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Accepted version. *Human Movement Science*, Vol 42 (August 2015): 212-224. DOI. © 2015 Elsevier. Used with permission.

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Using Swing Resistance and Assistance to Improve Gait Symmetry in Individuals Post-Stroke

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Abstract: A major characteristic of hemiplegic gait observed in individuals post-stroke is spatial and temporal asymmetry, which may increase energy expenditure and the risk of falls. The purpose of this study was to examine the effects of swing resistance/assistance applied to the affected leg on gait symmetry in individuals post-stroke. We recruited 10 subjects with chronic stroke who demonstrated a shorter step length with their affected leg in comparison to the non-affected leg during walking. They participated in two test sessions for swing resistance and swing assistance, respectively. During the adaptation period, subjects counteracted the step length deviation caused by the applied swing resistance force, resulting in an aftereffect consisting of improved step length symmetry during the post-adaptation period. In contrast, subjects did not counteract step length deviation caused by swing assistance during adaptation period and produced no aftereffect during the post-adaptation period. Locomotor training with swing resistance applied to the affected leg may improve step length symmetry through error-based learning. Swing assistance reduces errors in step length during stepping; however, it is unclear whether this approach would improve step length symmetry. Results from this study may be used to develop training paradigms for improving gait symmetry of stroke survivors.

Keywords: Adaptation, Resistance, Assistance, Stroke, Gait, Symmetry

1. Introduction

A major characteristic of hemiplegic gait observed in individuals post-stroke is spatial and temporal asymmetry (Patterson et al., 2008). An asymmetrical gait pattern may increase energy expenditure (Zamparo, Francescato, Luca, Lovati, & Prampera, 1995), increase the risk of falls (Di Fabio, Kurszewski, Jorgenson, & Kunz, 2004), and cause loss of bone density of the affected leg (Jørgensen, Crabtree, Reeve, & Jacobsen, 2000). These deficits can restrict functional

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mobility in individuals post-stroke and have a negative impact on the quality of life. Thus, restoration of gait symmetry is an important goal in stroke rehabilitation.

There is a growing interest in robotic-assisted gait training paradigms. Robots can generate external forces to alter abnormal gait patterns in individuals post-stroke, which is one benefit to this training approach. Previous findings have shown how individuals adapt to a force perturbation applied to the leg (Emken & Reinkensmeyer, 2005; Noble & Prentice, 2006; Savin, Tseng, & Morton, 2010). Initially, the perturbation deviates the leg trajectory away from the baseline, which creates movement errors. A few steps later, the leg trajectory goes back to the baseline as individuals learn to take into account the perturbation and correct the errors. When the force perturbation is removed, the leg trajectory skews to the opposite direction of the force perturbation (i.e., aftereffect). Studies have used this adaptation process to increase step length in humans with incomplete spinal cord injury (Houldin, Luttin, & Lam, 2011; Yen, Schmit, Landry, Roth, & Wu, 2012; Yen, Landry, & Wu, 2013) and individuals post-stroke (Savin, Tseng, Whitall, & Morton, 2013). Specifically, a resistance force was applied to a subject's leg to hinder leg swing during treadmill walking. When the force was removed, subjects demonstrated an aftereffect consisting of increased step length. However, for individuals post-stroke with a longer step length on the affected side at baseline, applying swing resistance force to the affected leg actually induced an increase in asymmetry after load release (Savin, Tseng, Whitall, & Morton, 2013). Thus, we postulated that for individuals post-stroke with a *shorter* step length on the affected side (in comparison to the non-affected side) at baseline, applying swing resistance to the affected leg during motor adaptation may improve step length symmetry after load release.

In clinics, applying swing assistance to the affected leg, particularly for individuals with a shorter step length on the affected leg, as needed has been used to increase step length of the affected leg during locomotor training. In traditional robotic gait training, the function of the robot is to provide assistive force (as opposed to resistive force) to move the leg into a predetermined "normal" trajectory. However, because the direction of the aftereffect is in the opposite direction of the force perturbation (i.e. individuals may

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demonstrate a shorter step length after swing resistance is removed), the step length in individuals post-stroke could actually deviate further away from normal after removing the assistance.

This phenomenon has been shown in a previous adaptation study with individuals post-stroke using a split-belt treadmill paradigm (Reisman et al. 2007). During the adaptation period, the two belts of the treadmill were moving at different speeds to induce a more symmetrical or asymmetrical gait pattern in individuals post-stroke. However, some subjects tended to adapt back to the baseline asymmetry later in the adaptation period, and show a deterioration in step length symmetry once the two belts of the treadmill were adjusted to move at the same speed during the post-adaptation period. Thus, we postulated that while the application of swing assistance to the affected leg for individuals post-stroke (with a shorter step length on the affected side at baseline), may improve step length symmetry during the adaptation period, it may actually have the opposite affect after the assistance load is released, leading to a greater deterioration in step length symmetry.

The purpose of this study was to examine how individuals poststroke with a shorter step length on the affected leg adapt to robotic assistance and resistance forces applied to the affected leg during treadmill walking. We were particularly interested in this subgroup of individuals post-stroke, i.e., individuals with a shorter step length on the affected leg, because these individuals post-stroke generally have lower walking function (based on our unpublished clinical results), and may actually need more gait training to improve their walking function compared with higher functioning individuals post-stroke. We hypothesized that following the removal of a resistance force, these individuals post-stroke would produce an aftereffect consisting of an increase in step length of the affected leg, resulting in an improvement in step length symmetry during the post-adaptation period. In contrast, we hypothesized that following removal of an assistance force, individuals post-stroke with a shorter step length on the affected leg would produce an aftereffect consisting of a decrease in step length of the affected leg, resulting in a greater level of step length asymmetry during the post-adaptation period. These hypotheses have been tested using robotic-generated swing assistance (force assists the affected leg forward during the swing phase of gait) and swing

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resistance (force resists the affected leg from moving forward during the swing phase of gait) loads.

2. Methods

2.1 Subjects

Ten subjects with chronic stroke (> 6 months) were recruited in this study (Table 1). In the recruitment interview, we visually inspected subjects' step length symmetry during overground walking. We included subjects who showed an observable shorter step length with the affected leg. Other inclusion criteria for participation in the study included: a) age between 18 and 75 years; b) history of unilateral, supratentorial, ischemic or hemorrhage stroke with lesion location confirmed by radiographic findings; c) unilateral lower extremity paresis; d) able to walk independently with or without the use of assistive devices or orthoses. Exclusion criteria include a) significant cardiorespiratory/metabolic disease, or other neurological or orthopedic injury that may limit exercise participation or impair locomotion; b) scores on the Mini Mental Status examination (MMSE) < 24 (Folstein, Folstein, and McHugh, 1975); c) stroke of the brainstem or cerebellar lesions; d) uncontrolled hypertension (systolic > 200 mm Hg, diastolic > 110 mm Hg); e) botox injection within 6 months of starting the study. Subjects were excluded if they were unable to tolerate 30 minutes of standing. Before participation, all subjects signed the informed consent approved by the Institutional Review Board of Northwestern University.

Case	Age	Years Post Injury	Affected Side	*Test Speed (m/s)	Resistive Load (N)	Assistive Load (N)	Assistive Device Used during Experiment
S1	67	17	Left	0.4	16	14	AFO
S2	64	20	Left	0.5	14	24	AFO
S3	67	13	Left	0.68	18	24	None
S4	63	20	Right	0.3	21	18	None
S5	57	24	Left	0.66	26	22	None
S6	64	9	Right	0.6	18	18	AFO
S7	67	1	Right	0.45	18	18	AFO
S8	36	5	Left	0.55	18	23	None
S9	51	2	Right	0.6	18	18	None

Table 1. Subject Information

Case	Age	Years Post Injury	Affected Side	*Test Speed (m/s)	Resistive Load (N)	Assistive Load (N)	Assistive Device Used during Experiment
S10	54	4	Right	0.61	28	32	None
*Tost s	nood v	was sot a	t oach subi	act's sale	octed comfo	rtahla snaad	during treadmill

*Test speed was set at each subject's selected comfortable speed during treadmill walking

2.2 Procedures

Subjects participated in two data collection sessions. In the first session, subjects walked with swing resistance applied to the affected leg, and in the second session subjects walked with swing assistance applied to the affected leg. The two data collection sessions were scheduled at least 2 weeks apart to minimize any lingering carryover. We used a customized cable-driven robot to generate the controlled assistance and resistance forces, see Figure 1, which has been described previously (Wu, Hornby, Landry, Roth, & Schmit, 2011). In brief, four nylon-coated stainless-steel cables (diameter 1.6 mm), driven by four motors, are affixed to custom braces that are strapped to the subject's affected leg at the ankle to provide controlled assistance or resistance forces to the legs while the subject walks on a treadmill. The cable driven system is compliant and highly backdrivable (Wu et al., 2011), which allows subjects to make and correct errors across steps.



Figure 1. Illustration of the cable robot and the application of the swing resistance and swing assistance.

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Two custom designed 3D position sensors were attached at the leg above the ankle and were used to record ankle position during treadmill walking. The ankle position signals were used to trigger swing assistance/resistance approximately from late stance (\sim 10% before toe off) to mid-swing (i.e., \sim 50% of swing phase). The timing of the force application was determined based on each subject's *average* stance and swing times determined prior to data collection using a custom designed program.

During data collection, subjects went through baseline, adaptation, and post-adaptation periods continuously while walking on a treadmill at a self-selected comfortable speed. An overhead harness was used for safety only with no body weight support was provided. In the baseline period, the force was turned off while subjects walked for one minute. In the adaptation period, the force was turned on and subjects walked with either a resistance or assistance applied to their affected leg, depending on test condition, for seven minutes. In the post-adaptation period, the force perturbation was unexpectedly removed while subjects continued walking on the treadmill for another two minutes.

While holding onto the support rail could potentially affect the walking pattern, subjects were allowed to use the support rail during walking for safety reasons. When the robot was in the "off" mode during the baseline and post-adaptation periods, there was still \sim 4 N of pretension applied to the cable. This pretension force is required to prevent the cable from slacking and interfering with the subject's walking on a treadmill.

The magnitudes of the resistance and assistance forces were both set at ~18% of the hip flexors maximal voluntary contraction (MVC) of the affected side, although it was adjusted based on each individual's tolerance (Table 1). In pilot testing, we found that 18% of the hip flexors MVC was approximately the maximum amount of resistance that individuals with neurological disorders could comfortably tolerate for ~10 minutes of walking, without significant fatigue. The procedure of measuring MVC is described in detail in a previous study (Yen et al., 2013). The average peak forces were 21.1 \pm 5.0 N and 19.5 \pm 4.4 N for assistance and resistance conditions,

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respectively, with no significant difference between the two conditions (p = 0.29, ANOVA).

2.3 Data analysis

The primary outcome measures of this study were step length of the affected and sound legs and step length symmetry while walking on a treadmill. The secondary outcome measures were swing time of the affected and sound legs and swing time symmetry. These measures were derived from subjects' ankle trajectory during treadmill walking, which were recorded using two ankle position sensors (Figure 1) (Yen et al., 2012). We defined heel contact as the point when the ankle trajectory changed its moving direction from forward to backward, while we defined toe-off as the time when the ankle trajectory changed its moving direction from backward to forward (Zeni, Richards, & Higginson, 2008). Step length was quantified as the horizontal distance between the two legs' ankle positions at heel contact. Swing time was quantified as the time between toe-off and heel contact normalized to the gait cycle time. Symmetry was calculated using the following symmetry index (SI):

$$SI = \frac{Xa - Xs}{0.5 \times (Xa + Xs)} \tag{1}$$

where Xa is the affected leg data and Xs is the sound leg data (Sadeghi, Allard, Prince, & Labelle, 2000). A SI score that is closer to 0 indicates a more symmetrical gait.

2.4 Statistical Analysis

Each outcome measure was averaged across the last 20 cycles of the baseline period (baseline value), across the first 3 cycles of adaptation period (early adaptation value), across the last 3 cycles of adaptation period (late adaptation value), and across the first 3 cycles of post-adaptation period (post-adaptation value). The baseline value was determined using more cycles in order to obtain a more robust baseline, in consideration of the variability of gait parameters in stroke survivors. The early, late, and post-adaptation values were determined

based on 3 cycles because: (a) the error in step length or swing time caused by the force perturbation during adaptation could be counteracted in a few steps; (b) the aftereffect after removal of the perturbation during post-adaptation could be washed out in a few steps. This method has been used by other investigators (Malone & Bastian, 2014).

We used Generalized Estimating Equations (GEE) to compare the values from the four different time periods. We did not use traditional repeated measures ANOVA for the analysis as it assumes correlations between repeated observations are equal, which is not necessarily correct. The GEE allowed us to model the correlation between repeated observations, and here we used an "unstructured" matrix to allow the observed data to dictate the correlations between repeated measurements. When significance was detected, post-hoc tests with sequential Bonferroni adjustment were conducted to compare the values during adaptation and post-adaptation to baseline. All statistical analyses were conducted using SPSS version 19 (IBM SPSS, Chicago, IL). The significance level for all analyses was set at P < 0.05.

3. Results

Case 8 was removed from data analysis as we found this subject showed a change in the direction of leg symmetry (step length of the affected leg became larger) when walking with the cable robot even during the baseline period. The results presented were based on 9 subjects.

3.1 Step length symmetry: resistance condition

In the resistance condition, step length of the affected leg decreased during the early adaptation period (i.e., augmented error), returned to the baseline (i.e., error correction) or slightly dropped below the baseline during the late adaptation period, increased above the baseline following the release of the resistance, and gradually returned to the baseline after approximately 15 steps (Figures 2A). The GEE detected significance differences in step length of the affected leg during the baseline, early adaptation, late adaptation, and post

adaptation periods (p < 0.01). Post-hoc tests indicated significant differences in step length of the affected leg during the early adaptation and post adaptation periods vs. baseline (Figure 2B). Specifically, the step length of the affected leg decreased 1.25 ± 0.46 cm (mean \pm standard error, same for below) from the baseline (p = 0.01) during the early adaptation period, decreased 1.99 ± 1.32 cm from the baseline (p = 0.33 > 0.05 due to large variability) during the late adaptation period, and increased 3.54 ± 0.72 cm from the baseline (p = 0.01) during the post adaptation period.



Figure 2. Kinematic data from the resistance condition. (A) Cycle-to-cycle variation of the mean step length of each leg (N = 9). (B) Changes in step length in comparison to the baseline. (C) Cycle-to-cycle variation of the mean step length SI (N = 9). (D) Changes in step length symmetry index in comparison to the baseline. (E) Cycle-to-cycle variation of the mean swing time of each leg (N = 9). (F) Changes in swing time in comparison to the baseline. (G) Cycle-to-cycle variation of the mean swing time symmetry index (N = 9). (H) Changes in swing time symmetry index in comparison to the baseline. Error bars represent standard error. *p < 0.05. EA = Early Adaptation; LA = Late Adaptation; PA = Post Adaptation. SI = Symmetry Index.

Step length of the sound leg had slight increases during the early adaptation period, and further increased during the late adaptation period (Figure 2A). The step length of the sound leg

returned to a level close to baseline within one step during the post adaptation period (Figure 2A). The GEE detected significant differences in step length of the sound leg during the baseline, early adaptation, late adaptation, and post adaptation periods (p = 0.03). Post hoc tests indicated that the step length of the sound leg significantly increased 3.6 ± 1.33 cm (p = 0.01, Figure 2B) from the baseline during the late adaptation period. There was no significant difference in step length of the sound leg during the early adaptation period vs. baseline (modestly increased 0.07 ± 0.78 cm, p = 1.00), and during the post adaptation period vs. baseline (increased 1.45 ± 1.11 cm, p = 0.5), Figure 2B.

The symmetry of step length had a modest change during the early adaptation period (i.e., SI slightly further deviated from 0) (Figure 2C), and became more asymmetrical compared to baseline during the late adaptation period. When the resistance was removed during the post adaptation period, the step length became more symmetrical compared to the baseline, and this improved symmetry of step length was retained for approximately 10 steps.

The GEE detected significant differences in the symmetry of step length during the baseline, early adaptation, late adaptation, and post adaptation periods (p = 0.01). Post hoc tests indicated that the symmetry of step length during the late adaptation period and post adaptation period were significantly different from baseline (Figure 2D). Specifically, the SI was 0.11 ± 0.04 units below baseline (p =0.01) during the late adaptation period and was 0.06 ± 0.03 units above baseline value (p = 0.045) during the post adaptation period. There was no significant difference in the symmetry of step length during the early adaptation period in comparison with baseline (the SI was 0.04 ± 0.02 units below baseline, p = 0.32).

3.2 Swing time symmetry: resistance condition

In the resistance condition, subjects demonstrated greater swing time on the affected leg than the sound leg during the baseline, early adaptation, late adaptation, and post adaptation periods (Figure 2E). Compared to baseline, swing time of the affected leg increased during the early adaptation period for approximately 8 steps before returning towards baseline. During the late adaptation period, swing

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time of the affected leg was slightly above the baseline. When the resistance was removed during the post adaptation period, swing time of the affected leg slightly dropped below the baseline.

The GEE detected significant differences in swing time of the affected leg during baseline, early adaptation, late adaptation, and post adaptation periods (p = 0.045). Post hoc tests indicated that the swing time of the affected leg during the early adaptation period was greater than the baseline by 1.1 ± 0.3 % of the gait cycle (p = 0.02). There were no significant differences in the swing time of the affected leg during the late adaptation period vs. baseline (slightly increased 0.72 ± 1.14 % of the gait cycle; p = 1.00), and during the post adaptation period vs. baseline (slightly decreased 0.63 ± 0.91 % of the gait cycle; p = 1.00), Figure 2F.

There was a modest change in swing time of the sound leg across the four different periods examined (Figure 2E). The GEE detected no significant difference in swing time of the sound leg between the baseline, early adaptation, late adaptation, and post adaptation periods (p = 0.09).

Compared to baseline, swing time became more asymmetrical when resistance was applied, especially in the first few steps, during the early adaptation period (Figure 2G). Following removal of the resistance, swing time symmetry returned towards baseline during the post-adaptation period. The GEE detected significant differences in the swing time symmetry index between the baseline, early adaptation, late adaptation, and post adaptation periods (p = 0.004). Post hoc tests indicated that the swing time symmetry index during the early adaptation period was greater than the baseline by 0.04 ± 0.01 units (p = 0.01). There were no significant differences in swing time symmetry index during the late adaptation period vs. baseline (slightly increased 0.03 ± 0.03; p = 0.49), and during the post adaptation period vs. baseline (slightly decreased 0.02 ± 0.02; p = 0.45), Figure 2H.

3.3 Step length symmetry: assistance condition

Step length of the affected leg increased when swing assistance was applied during the early adaptation period, and further increased

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during the late adaptation period (Figures 3A). Following the release of assistance, step length of the affected leg decreased, although it remained slightly above baseline during the post-adaptation period. The GEE detected significant differences between the step length of the affected leg during the baseline, early adaptation, late adaptation, and post adaptation periods (p < 0.01). Post hoc tests indicated that the step length of the affected leg during the early adaptation and late adaptation periods was significantly different from baseline (Figure 3B). Specifically, the step length of the affected leg increased 3.2 ± 0.89 cm from baseline during the early adaptation period (p < 0.01), and increased 5.5 ± 1.15 cm from baseline during the late adaptation period (p < 0.01). There was no significant difference in step length of the affected leg during the post adaptation period vs. baseline (increased 1.92 ± 1.86 cm; p = 0.82).



Figure 3. Kinematic data from the assistance condition. (A) Cycle-to-cycle variation of the mean step length of each leg (N = 9). (B) Changes in step length in comparison to the baseline. (C) Cycle-to-cycle variation of the mean step length SI (N = 9). (D) Changes in step length symmetry index in comparison to the baseline. (E) Cycle-to-cycle variation of the mean swing time of each leg (N = 9). (F) Changes in swing time in comparison to the baseline. (G) Cycle-to-cycle variation of the mean swing time symmetry index (N = 9). (H) Changes in swing time symmetry index in comparison to the baseline.

the baseline. Error bars represent standard error. *p < 0.05. EA = Early Adaptation; LA = Late Adaptation; PA = Post Adaptation. SI = Symmetry Index.

Step length of the sound leg slightly decreased during the early adaptation period, returned towards baseline during the late adaptation period, and increased above baseline during the postadaptation period (Figure 3A). The GEE detected significant differences in the step length of the sound leg between the baseline, early adaptation, late adaptation, and post adaptation periods (p = 0.01). Post hoc tests indicated that the step length of the sound leg during the post adaptation period was significantly greater than the baseline by 2.08 ± 0.78 cm (p = 0.01, Figure 3B). There was no significant difference in the step length of the sound leg during the early adaptation period vs. baseline (decreased 0.63 ± 1.74 from baseline; p = 1.00), and during the late adaptation period vs. baseline (slightly increased 0.44 ± 0.62 from baseline; p = 1.00), Figure 3B.

Step length symmetry improved when swing assistance was applied during the early and late adaptation periods (SI moved towards to 0, Figures 3C), and returned to baseline after release of assistance during the post-adaptation period. The GEE detected significant differences in step length symmetry between the baseline, early adaptation, late adaptation, and post adaptation periods (p < 0.01). Post hoc tests indicated that the step length symmetry during the early adaptation and late adaptation periods were significantly different from baseline (Figure 3D). Specifically, the SI during the early adaptation period was 0.10 ± 0.04 units above baseline (p = 0.03), and the SI during the late adaptation period was 0.12 ± 0.02 units above baseline (p < 0.01). There was no significant difference in step length symmetry during the post adaptation period vs. baseline (SI slightly decreased 0.007 ± 0.05 ; p = 1.00).

3.4 Swing time symmetry: assistance condition

No apparent trend of change was noticed in swing time for either the affected or sound leg when swing assistance was applied during the early and late adaptation periods (Figures 3E), and when swing assistance was released during the post-adaptation period. The GEE detected no significance difference in swing time across the four different test periods for both the affected leg (p = 0.05) and the

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sound leg (p = 0.53). The differences in swing time of both the affected and sound legs during the early adaptation, late adaptation, and post adaptation periods vs. baseline were below 1% of the gait cycle, see Figure 3F.

Swing time symmetry showed no apparent change when swing assistance was applied during the early and late adaptation periods (Figure 3G). Following release of assistance, swing time symmetry had modest changes during the post-adaptation period. While the GEE detected significant differences in swing time symmetry between the baseline, early adaptation, late adaptation, and post adaptation periods (p = 0.01), Figure 3H, post hoc tests indicated there were no significant differences in swing time symmetry during the early adaptation period vs. baseline (slightly decreased 0.03 ± 0.02 ; p = 0.62), during the late adaptation period vs. baseline (slightly decreased 0.02 ± 0.02 ; p = 0.40), and during the post adaptation period vs. baseline (slightly increased 0.03 ± 0.02 ; p = 0.42).

4. Discussion

We examined how stroke survivors with a shorter step length of the affected leg responded to unilateral swing resistance and assistance applied to the affected leg during treadmill walking. Subjects counteracted step length deviation caused by swing resistance during the adaptation period and produced an aftereffect consisting of an increase in step length, resulting in an improvement in step length symmetry during the post-adaptation period. In contrast, subjects did not counteract step length deviations caused by swing assistance during the adaptation period, as it worked to their advantage to use the assistance provided. No aftereffect was produced and therefore no significant changes in step length symmetry were observed during the post-adaptation period. These results suggest that different motor learning mechanisms may be involved during locomotor adaptation to swing resistance and assistance forces applied to the affected leg of individuals post-stroke.

Error-based motor learning mechanisms may be involved during locomotor adaptation to swing resistance applied to the affected leg for stroke subjects with a shorter step length of the affected leg. While swing resistance initially augmented errors in step length of the

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affected leg (i.e., further shortened the step length) during the early adaptation period, subjects gradually adapted to the swing resistance force and counteracted the errors induced by it, although further slightly decreased (may be due to the fatigue) during late adaptation period. Physiologically, the underlying motor learning process may be occurring in the neural system as feed-forward control command is modified in anticipation of swing resistance in upcoming steps, resulting in an aftereffect consisting of an increase in step length with the affected leg during the post-adaptation period. The increase in step length leads to an improvement in step length symmetry, which we observed in individuals post-stroke after 7 minutes of treadmill training with swing resistance applied to the affected leg.

Error information from sensorimotor feedback, particularly afferent feedback from hip flexor muscles and proprioceptors (Lam and Pearson 2001) of the affected leg, may drive alterations and continuously update the internal representation (Smith and Shadmehr 2005) of the desired step length of the affected leg. It was suggested that the cerebellum may be a key structure involved in error-based motor learning mechanisms during locomotor adaptation, although many other regions of the brain and spinal cord may also be involved (Morton and Bastian 2006; Drew et al. 2004).

In contrast, for stroke survivors taking a longer step length with the affected leg, applying swing resistance to the affected leg may initially decrease step length of the affected leg during the early adaptation period (due to the added force in the opposite direction of swing), but may result in an increase in step length with the affected leg during the post-adaptation period (Savin et al., 2013) as the subject adapts to the added resistance. This would result in a decrease in step length symmetry during the post-adaptation period for these subjects. Thus, applying swing resistance to the affected leg during treadmill walking may or may not improve step length symmetry of individuals post-stroke, depending on the direction of step length symmetry at baseline.

On the other hand, a use-dependent motor learning mechanism may be involved in the swing assistance condition (Diedrichsen et al., 2010). In this condition, step length of the affected leg further increased with swing assistance, rather than returning to the baseline

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level during the late adaptation period, indicating that neural systems did not generate additional torgues to counteract the deviation in step length of the affected leg induced by swing assistance. Instead, subjects walked with swing assistance and demonstrated a greater level of step length symmetry, which may be driven by the improved energy efficiency of symmetric gait (Ellis et al., 2013). Following the release of swing assistance, step length of the affected leg did not fall below the baseline, i.e., no aftereffect was observed. Thus, we postulate that use-dependent motor learning mechanisms (instead of error-based motor learning mechanisms) may be more involved in the swing assistance condition, although other factors, such as floor effect for these subjects with shorter step length on the affected side may also be involved. In contrast to error-based motor learning, which attempts to cancel the perturbation (i.e., swing assistance), usedependent motor learning modifies movement in the direction of a perturbation (Diedrichsen et al. 2010).

Our observations in the changes of step length of the affected leg following swing assistance were not consistent with results from a previous study using a split-belt treadmill paradigm, which showed that when the belt was used to "correct" step length asymmetry during adaptation, individuals post-stroke only showed improvement in the early adaptation period, but returned towards baseline asymmetry during late adaptation (Lauziere et al. 2014; Reisman et al. 2007). One of possible reasons for this difference may be due to the energy cost requirements needed to counteract perturbations induced by swing assistance vs. the split-belt treadmill, which are introduced during swing and stance phase of gait, respectively (Gottschall and Kram 2005; Finley et al. 2013). In the current study, individuals poststroke walked with a more symmetrical gait pattern during the early adaptation period when swing assistance was applied to the paretic leg. The reason why subjects did not return to the baseline asymmetrical gait pattern during the late adaptation period may be because this symmetrical gait pattern is more energy efficient than the asymmetrical gait pattern during baseline (Finely et al. 2013). However, when a perturbation was applied through the split-belt treadmill, for instance, and the subject's stepping pattern on the paretic leg was facilitated by a slower speed belt (compared to the belt speed under the non-paretic leq), the stance time of the non-paretic leg decreased (due to the faster belt speed), as well as an increase in

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the step length and stance time of the paretic leg (Reisman et al., 2007), resulting in an improvement in step length and stance time symmetry, during early adaptation period. However, an increase in step length of the paretic leg may increase the energy cost because the metabolic power increases as step length becomes longer (Donelan et al. 2002; Kuo, 2002). In addition, a decrease in stance time on the non-paretic leg may reduce the energy input from the non-paretic leg, and require more energy input from the paretic leg, which is less efficient in most cases. The energy cost required for leg stance is much greater than that for leg swing (Gottschall and Kram 2005). Thus, it is possible that the symmetrical gait pattern induced by split-belt during early adaptation period may be less energy efficient than the baseline asymmetrical gait pattern. As a consequence, the neural system of individuals post-stroke may modify spatial-temporal parameters to reach a more energy efficient gait pattern, even though it is not an "optimal" gait pattern, during the late adaptation period because energy cost is one of crucial factors that drive locomotor adaptation (Finely et al. 2013).

Applying unilateral swing resistance may induce a bilateral response in both perturbed and unperturbed legs in healthy adults during treadmill walking (Dietz et al. 1986; Eng et al. 1994). Further, repeated exposure to swing resistance may induce motor adaptation and show aftereffect from both perturbed and unperturbed legs, suggesting that a feed-forward control mechanism may be involved in this adaptation process (Savin et al. 2010). However, in the current study, we did not observe significant changes in the step length of the unaffected leg (i.e., unperturbed leg) during the post-adaptation period. While this could be simply due to the fact that the perturbation force was not strong enough, it may suggest a potential impairment in coordination across the two legs for stroke survivors.

While swing resistance induced an aftereffect of increased step length in the affected leg, it did not induce an aftereffect in swing time. This is consistent with our previous studies, in which the force perturbation generated by the cable robot did not induce an aftereffect in swing time (Yen et al, 2012; 2013). A possible mechanism may be due to different neural pathways being involved in the control of spatial and temporal parameters of gait in individuals post-stroke during walking (Yen et al., 2012). This argument is supported by

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results from previous studies in healthy controls. For instance, children develop the ability to adapt temporal parameters earlier than they do to spatial parameters (Vasudevan et al. 2011). In addition, applying distraction during locomotor adaptation to split-belt perturbation affects the adaptation rate of spatial parameters but not of temporal parameters in healthy controls (Malone and Bastian 2010). These results collectively suggest that different adaptive approaches may be needed for improving spatial vs. temporal parameters of individuals post-stroke. For instance, applying swing resistance/assistance may be more effective in modulating spatial than temporal parameters of individuals post stroke.

The results of this study may have clinical applications. Locomotor training has been used in clinics to help individuals post stroke improve walking function. Results from our previous study provided initial evidence to support that long-term locomotor training with either swing resistance or swing assistance applied to the affected leg can improve walking function (as measured by walking speed) in stroke survivors (Wu, Landry, Kim, Schmit, Yen, & Macdonald, 2014). The results from the current study suggest that applying resistance load to the affected leg may be used for improving gait symmetry for individuals post-stroke taking a shorter step length with the affected leg. It remains unclear whether swing assistance may be used to improve step length symmetry for these same individuals post-stroke, although swing assistance did not lead to a deterioration in step length symmetry after swing assistance training, which was opposite of one of our hypotheses.

While the changes in step length (and therefore symmetry) induced by one session of swing resistance were small in magnitude and are not likely to be retained for a long period of time, results from this study may be used to develop a long-term robotic training paradigm for improving step length symmetry in individuals poststroke taking a shorter step length with their affected leg. Understanding the motor learning mechanism underlying the improvements in step length and step length symmetry that we observed with the application of swing resistance, as well as identifying a subset of subjects post-stroke that may benefit from this method of training are critical in maximizing treatment outcomes

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through the development of evidence-based treatments for individuals post-stroke.

One limitation of this study was that subjects were allowed to hold onto a handrail during treadmill adaptation for safety reasons. The pulling force from the subject's arms could confound the results, especially with the resistance condition. Also, we only tested a single magnitude of resistance/assistance. How different amounts o f resistance/assistance affect error correction and the subsequent aftereffect is unknown. In addition, subjects tested in the current study were stroke survivors taking a shorter step length with the affected leg. Thus, there is a limited subset of individuals post-stroke that this approach will be successful with. For individuals post-stroke taking a longer step length with the affected leg, applying swing resistance to the affected leg may worsen symmetry in step length during the post-adaptation period (Savin et al. 2013).

We did not randomize the resistance and assistance conditions in this experiment, and were not able to rule out all potential order effects on the results. However, the adaptation period was short (7 minutes) in each condition, and the resulting aftereffect was not likely to persist for two weeks. Thus, we do not believe that our results were systematically biased by the test order. In addition, a pretension force $(\sim 4 \text{ N})$ was applied to the cable during the baseline period, otherwise, the cable would slack and interfere with stepping. This pretension force may have affected the step length symmetry of some lower functioning subjects during the baseline period because the weaker paretic leg muscles of these subjects. For instance, for case 8 the direction of the symmetry changed when walking with the cable robot during the baseline period, and was removed from analysis. However, because this cable pretension force was retained during both the baseline and post adaptation periods, our results may not be systematically biased by this technique limitation as the force was consistent throughout. The sample size of this study was small, and we did not measure neurological impairment of each subject. Therefore, we are not able to draw a conclusion about the relationship between the level of impairment and direction of adaptation. Lastly, 4 subjects wore ankle foot orthosis (AFO) during the test, which may have affected their ability to adapt to swing resistance/assistance. However, we were not able to make a conclusion about the impact of

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an AFO on the locomotor adaptation of individuals post-stroke due to the small sample size of subjects of this study.

Conclusion

We investigated how stroke survivors respond to the application of swing resistance and assistance during treadmill walking. Subjects counteracted errors induced by swing resistance and produced an aftereffect that improved step length symmetry during the postadaptation period. In contrast, subjects did not counteract deviations caused by swing assistance and produced no aftereffect during the post-adaptation period. Results of this study suggest that locomotor training with swing resistance may improve step length symmetry through error-based learning. On the other hand, swing assistance reduced error in leg swing and may facilitate use-dependent motor learning. Results from this study may be used to develop individual robotic gait training paradigms for improving gait symmetry of stroke survivors.

Highlights

- 1. Swing resistance/assistance was added to the stroke patients' affected leg during treadmill walking.
- 2. Step length symmetry improved following removal of swing resistance but no aftereffects were observed following removal of swing assistance.
- 3. Swing resistance and assistance may engage different motor learning pathways.

Acknowledgement: This work was supported by the NIH under Grant 1R21HD058267 (Wu). We thank Miss Jill Landry for her comments.

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