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Motor adaptation to lateral pelvis assistance force during treadmill walking in individuals post-stroke

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Abstract

The goal of this study was to determine how individuals post-stroke response to the lateral assistance force applied to the pelvis during treadmill walking. Ten individuals post chronic (> 6 months) stroke were recruited to participate in this study. A controlled assistance force (~10% of body weight) was applied to the pelvis in the lateral direction toward the paretic side during stance of the paretic leg. Kinematics of the pelvis and legs were recorded. Applying pelvis assistance force facilitated weight shifting toward the paretic side, resulting in a more symmetrical gait pattern but also inducing an enlarged range of motion of the pelvis during early adaptation period. The neural system of individuals post stroke adapted to the pelvis assistance force and showed an aftereffect consists of reduced range of motion of the pelvis following load release during post adaptation period.

Keywords

Stroke; locomotion; pelvis; symmetry; motor adaptation

I. Introduction

One of the goals of individuals post stroke is to improve ambulation and gait function because impairments in gait function may significantly impact the quality of life of these patients. Gait function, represented by gait speed and endurance, is highly correlated to gait asymmetry. Gait asymmetry is commonly observed in individuals post stroke and represented as a shorter duration of support time on the paretic side in comparison to the non-paretic side [1], and less weight shifting toward the paretic side [2]. In addition, many stroke survivors walked with a large lateral pelvis movement [3], which is less efficient gait pattern because increased lateral movement of the pelvis may increase the energy cost [4].

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Applying lateral assistance force to the pelvis may facilitate weight shifting toward the paretic side, resulting in a more symmetrical gait pattern and forcing subject to use more of their paretic leg [5]. However, enhanced weight shifting toward the paretic side may also increase the displacement of the pelvis toward the paretic side, resulting in a large range of motion of the pelvis, which may be less efficient due to the cost of energy induced by enhanced lateral movement of the pelvis [4]. Given the energy cost is one of key criterion that drive locomotor adaptation [6], we hypothesized that stroke subjects would gradually adapt to the pelvis assistance force and reduce the magnitude of displacement toward the paretic side, i.e., reducing the range of motion of the pelvis during the late adaptation period.

II. Method

A. Subjects

Ten subjects with chronic stroke (> 6 months) were recruited from the Sensory Motor Performance Program (SMPP) stroke database and the Rehabilitation Institute of Chicago (RIC) outpatient clinic for this study. Demographic information for these subjects is shown in Table 1. Six participants are male and 4 are female. The average age was 51.4 ± 13.1 years old, and time post stroke was 8.1 ± 6.1 years.

The inclusion criteria were: (1) age ranged from 21 to 75 years old, (2) unilateral, supratentorial, ischemic or hemorrhagic stroke confirmed with radiographic data, (3) no prior history of stroke, (4) independent ambulation with/without the use of assistive device or below knee orthoses, (5) self-selected walking speed ≤ 0.80 m/s.

Exclusion criteria were: (1) brainstem or cerebellar stroke, (2) a score on the Mini Mental Status examination < 24 [7], (3) other neurological conditions, cardiorespiratory/metabolic disorders, or orthopedic conditions affecting ambulation ability, (4) botox injection within 6 months of study visit.

Written and verbal information about the study procedure were provided for all subjects before giving written consent. The study was approved by the Institutional Review Board of the Medical School of Northwestern University.

B. Experimental Apparatus

A customized cable-driven robotic system was used to provide a controlled assistance force to the pelvis in the lateral direction toward the paretic side while subjects walked on a treadmill (Woodway, WOODWAY USA Inc., Waukesha, WI), Figure 1, which has been reported before [8]. In brief, the robotic system consists of 2 nylon-coated stainless-steel cables (1.6 mm), driven by 2 motors (AKM 33H, Kollmorgen, Radford, VA) through 2 cable spools that are located on the side of the treadmill. The cables were attached to a custom waist belt that was strapped to subjects' pelvis to provide an assistance force at targeted phase of gait. Two sets of custom designed 3-dimensional position sensors were used to record the pelvis and ankle positions. Each sensor consists of a retractable detection rod and 2 universal joints attached to the ends of the rod. Two potentiometers (P2201, Novotechnik, Southborough, MA) were used to measure rotational movements of the rod in the anterior-posterior and medial-lateral directions; one potentiometer (SP-2, Celesco, Chatsworth, CA) was used to measure the linear movement of the retractable detection rod in the diagonal direction. The recorded ankle position signals were used to trigger the pelvis assistance force applied from initial contact of the paretic leg, which was defined as the timing when the ankle position changed its movement direction from forward to backward direction, for 400 ms. A custom-written LabVIEW program was used to collect pelvis and ankle position signals at 500 Hz, as well as to send force command signals to the driver to generate controlled assistance.

C. Experimental protocol

An overhead harness was used for protection only (no body weight support was provided) while subjects walked on a treadmill. The treadmill speed was set at self-selected comfortable walking speed for each subject, which was determined by a research physical therapist at the beginning of test session. Each subject walked on a treadmill with no load for 1 minute, i.e., baseline, then a controlled pelvis assistance load was applied to facilitate weight shifting toward to the paretic side for 7 minutes. After this, the load was released without warning and subjects were instructed to continue walking on the treadmill for another 1 minute. The peak magnitude of pelvis assistance force was set at ~10% of body weight, although adjusted based on the tolerance of each subject. For the sake of safety, subjects were allowed to hold on a front bar while walking on a treadmill, although they were requested to refrain from holding on the side bar.

Kinematic data of the pelvis and legs were recorded using position sensors starting from the baseline to the end of session with the sample frequency at 500 Hz.

D. Data Analysis

Custom written program in Matlab (The Mathworks, Natick, MA) was used for all data analysis. All kinematic data were smoothed using a 4^{th} order Butteworth low-pass filter (cut-off frequency: 8 Hz) with zero lag. The ankle position signals were used to segment the gait cycles. Specifically, toe-off was defined as the time during which the ankle position, which was measured using ankle position sensor, changed its moving direction from backward to forward; heel-contact was defined as the time during which the ankle position changed its moving direction from forward to backward [9]. Stance time was calculated as the time from initial contact to toe-off and normalized to the whole gait cycle time. Step length was calculated as the anterior-posterior distance between two feet at the time of initial heel contact. In addition, because there was a variation in foot position on a treadmill belt in the mediolateral direction from one gait cycle to the next cycle, we defined the mid-point between two feet in the mediolateral direction at the time of initial contact as a reference point for each gait cycle. The pelvis lateral displacement relative to the reference point was calculated by subtracting the mid-point from the gross pelvis lateral displacement, which was measured using two pelvis position sensors.

The range of motion of the pelvis in the mediolateral direction was used to quantify the motor adaptation to pelvis assistance force during walking in individuals post-stroke. The symmetry index of pelvis displacement was quantified as 1-maxdP/maxdNP, where the maxdP is the maximum displacement of the pelvis in the mediolateral direction to the paretic side, and maxdNP is the maximum displacement of the pelvis to the non-paretic side. A symmetry index of 0 indicated a perfect symmetrical pelvis displacement from side to side. Similarly, the symmetry indexes of stance time and step length were defined as 1-timestP/timestNP, where timestP is the stance time of the paretic leg, and timestNP is the stance time of the non-paretic leg, and 1-steplengthP/steplengthNP, where steplengthP is step length of the paretic leg, and steplengthNP is the step length of the non-paretic leg. The symmetry indexes of stance time, step length, and pelvis displacement during the course of motor adaptation were quantified as the average of 5 steps before load (baseline), the first 5 steps after load was applied (early adaptation), the last 5 steps before load release (late adaptation), first 5 steps after load was released (post adaptation), and the 20th to 25th steps after load was released (late post adaptation).

Repeated measure ANOVA was used to compare range of motion of the pelvis, symmetry ratios of the pelvis displacement, stance time, and step length between baseline and 4 other sessions, with significance noted at $p < 0.05$. If the ANOVA revealed significant differences, Tukey-Kramer post-hoc tests were used to identify differences between conditions, again with significance noted at $p < 0.05$.

III. Results

Data from 10 subjects were analyzed. The peak displacement of the pelvis in the mediolateral direction from one typical subject is shown in Figure 2. The subject shifted the pelvis more toward the paretic side during early adaptation period when a pelvis assistance force was applied, Figure 2A. As a result, the range of motion of the pelvis in the mediolateral direction increased during early adaptation period, see Figure 2B. In addition, the symmetry of the pelvis displacement also improved, i.e., smaller symmetry index ratios, during this time of period. Then, the subject gradually adapted to the pelvis assistance force. The peak displacement of the pelvis toward the paretic side gradually reduced to a level that was similar to that during baseline. As a result, the range of motion of the pelvis also decreased during late adaptation period. Following the load release, we observed that the peak displacement of the pelvis toward the paretic side further decreased, resulting in a reduced range of motion of the pervious displacement during post adaptation period, see Figure 2B.

The average of the range of motion of the pelvis displacement across all subjects is shown in Figure 3. The pelvis assistance force had a significant impact on the range of motion of the pelvis ($F(4, 9) = 45.0$, $p < 0.001$, ANOVA). Post-hoc tests indicated that the range of motion of the pelvis significantly increased during early adaptation period, i.e., increased from 108.2 ± 22.0 mm at baseline to 154.4 ± 25.3 mm (45.4% increase, $p < 0.001$), and significantly decreased during post adaptation period, i.e., decreased to 88.8 ± 15.2 mm (17.2% decrease, $p = 0.01$). The reduction in the range of motion of the pelvis partially retained during late post adaptation period, although this was not significant (i.e., 93.9 ± 17.7 mm, $p = 0.09$).

Applying pelvis assistance force toward the paretic side also had a significant impact on the symmetry of the pelvis lateral movement ($F(4, 9) = 13.9$, $p < 0.001$), see Figure 4A. Post-hoc tests indicated that the symmetry index of the pelvis displacement significantly improved when pelvis assistance force was applied during early adaptation period compared to baseline (baseline: 0.80 ± 0.41, early adaptation: 0.27 ± 0.61, $p = 0.002$). The symmetry index had a tendency to increase between baseline to post adaptation period, but did not reach significance ($p = 0.07$).

Applying pelvis assistance force toward the paretic side had a significant impact on the symmetry index of the step length ($F (4, 9) = 2.8$, $p = 0.04$), see Figure 4B. Post-hoc tests indicated no significant difference between different conditions ($p > 0.05$). In addition, applying pelvis assistance for had no significant impact on the symmetry index of the stance time ($F (4, 9) = 0.61$, $p = 0.66$).

IV. Discussion

Applying lateral pelvis assistance force toward the paretic side may facilitate weight shifting toward the paretic leg during stance, which may result in a more symmetrical pelvis lateral movement but also induce a larger range of motion of the pelvis during early adaptation period. Pelvis assistance may also improve the symmetry of the step length during the early adaptation period. Individuals post-stroke may gradually adapt to the pelvis assistance force during late adaptation period, and show an aftereffect that consists of reduced range of the motion of the pelvis after load release during post adaptation period. The symmetry of the pelvis and step length had no significant changes during post adaptation period.

Many individuals post-stroke have impaired weight shifting toward the paretic side, and walk with an asymmetrical pelvis movement in the mediolateral direction [2]. Results from this study indicated that applying lateral assistance force to the pelvis may improve the symmetry of the pelvis displacement and step length of individuals post-stroke during early adaptation period. However, the neural system of individuals post-stroke gradually adapted to the external applied pelvis assistance force during the late adaptation period. This adaptation process is consistent with previous studies in which force perturbation was applied to the leg in individuals post-stroke [10, 11]. The neural system of individuals post-stroke may anticipate the incoming pelvis assistance force, and generate additional joint torque to counteract the external pelvis assistance force, and overtime, subjects may return a state of pelvis movement that is similar to a state during baseline.

Error-based motor learning mechanisms may be involved during motor adaptation to lateral assistance applied to the pelvis in individuals post-stroke. While lateral assistance force initially increased errors in ranges of motion of the pelvis during early adaptation period, subjects gradually adapted to the pelvis assistance force and generated additional joint torques to correct the errors induced by pelvis assistance force during late adaptation period. The neural system may adapt to this pelvis assistance force through a feed-forward control mechanism, i.e., the motor command is modified in anticipation of pelvis assistance force in upcoming steps, resulting in an aftereffect consisting of reduced range of motion of the pelvis during the post-adaptation period.

One potential mechanisms which drives this motor adaptation may be due to the energy cost [6]. While applying pelvis assistance force may improve the symmetry of the pelvis displacement and step length, the range of motion of the pelvis also increased, which may increase the energy cost and is a less efficient gait pattern.

This study has several limitations. For instance, subjects were allowed to hold on a handrail during treadmill walking for the sake of safety. The pulling force from the subject's arms could confound the results. In addition, we only tested a single magnitude of pelvis assistance force. It remains unclear whether different levels of pelvis assistance force will induce a similar response.

V. Conclusion

We investigated how stroke survivors respond to the application of pelvis assistance force toward the paretic side. Individuals post stroke may adapt to the lateral assistance force applied to the pelvis toward the paretic side and show an aftereffect consists of reduced range of motion of the pelvis after load release. In addition, while applying pelvis assistance toward the paretic side may improve the symmetry of the pelvis and step length during early adaptation period, stroke survivors may adapt to this assistance force and return to its asymmetrical pelvis movement pattern during the late adaptation period.

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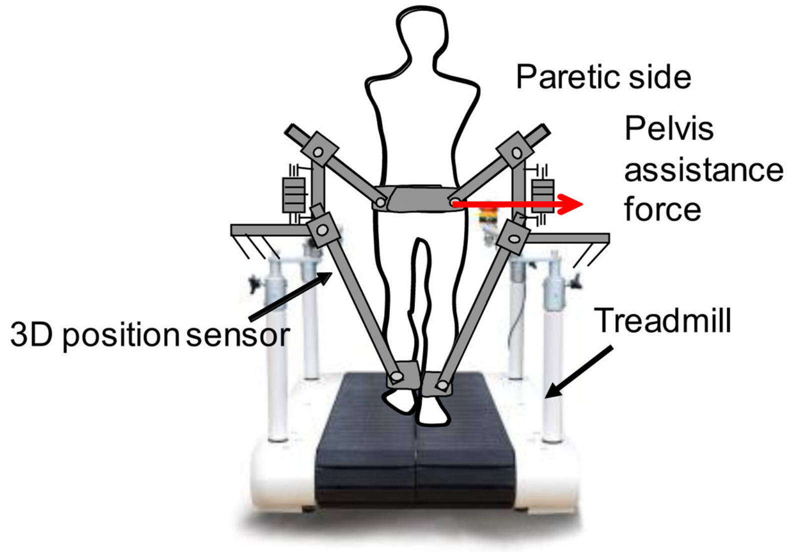
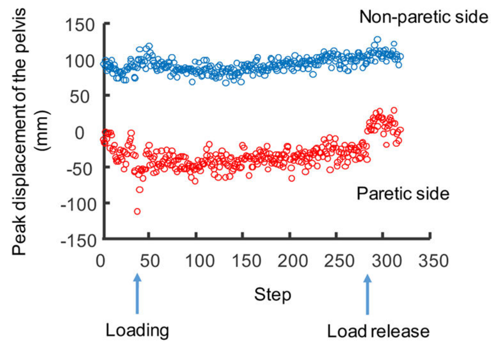


Figure 1.

Experimental setup for the pelvis assistance while a subject walked on a treadmill. The motor and cable-spool located on the right side of treadmill was used to apply a controlled assistance force to the pelvis toward the paretic side, i.e., the right side of subject in this case. A computer was used to control the movement of motors through a custom program, which was written in LabVIEW.

A.



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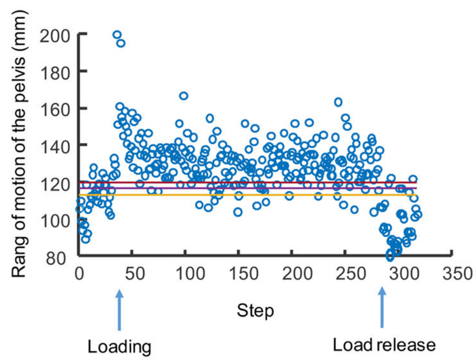

Figure 2.

Peak displacement, **A**, and range of motion, **B**, of the pelvis from one individual post stroke. A controlled pelvis assistance force was applied to the paretic side, i.e., the downward in figure A, while the subject walked on a treadmill. Three horizontal lines indicated the mean of range of motion of the pelvis during baseline and 95% confidence interval.

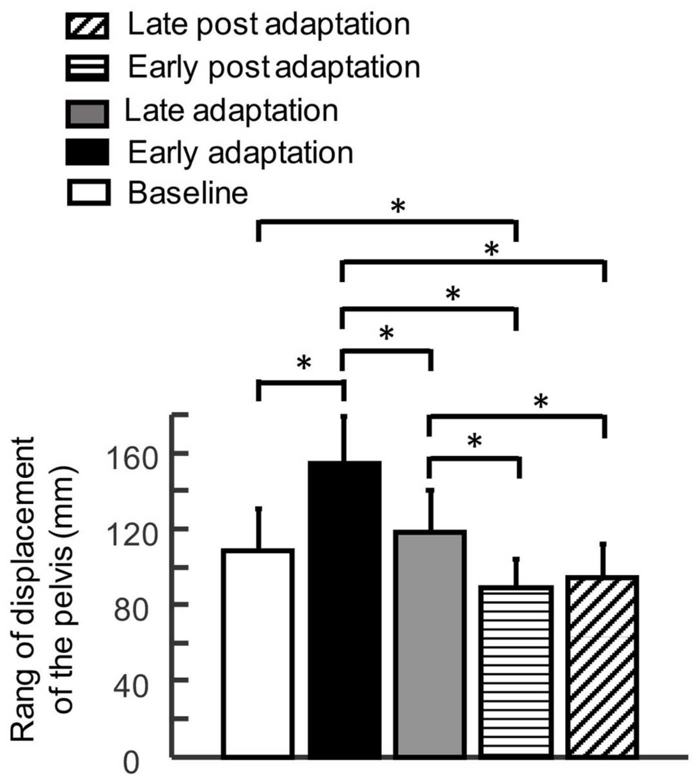

Figure 3.

Group average of the range of motion of the pelvis across 10 subjects at 5 different testing conditions. The bar and error bar indicate the mean and standard deviation of the range of motion the pelvis across subjects. Asterisk (*) indicates significant effect of loading conditions.

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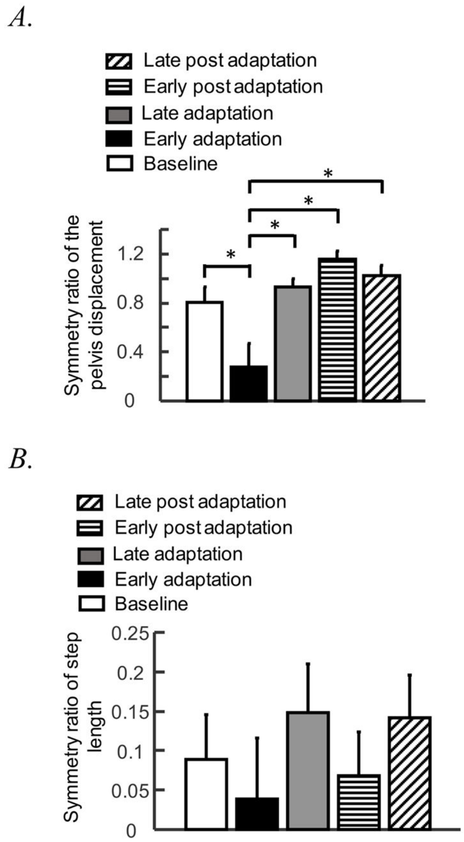

Figure 4.

Group average of the symmetry ratio of the displacement of the pelvis, **A**, and step length, **B**, across 10 subjects at 5 different testing conditions. The bar and error bar indicate the mean and standard error of the symmetry ratio of the pelvis and step length across subjects. Asterisk (*) indicates significant effect of loading conditions.

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Table 1

Participate information, including gender, age, time post stroke, affected side, and assistive devices. Abbreviations: AFO, ankle foot orthosis; AD, assistive devices.

No.	Gender	Age	Time post stroke (yrs)	Affected side	AD/Brace
1	M	46	4	Left	AFO
2	F	27	2	Left	AFO
3	F	47	10	Left	None
4	F	57	3	Left	AFO
5	M	41	9	Left	AFO
6	M	68	3	Left	AFO
7	F	62	4	Left	AFO
8	M	47	12	Left	AFO
9	M	70	21	Right	AFO
10	M	49	13	Right	AFO