretic leg than for the non-paretic leg. Over time, stroke survivors might adjust the force production level from each leg to induce a comparable effort level of both legs. Too high an effort level might induce fatigue of leg muscles [33]. This motor strategy was also reported in previous studies in stroke survivors during walking or other dynamic lower limb movement [34], [35]. Thus, stroke survivors might further increase force output from the non-paretic leg, i.e., rely more on the non-paretic leg, to counteract the resistance force. Over time, this compensatory motor strategy may be partially retained, even when the resistance force was removed during the late post-adaptation period, which might be achieved through a use-dependent motor learning mechanism [28]. This possibility was supported by the observation of greater integrated EMG of MH from the non-paretic leg, compared to that from the paretic leg during the late adaptation period, Fig. 3E, F. This declining asymmetrical force production pattern also induced a more asymmetrical gait pattern, Fig. 3I.

The reduction in step length asymmetry after the Targeted Resistance training transferred to overground walking. Specifically, participants exhibited smaller step length asymmetry while walking overground at 10-minute after treadmill walking with the targeted resistance, compared to that after the constant resistance training (T2; Fig. 4A). This finding parallels with previous findings demonstrating transfer of the motor adaptation from treadmill to overground walking in stroke survivors [17], [36], [37]. Here, individuals with stroke-related hemiparesis exhibited transfer of symmetrical step length from treadmill to overground walking [17], [37]. One possible mechanism is that some neural circuits controlling

locomotion during treadmill and overground walking environmental contexts may be partially shared, and the temporal sequence of muscle activations and the limb movements are broadly similar between two contexts [30]. Therefore motor skills obtained from one context, i.e., treadmill, may potentially transfer to other context, i.e., overground walking [38], [39]. However, there were some differences in environmental contexts between treadmill and overground walking [37]. For example, when stroke individuals walked on a treadmill, they did not actually move through space as much as when they walked overground. In addition, participants were holding onto the front handrail and were wearing a safety harness during treadmill walking. These external balance support might improve their gait performance [40]. These differences in the environmental context could have an impact on peripheral inputs and perceptual interactions [41], [42], resulting in only partial transfer of motor skills from treadmill to overground walking (T2; Fig. 4A). In particular, we did not observe the transfer effect immediate (T1) after treadmill walking, which might due to the fatigue of the paretic leg muscles after ~30 minutes of treadmill walking with the resistance force, and 20 minutes after treadmill walking, which might due to the washout of motor skills that occurred during the previous two walking sessions (i.e., T1 and T2).

Results obtained from this study may have potential applications for gait rehabilitation in chronic-stroke patients. For instance, in clinical settings, locomotor training has been widely used to improve walking function in patients with stroke, but patients may rely more on their non-paretic leg for locomotion using compensatory strategies, resulting in limited improvements in motor function of the paretic leg. Our results suggest that applying a targeted resistance force to the pelvis during the early to mid-stance phase of the paretic leg during locomotor training may induce a forced use of the paretic hip extensor and improve spatial gait symmetry. Therefore, physical therapists may consider the application of a targeted resistance to the pelvis at early to mid-stance using an elastic therapy belt [43] to enhance a forced use of the paretic leg, particularly hip extensors contributing to propulsive force, in stroke survivors.

In this paper, we focused entirely on short-term changes in behavior and muscle activity. Thus, these results cannot be generalized to longer term training effects. Future studies should examine the ability to retain or transfer performance following long period of time training. In addition, all participants held onto the frontal bar using their non-paretic arm to maintain balance during treadmill walking when the resistance force was applied, but holding onto the frontal bar might have dampened the training effect because they might also use their non-paretic arm as a compensatory strategy to counteract the resistance force. Future studies should measure the force applied by non-paretic arm and determine to what extent stroke patients were using their non-paretic arm to compensate the resistance force applied to the pelvis. The duration of force was set as 500 ms across all participants, we do not know whether personalization of this parameter will affect the outcomes. We also did not record soleus muscle activity, which contributes to propulsion during walking. In addition, the ground reaction force was not recorded during treadmill walking in this study. We have no direct evidence regarding whether participants increased their propulsion force of the paretic/non-paretic leg when the resistance force was applied. In this study, 11 of 13 participants wore an AFO on their paretic leg (Table 1), their ankle dorsiflexion and

plantarflexion movements might be highly restricted, which might reduce muscle activities of TA and MG muscles [44].

V. CONCLUSION

Applying a targeted backward resistance force to the pelvis during the stance phase of the paretic leg may induce forced use of the paretic leg and improve gait symmetry in hemispheric stroke survivors. Further, improved gait symmetry may transfer from treadmill to overground walking. Knowledge obtained from this study such as the timing that a resistance force can be applied provides insights for developing locomotor training protocols to improve motor function of the paretic leg in stroke survivors.

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Fig. 1.

Experimental setup and protocol. (A) Participants walked on a treadmill with a backward resistance force applied using a cable-driven robotic system. Two cables, which were driven by two motors, were used to apply a backward resistance force. (B) Participants were randomly assigned into a constant–targeted or targeted–constant group. Each participant visited the lab for two different days. For the constant–targeted group, participants were under the constant resistance training condition first (Day 1) and at least a week later they were under the targeted resistance condition (Day 2). For the targeted–constant group, the order of the training condition was reversed. (C) The treadmill walking session consisted of 1-min walking without a resistance force (baseline), 15-min treadmill walking with resistance force (adaptation), and 2-min walking without resistance (post-adaptation). Then, after 5-min seated break on a treadmill, participants performed another 15-min treadmill walking with resistance. Also, participants conducted overground walking before treadmill walking, immediately after treadmill walking, 10 min after treadmill walking, and 20 min after treadmill walking. (D) A backward resistance force was applied to the pelvis either only during stance phase of the paretic leg (bottom; black line; targeted resistance) or during the whole gait cycle (bottom; grey line; constant resistance).

Fig. 2.

Integrated EMG of medial hamstring from the paretic leg (A) and the non-paretic leg (B) and stride-by-stride of step length asymmetry (C) from one representative participant for the targeted and constant resistance force conditions.

Fig. 3.

Group average of integrated medial hamstring (MH) muscle activity and step length asymmetry for the targeted and constant resistance force conditions. (A) Group average of integrated EMG of MH from the paretic leg during the course of treadmill walking for the targeted and constant resistance force conditions. (B) Group average of integrated EMG of the paretic leg during baseline, early adaptation period, and late post-adaptation period for both force conditions. (C) Group average of changes in integrated EMG of MH from the paretic leg during the early adaptation period (i.e., from baseline to early adaptation period), and during the late post-adaptation period (i.e., from baseline to late post-adaptation period) for both force conditions. (D) Group average of integrated EMG of MH from the non-paretic leg during the course of treadmill walking for both force conditions. (E) Group average of integrated EMG from the non-paretic leg during baseline, early adaptation period, and late post-adaptation period for both force conditions. (F) Group average of changes in integrated EMG of MH from the non-paretic leg during the early adaptation period and during the late post-adaptation period for both force conditions. (G) Average of step length asymmetry during the course of treadmill walking for the targeted and constant resistance force conditions. (H) Group average of step length asymmetry during baseline, early adaptation, and late post-adaptation periods for the targeted and constant resistance force conditions. (I) Group average of change in step length asymmetry during the early adaptation period (i.e., from baseline to early adaptation period), and during the late post-adaptation period (i.e., from baseline to late post-adaptation period) for both force conditions. Data shown above

are means $\pm$ standard errors. Abbreviations: B, baseline; R1–5, adaptation periods; P1–2, post-adaptation periods (see detail in method section). * indicates significant difference.

Fig. 4.

Average of step length asymmetry (A) and overground gait velocity (B) before treadmill walking, i.e., baseline, immediately after treadmill walking, 10 min after treadmill walking, and 20 min after treadmill walking for the targeted resistance and constant resistance force conditions. Abbreviations: T1, immediately after treadmill walking; T2, 10 min after treadmill walking; T3, 20 min after treadmill walking. * indicates significant difference. Data shown above are means and standard errors across subjects.

TABLE I

DEMOGRAPHIC INFORMATION FOR THE PARTICIPANTS

Participant	Gender	Age (y)	Weight (kg)	Post-injury (y)	Paretic side	Assistive device	Self-selected comfortable speed (m/s)	Resistance force (N)
P1	M	59	74	15	L	AFO	0.76	58
P2	F	29	71	5	L	AFO	0.80	42
P3	F	59	61	10	L	SPC/AFO	0.35	36
P4	F	22	137	2	L	LBQC/AFO	0.38	54
P5	M	49	102	9	L	AFO	0.41	40
P6	M	61	100	16	R	SPC	0.41	49
P7	M	53	116	6	L	SPC/AFO	0.66	68
P8	F	64	60	5	R	SPC/AFO	0.40	47
P9	M	58	67	12	R	AFO	0.62	52
P10	F	57	70	13	R	SBQC/AFO	0.30	41
P11	F	63	68	6	L	SPC/AFO	0.50	47
P12	M	47	91	3	R	None	0.80	71
P13	M	58	78	2	R	AFO	0.63	42

Abbreviations: y, years; M, male; F, female; L, left; R, right; AFO, ankle foot orthosis; SPC, single point cane; LBQC, large base quad cane; SBQC, small base quad cane.

TABLE II
STEP LENGTH AND MUSCLE ACTIVITY (TA, MG, RF) OF THE TARGETED AND CONSTANT RESISTANCE CONDITIONS

		B	R1	P2	from B to R1	from B to P2
Step Length (Paretic > Non-paretic; 5)						
Targeted	Paretic (cm)	44.9 (5.1)	46.8 (5.9)	41.4 (6.2)	1.9 (1.0)	-3.5 (2.0)
	Non-paretic (cm)	35.2 (7.0)	33.4 (6.0)	38.0 (6.4)	-1.8 (1.1)	2.7 (1.9)
	Asymmetry (%)	38.0 (18.3)	49.1 (11.1)	12.4 (4.9)	11.1 (8.0)	-25.7 (13.9)
Constant	Paretic (cm)	46.3 (4.8)	46.8 (5.1)	47.4 (4.8)	0.5 (0.7)	1.1 (2.8)
	Non-paretic (cm)	33.5 (7.3)	32.5 (6.7)	33.2 (6.6)	-1.0 (1.1)	-0.3 (1.5)
	Asymmetry (%)	52.4 (21.5)	58.8 (20.0)	55.3 (21.8)	6.4 (5.3)	2.9 (12.5)
Step Length (Paretic < Non-paretic; 7)						
Targeted	Paretic (cm)	32.0 (2.8)	33.2 (3.9)	36.2 (2.6)	1.2 (1.9)	4.2 (3.5)
	Non-paretic (cm)	44.0 (4.2)	39.4 (5.1)	39.8 (4.1)	-4.6 (2.0)	-4.2 (3.2)
	Asymmetry (%)	10.3 (6.8)	11.6 (7.6)	5.1 (4.6)	1.4 (3.7)	-5.1 (5.5)
Constant	Paretic (cm)	36.0 (2.5)	35.7 (2.5)	33.6 (2.7)	-0.3 (0.7)	-2.4 (1.1)
	Non-paretic (cm)	47.0 (3.3)	46.4 (3.1)	50.5 (3.4)	-0.5 (1.9)	3.5 (0.7)
	Asymmetry (%)	23.7 (2.4)	24.5 (3.3)	32.9 (4.3)	0.8 (1.5)	9.2 (2.1)
Muscle activity						
Targeted	TA (%)	102.4 (8.6)	118.5 (9.5)	102.1 (10.2)	16.1 (12.7)	-0.3 (3.9)
	Non-paretic	102.1 (10.2)	103.1 (8.7)	92.8 (12.3)	1.0 (5.3)	-9.3 (6.9)
	MG (%)	90.5 (10.0)	92.5 (9.1)	90.5 (10.9)	2.0 (6.0)	0.0 (3.0)
Constant	TA (%)	96.3 (8.3)	103.3 (7.8)	96.0 (9.6)	7.1 (2.7)	-0.3 (2.9)
	Non-paretic	85.4 (10.7)	93.2 (11.5)	83.8 (13.2)	7.8 (3.5)	-0.5 (5.3)
	MG (%)	92.0 (11.1)	89.8 (10.2)	89.8 (12.8)	-2.1 (2.3)	-2.2 (3.9)
Targeted	RF (%)	96.0 (6.3)	118.5 (12.7)	96.2 (10.3)	22.5 (11.8)	0.2 (6.5)
	Non-paretic	96.2 (9.8)	95.2 (8.3)	83.9 (13.1)	-1.0 (7.3)	-12.2 (6.9)
	MG (%)	97.0 (10.8)	82.6 (5.8)	99.6 (10.1)	-14.3 (7.6)	2.7 (7.1)
Constant	RF (%)	91.2 (6.3)	92.7 (4.6)	85.3 (10.45)	1.6 (5.4)	-5.9 (7.0)
	Non-paretic	92.5 (8.5)	87.9 (4.3)	99.3 (15.3)	-4.6 (7.0)	6.8 (8.5)
	MG (%)	101.9 (6.3)	98.4 (7.1)	103.4 (9.0)	-3.5 (7.4)	1.5 (4.4)

Values in parenthesis refer to standard error. Abbreviations: B, baseline; R1, early adaptation; P2, late post-adaptation.