Published in final edited form as:

Clin Neurophysiol. 2017 October ; 128(10): 1915–1922. doi:10.1016/j.clinph.2017.07.409.

Applying a pelvic corrective force induces forced use of the paretic leg and improves paretic leg EMG activities of individuals post-stroke during treadmill walking

Chao-Jung Hsu, PT, PhD¹, Janis Kim, PT, MS¹, Rongnian Tang, PhD¹, Elliot J. Roth, MD^{1,2}, William Z. Rymer, MD, PhD^{1,2}, and Ming Wu, PhD^{1,2}

¹Legs and walking lab, Shirley Ryan AbilityLab, Chicago, IL, USA

²Department of Physical Medicine and Rehabilitation, Northwestern University, Chicago, IL, USA

Abstract

Objective—To determine whether applying a mediolateral corrective force to the pelvis during treadmill walking would enhance muscle activity of the paretic leg and improve gait symmetry in individuals with post-stroke hemiparesis

Methods—Fifteen subjects with post-stroke hemiparesis participated in this study. A customized cable-driven robotic system based over a treadmill generated a mediolateral corrective force to the pelvis toward the paretic side during early stance phase. Three different amounts of corrective force were applied. Electromyographic (EMG) activity of the paretic leg, spatiotemporal gait parameters and pelvis lateral displacement were collected.

Results—Significant increases in integrated EMG of hip abductor, medial hamstrings, soleus, rectus femoris, vastus medialis and tibialis anterior were observed when pelvic corrective force was applied, with pelvic corrective force at 9% of body weight inducing greater muscle activity than 3% or 6% of body weight. Pelvis lateral displacement was more symmetric with pelvic corrective force at 9% of body weight.

Conclusions—Applying a mediolateral pelvic corrective force toward the paretic side may enhance muscle activity of the paretic leg and improve pelvis displacement symmetry in individuals post-stroke.

Significance—Forceful weight shift to the paretic side could potentially force additional use of the paretic leg and improve the walking pattern.

Keywords

stroke; gait; pelvic corrective force; EMG; symmetry; forced use

Correspondence to: Ming Wu, PhD, Legs and walking lab, Shirley Ryan AbilityLab, 355 East Erie Street, Chicago, IL, USA, 60611, Tel: +1-312-238-0700, Fax: +1-312-238-2208, w-ming@northwestern.edu.

Publisher's Disclaimer: This is a PDF file of an unedited manuscript that has been accepted for publication. As a service to our customers we are providing this early version of the manuscript. The manuscript will undergo copyediting, typesetting, and review of the resulting proof before it is published in its final citable form. Please note that during the production process errors may be discovered which could affect the content, and all legal disclaimers that apply to the journal pertain.

Conflict of Interest Statement

No potential conflict of interest was reported by the authors.

1. Introduction

Walking dysfunction is one of the commonly reported physical limitations after stroke (Perry et al., 1995). Individuals with post-stroke hemiparesis typically demonstrate slow gait velocity, reduced stride and step length, and both decreased period of stance and increased period of swing of the paretic leg (Balaban and Tok, 2014; Patterson et al., 2010a). As walking dysfunction can increase the risk of falls (Hausdorff et al., 2001), restrict functional mobility and negatively affect quality of life (Maclean et al., 2000; Perry et al., 1995; Schmid et al., 2007), an important goal of stroke rehabilitation is to improve symmetrical gait patterns.

The asymmetrical gait characteristics after stroke are associated with reduced weight bearing toward the paretic leg (Hendrickson et al., 2014; Olney and Richards, 1996; Tyson, 1999) and altered timing and amplitude of paretic leg muscle activation (Burridge et al., 2001; Hsu et al., 2003; Lamontagne et al., 2007). Reduced weight bearing toward the paretic leg may be a consequence of a “learned nonuse”, (Aruin et al., 2012) which can be reversed by forced use and intensive function training of the paretic leg (Schaechter, 2004; Taub et al., 2006a).

Constraint induced movement therapy has been used to induce a forced use of the paretic arm in individuals with post-stroke hemiparesis (Taub et al., 2006; Wolf et al., 2008). This training paradigm has not been transferred to paretic leg because restraint of the non-paretic leg does not allow bipedal locomotion. Recent studies indicated that using a shoe insole under the non-paretic leg may promote body weight shift towards the paretic leg (Aruin et al., 2012; Mohapatra et al., 2012). However, the positive effects of shoe insole on walking function may be limited (Aruin et al., 2012), suggesting the shoe insole may not be effective in inducing forced use of the paretic leg during locomotion.

It has been reported that giving a downward force on the pelvis could induce a prolonged stance phase of gait in infants (Yang et al., 1998) and enhance soleus muscle activity in neurological intact adults (Stephens and Yang, 1999). Potentially, applying additional loading to the paretic leg during stance phase would enhance muscle activity of the paretic leg. Therefore, the purpose of this study was to determine whether applying a mediolateral corrective force to the pelvis during treadmill walking would facilitate weight shifting toward the paretic leg, and thereby enhance muscle activity of the paretic leg and improve gait symmetry in individuals with post-stroke hemiparesis. The pelvic corrective force was applied through a customized cable-driven robotic system (Wu et al., 2011). We hypothesized that the application of pelvic corrective force toward the paretic side during early stance phase of gait would facilitate weight shifting toward the paretic leg, enhance muscle activity of the paretic leg and improve gait symmetry in individuals with post-stroke hemiparesis.

2. Methods

2.1 Subjects

A total of 15 subjects with hemiparesis due to chronic stroke (> 6 months) were recruited from the Sensory Motor Performance Program (SMPP) stroke database and the Rehabilitation Institute of Chicago (RIC) outpatient clinic. Demographic information for the subjects is shown in Table 1. The inclusion criteria were: (1) age 21–75 years, (2) unilateral, supratentorial, ischemic or hemorrhagic stroke confirmed with radiography, (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 (Folstein et al., 1975), (3) other neurological conditions, cardiorespiratory/metabolic disorders, or orthopedic conditions affecting ambulation ability, (4) botox injection within 6 months of study enrollment visit. All subjects received written and verbal information about the study procedure before giving written consent. The study was approved by the Northwestern University Institutional Review Board.

2.2 Experimental Apparatus

A customized cable-driven robotic system based over a treadmill was used to generate a corrective force to the pelvis in the mediolateral direction (Figure 1A). The robotic system consists of 2 nylon-coated stainless-steel cables, driven by 2 motors (AKM 33H, Kollmorgen, Radford, VA) through cable spools that are located on the side of the treadmill. The cables are affixed to a custom waist belt that is strapped to subjects' pelvis to provide a corrective force during treadmill walking. The pelvis and ankle positions were recorded using 4 custom designed 3-dimensional position sensors (Figure 1B). Each sensor includes a rod and 2 universal joints attached to the ends of the rod. The universal joints are connected to each side of the custom waist belt and custom ankle straps (Wu et al., 2011). Two potentiometers (P2201, Novotechnik, Southborough, MA) were used to measure rotational movements of the rod in the anteroposterior and mediolateral directions; one potentiometer (SP-2, Celesco, Chatsworth, CA) was used to measure the linear movement of the rod in the diagonal direction. The recorded ankle position signals were used to trigger the corrective force applied from initial contact to mid-stance of the paretic leg. A custom-written LabVIEW program was used to collect pelvis and ankle positions at 500 Hz as well as to command corrective force signals to the motors.

2.3 Experimental Procedures

Each subject participated in 5 test sessions on the treadmill, which included 1 session of walking at self-selected comfortable speed (baseline), 1 session of walking at maximum walking speed, and 3 sessions of walking at self-selected comfortable speed with different amounts of corrective force (3%, 6% and 9% of body weight). The order of 3 corrective force sessions was randomized across subjects. For the baseline and maximum walking speed sessions, each session required subjects to walk 30 strides. For the 3 corrective force sessions, each session was 1 minute with a 1-minute standing rest between sessions. During all sessions, subjects wore an overhead harness for safety, and no body support provided.

Subjects were also allowed to use the front handrail for safety but not encouraged to use for support.

2.4 Measurements

Surface electrodes (Delsys DE 2.1, Delsys Inc., Boston, MA) were used to record the electromyograms (EMG) during treadmill walking from 7 muscles on the paretic leg: hip abductor (gluteus medius), medial hamstrings, medial gastrocnemius, soleus, rectus femoris, vastus medialis and tibialis anterior. Electrodes were applied to lightly abraded, degreased skin over the respective muscle belly. EMG signals were amplified (gain 1,000) and band-pass filtered (20–450 Hz) in hardware. All signals were then sampled at 500 Hz via a 12-bit analog-to-digital converter (National Instruments, Austin, TX) on a PC running custom-written LabVIEW program.

2.5 Data Processing and Analysis

Custom software written in Matlab (The Mathworks, Natick, MA) was used for all data analyses. The EMG data was first low-pass filtered at 200 Hz, then high-pass filtered at 10 Hz and band-stop filtered at 55–65 Hz using a second order Butterworth filter. All EMG data were rectified and smoothed using a second order Butterworth filter with a low-pass cutoff frequency at 20 Hz. The smoothed EMG signals were segmented into step cycles from initial contact to the next initial contact of the paretic leg based on ankle positions. Because of the variability in the duration of each step cycle, the EMG data were interpolated, resampled and time-normalized to percentage of the gait cycle and then averaged across middle consecutive 20 strides. The EMG amplitude of each test session was normalized to the peak EMG amplitude of the maximum walking speed session. The integral of the EMG activity was calculated in the intervals of stance phase of the gait cycle (Perry and Burnfield, 1992).

Stance time and step length of the paretic leg were computed using ankle positions (Zeni Jr. et al., 2008). Stance time was calculated as the time from initial contact to toe-off and normalized to the whole gait cycle time (% gait cycle). Step length was defined as the anteroposterior distance between the two legs' ankle positions at initial contact (Yen et al., 2015). The symmetry ratio of stance time and step length was quantified as: $(\text{paretic} / \text{nonparetic}) * 100\%$ (Patterson et al, 2010b), where *paretic* is the variable of the paretic leg, and *nonparetic* is the variable of the non-paretic leg. A symmetry ratio of 100% represented perfect symmetry. Absolute deviation from symmetry of stance time and step length was calculated as: $|100\% - \text{symmetry ratio}|$. The absolute value was used so that asymmetry in one direction did not cancel out asymmetry in the opposite direction. For each gait cycle, the pelvis lateral displacement (relative to the reference point) was calculated by subtracting the mid-point of ankle position sensors at the initial contact from the mid-point of pelvis position sensors in the mediolateral direction. We defined the mid-point of ankle position sensors at the initial contact as a reference point for each gait cycle because there was a step-to-step variation of foot position in the mediolateral direction. The symmetry index of pelvis displacement was quantified as: $[(\text{peaknonparetic} - \text{peakparetic}) / \text{peaknonparetic}] * 100\%$ (Lauziere et al., 2014), where *peakparetic* is the peak mediolateral displacement of the paretic side, and *peaknonparetic* is the peak mediolateral displacement of the non-paretic side. For the condition that the pelvis never reached the reference point throughout the gait

cycle, the peak mediolateral displacement was defined as a negative value. For example, *peakparetic* of the baseline in Figure 2 was -2.3cm . A symmetry index of 0% indicated equal displacement of the pelvis from side to side. The absolute deviation from symmetry of stance time and step length, and symmetry index of pelvis displacement were all averaged over middle consecutive 20 strides for statistical analysis.

2.6 Statistical Analysis

Statistical analyses were conducted with commercial software (SPSS 20, IBM Co. Armonk, NY). Repeated measures analysis of variance (ANOVA) was used to compare the integrated EMG of each muscle, symmetry ratio of stance time and step length, and symmetry index of pelvis displacement between baseline and 3 corrective force sessions. A further post-hoc analysis was conducted if ANOVA yielded significant results. An alpha level of 0.05 was set for significance in all statistical tests.

3. Results

Integrated EMG

The integral of the EMG activity is presented from 14 subjects. The EMG data from 1 subject could not be reliably analyzed due to artifact. Visual inspection of individual subject data indicated that there were no systematic changes in the timing of muscle activity (activation onset and duration) and thus only changes in amplitude of muscle activity are reported. Muscle activation patterns of each muscle for all sessions from a representative subject are shown in Figure 3. There was typically higher muscle activation when a corrective force was applied to the pelvis compared with the baseline.

Average integrated EMG during the stance phase in each session is shown in Figure 4. Integrated EMG of medial hamstrings, soleus, and rectus femoris significantly increased in the corrective force sessions compared with the baseline. In addition, the corrective force at 9% of body weight induced greater muscle activity. Similarly, integrated EMG of tibialis anterior significantly increased with corrective force at 9% of body weight compared with the baseline, 3% or 6% of body weight. Integrated EMG of hip abductor also significantly increased with corrective force at 9% of body weight compared with the baseline and 3% of body weight. For medial gastrocnemius, significant differences in integrated EMG was only observed between corrective force at 6% and 9% of body weight. For vastus medialis, significant difference in integrated EMG was observed between the baseline and corrective force at 9% of body weight.

Absolute deviation from symmetry of stance time and step length

Symmetry ratio and absolute deviation from symmetry of stance time and step length for each subject is shown in Table 2 and Table 3. No significant differences in absolute deviation from symmetry of stance time (baseline= $4.9 \pm 3.7\%$, 3%= $5.2 \pm 3.6\%$, 6%= $5.1 \pm 3.5\%$, 9%= $4.9 \pm 3.2\%$; $p=0.920$) and step length (baseline= $20.7 \pm 17.5\%$, 3%= $24.5 \pm 22.9\%$, 6%= $24.3 \pm 19.1\%$, 9%= $27.4 \pm 34.1\%$; $p=0.309$) between baseline and 3 corrective force conditions were observed (Figure 5 and Figure 6), suggesting the pelvic corrective force had no significant impact on the symmetry of both stance time and step length.

Symmetry ratio of stance time and step length

Symmetry ratio of stance time and step length for each subject is shown in Table 2 and Table 3. No significant differences in symmetry ratio of stance time (baseline= $96.6 \pm 5.2\%$, 3%= $97.1 \pm 5.7\%$, 6%= $97.5 \pm 5.8\%$, 9%= $96.7 \pm 5.0\%$; $p=0.558$) and step length (baseline= $98.3 \pm 7.1\%$, 3%= $99.1 \pm 8.8\%$, 6%= $96.1 \pm 8.1\%$, 9%= $104.6 \pm 11.4\%$; $p=0.342$) between the baseline and 3 corrective force sessions were observed (Figure 5 and Figure 6), suggesting the pelvic corrective force had no significant impact on the symmetry of both stance time and step length.

Symmetry index of pelvis displacement

Pelvis displacements during a gait cycle from a representative subject are plotted in Figure 2. The trajectory of pelvis displacement stayed on the non-paretic side during the whole gait cycle at baseline, but shifted towards the paretic side when pelvis corrective force was applied. In addition, a greater corrective force induced a larger shifting to the paretic side. The symmetry index of pelvis displacement improved with corrective force compared to baseline (baseline= $101.5 \pm 29.4\%$, 3%= $98.4 \pm 27.5\%$, 6%= $88.4 \pm 41.4\%$, 9%= $86.7 \pm 36.2\%$), but did not reach statistical significance due to large variability (Figure 7). The symmetry index of pelvis displacement showed a significant improvement between 3% and 9% of body weight of corrective force ($p = 0.041$).

4. Discussion

The results of this study indicate that additional pelvic corrective force toward the paretic side during early stance phase of gait enhanced EMG activity of the paretic leg, particularly hip abductor, medial hamstrings, tibialis anterior, soleus, and rectus femoris muscles. The muscle activity improved progressively with increases in pelvic corrective force. In addition, pelvic corrective force improved the symmetry of pelvis displacement. Results from this study suggest that applying pelvis corrective force toward the paretic side during treadmill walking may be effective in inducing forced use of the paretic leg in individuals with post-stroke hemiparesis.

Sensory afferent feedback to the soleus during walking may decrease after stroke, (Mazzaro et al., 2007) which can result in decreased soleus muscle activity (Mazzaro et al., 2007; Nielsen and Sinkjaer, 2002). During the corrective force sessions, the amplitude of soleus and medial hamstrings muscle activity increased significantly due to increased weight-shift to the paretic leg. This finding is in line with previous studies (Bastiaanse et al., 2000; Dietz, 2002; Dietz, 2009; Dietz and Duysens, 2000; Dietz et al., 2002) indicating that increased leg loading stimulates extensor load receptors to facilitate leg extensor activation during walking. Specifically, sensory afferents from Golgi tendon organs in extensor muscles and cutaneous afferents from the sole of the foot may be elicited in response to increased leg loading (Bastiaanse et al., 2000; Dietz, 2002; Dietz and Duysens, 2000). The enhanced afferents may increase the amplitude of leg extensor muscle activity through spinal reflex pathways (Duysens et al., 2000).

Interestingly, increased tibialis anterior muscle activity was also observed during the stance phase. It has been reported that neurologically intact adults and individuals with spinal cord injury increased lower leg extensor and flexor muscle activities throughout the gait cycle as the initial response to novel perturbations (Gordon et al., 2009). This co-contraction strategy may help increase joint impedance and maintain postural stability (Lamontagne et al., 2000; Lamontagne et al., 2007; Rosa et al., 2014; van Soest et al., 2003). However, excessive co-contraction can potentially increase compressive loading of the joints and decrease movement flexibility (Busse et al., 2005). Future research with longer exposure to pelvis corrective force is needed to explore whether individuals with post-stroke hemiparesis would develop selective muscle responses or adopt a compensatory co-contraction strategy to ensure gait stability.

While individuals with post-stroke hemiparesis often show impairments in hip abductor muscle activity (Kirker et al., 2000), our results indicated that applying pelvic corrective force was effective in increasing muscle activity of the hip abductor during walking. Hip abductor muscle activity contributes to lower leg weight transfer and maintenance of lateral stability after stroke (Kirker et al., 2000). In particular, activation of the hip abductor is important for moving the center of mass over the stance leg when transitioning from double leg support to single leg support (Kirker et al., 2000; Pai et al., 1994). The increased hip abductor muscle activity in our study may improve the pelvis stability for single leg stance on the paretic leg (Jung et al., 2015; Mercer et al., 2009).

The smaller magnitudes of pelvic corrective force (3% and 6% of body weight) had no significant impact on the integrated EMG of medial gastrocnemius. This finding suggests that our stroke subjects might rely more on soleus for body weight support in these load conditions as soleus is more focused on providing body weight support and propulsion while the major role of the gastrocnemius is to accelerate the leg into swing (Neptune et al., 2001). However, when the magnitude of pelvis corrective force increased to 9% of body weight, the soleus might not be strong enough to counteract the increased load, medial gastrocnemius was therefore recruited to assist in body weight support.

Previous studies examining the response to loading in infants and adult rats have shown an increase in the stance phase as a result of a prolonged loading of leg extensor muscles (Fouad and Pearson, 1997; Yang et al., 1998). Therefore, we anticipated that the duration of the stance phase on the paretic leg would increase with pelvic corrective force applied and result in an improvement in stance phase symmetry. However, our subjects demonstrated no significant changes in stance phase symmetry ratio between baseline and the corrective force sessions. While the stance phase symmetry can be achieved by an increase in stance phase on the paretic leg or a decrease on the non-paretic leg, 1 subject in this study showed a substantial increase in stance phase on the paretic leg in corrective force sessions and the remaining subjects showed only minor changes on both paretic and non-paretic legs. Our result is similar to other studies indicating that temporal gait characteristics did not significantly change with additional leg loading in individuals with post-stroke hemiparesis (Duclos et al., 2014; Khanna et al., 2010). These findings suggest that in general, temporal gait characteristics might be maintained after stroke despite the addition of loading.

Although no significant changes in absolute deviation from symmetry of step length were observed between baseline and the corrective force sessions, 3 subjects still showed improvements in step length symmetry during pelvis corrective force sessions and the improvements were achieved by an increase in step length on the paretic leg. This desirable change might be attributed to the increased rectus femoris and vastus medialis muscle activities of the paretic leg during the swing phase, which could assist the knee to extend and induce the foot to further swing forward. Also, the increased pelvis lateral displacement to the paretic side during the stance phase could help trunk progression and facilitate non-paretic leg swing forward. On the other hand, 2 subjects appeared to take a shorter non-paretic step to achieve a symmetrical step. Interestingly, the asymmetry of pelvis displacement for these 2 subjects was even greater when pelvic corrective force was applied compared to baseline. This suggests that these subjects might push against the pelvic corrective force by bearing more weight through the non-paretic leg, or that the applied pelvic corrective force was not strong enough to facilitate weight shift to the paretic side. In this circumstance, trunk progression might be reduced during paretic stance, which could result in restricted forward non-paretic leg swing (Roerdink and Beek, 2011) and thus decrease non-paretic step.

This study has limitations to consider when interpreting the results. First, subjects were allowed to hold onto the front handrail during the testing. The use of handrail may influence walking patterns, including decreased lower leg loading (Zahradka, 2010), reduced proximal leg muscle activity (Stephenson et al., 2010) and increased paretic single-leg support time (Chen et al., 2005). It is possible that using the handrail could confound our results. In addition, our results indicated that higher pelvic corrective force resulted in greater gains in lower leg muscle activity. However, the small sample size and the small number of different pelvic corrective force amounts do not enable us to make inferences regarding a dose-response relationship. The study did not include functional tests, such as timed walk test. The treatment effect of pelvic corrective force on usual walking performance is still unclear. Finally, the duration of exposure to pelvic corrective force was short (i.e. 1 minute). It is unknown whether our results, such as muscle activation patterns, would be similarly altered with longer exposure to pelvic corrective force.

5. Conclusions

This study applied the concept of forced use to the paretic leg during treadmill training and reported effects of this innovative approach on lower leg muscle activity, spatial-temporal characteristics of gait and pelvis lateral displacement. The results of this study demonstrated that providing corrective force at the pelvis could improve lower leg muscle activity and pelvis displacement symmetry. Additionally, higher pelvic corrective force appeared to result in greater magnitude of improvements. This study provides initial evidence that applying pelvic corrective force may effectively induce forced use of the paretic leg and potentially overcome the learned nonuse after stroke. Knowledge gained from this study, including patterns of paretic leg muscle activity, may provide insights into the strategy that individuals with post-stroke hemiparesis use in response to external force application and help develop robotic gait training protocols for better walking function. Future research is necessary to establish a potential dose-response relationship between amount of pelvis

corrective force and magnitude of improvement in muscle activity and gait symmetry, and further identify an optimal dose for the gait training.

Acknowledgments

This study was funded by the National Institutes of Health, R01HD082216.

References

- Aruin AS, Rao N, Sharma A, Chaudhuri G. Compelled body weight shift approach in rehabilitation of individuals with chronic stroke. *Top Stroke Rehabil.* 2012; 19:556–63. [PubMed: 23192720]
- Balaban B, Tok F. Gait disturbances in patients with stroke. *PM & R.* 2014; 6:635–642. [PubMed: 24451335]
- Bastiaanse CM, Duysens J, Dietz V. Modulation of cutaneous reflexes by load receptor input during human walking. *Exp Brain Res.* 2000; 135:189–198. [PubMed: 11131503]
- Burridge JH, Wood DE, Taylor PN, McLellan DL. Indices to describe different muscle activation patterns, identified during treadmill walking, in people with spastic drop-foot. *Med Eng Phys.* 2001; 23:427–434. [PubMed: 11551819]
- Busse ME, Wiles CM, van Deursen RWM. Muscle co-activation in neurological conditions. *Phys Ther Rev.* 2005; 10:247–253. 2005.
- Chen G, Patten C, Kothari DH, Zajac FE. Gait deviations associated with post-stroke hemiparesis: improvement during treadmill walking using weight support, speed, support stiffness, and handrail hold. *Gait Posture.* 2005; 22:57–62. [PubMed: 15996593]
- Dietz V. Proprioception and locomotor disorders. *Nat Rev Neurosci.* 2002; 3:781–790. [PubMed: 12360322]
- Dietz V. Body weight supported gait training: from laboratory to clinical setting. *Brain Res Bull.* 2009; 78:I–VI. [PubMed: 19070780]
- Dietz V, Duysens J. Significance of load receptor input during locomotion: a review. *Gait Posture.* 2000; 11:102–110. [PubMed: 10899663]
- Dietz V, Müller R, Colombo G. Locomotor activity in spinal man: significance of afferent input from joint and load receptors. *Brain.* 2002; 125:2626–2634. [PubMed: 12429590]
- Duclos C, Nadeau S, Bourgeois N, Bouyer L, Richards CL. Effects of walking with loads above the ankle on gait parameters of persons with hemiparesis after stroke. *Clin Biomech.* 2014; 29:265–271.
- Duysens J, Clarac F, Cruse H. Load-Regulating Mechanisms in Gait and Posture: Comparative Aspects. *Physiol Rev.* 2000; 80:83–133. [PubMed: 10617766]
- Folstein MF, Folstein SE, McHugh PR. “Mini-mental state”. A practical method for grading the cognitive state of patients for the clinician. *J Psychiatr Res.* 1975; 12:189–198. [PubMed: 1202204]
- Fouad K, Pearson KG. Effects of extensor muscle afferents on the timing of locomotor activity during walking in adult rats. *Brain Res.* 1997; 749:320–328. [PubMed: 9138732]
- Gordon KE, Wu M, Kahn JH, Dhaher YY, Schmit BD. Ankle load modulates hip kinetics and EMG during human locomotion. *J Neurophysiol.* 2009; 101:2062–2076. [PubMed: 19193774]
- Hausdorff JM, Rios DA, Edelberg HK. Gait variability and fall risk in community-living older adults: A 1-year prospective study. *Arch Phys Med Rehabil.* 2001; 82:1050–1056. [PubMed: 11494184]
- Hendrickson J, Patterson KK, Inness EL, McIlroy WE, Mansfield A. Relationship between asymmetry of quiet standing balance control and walking post-stroke. *Gait Posture.* 2014; 39:177–181. [PubMed: 23877032]
- Hsu AL, Tang PF, Jan MH. Analysis of impairments influencing gait velocity and asymmetry of hemiplegic patients after mild to moderate stroke. *Arch Phys Med Rehabil.* 2003; 84:1185–1193. [PubMed: 12917858]

- Jung K, Kim Y, Cha Y, In TS, Hur YG, Chung Y. Effects of gait training with a cane and an augmented pressure sensor for enhancement of weight bearing over the affected lower limb in patients with stroke: a randomized controlled pilot study. *Clin Rehabil.* 2015; 29:135–142. [PubMed: 25009199]
- Khanna I, Roy A, Rodgers MM, Krebs HI, Macko RM, Forrester LW. Effects of unilateral robotic limb loading on gait characteristics in subjects with chronic stroke. *J Neuroeng Rehabil.* 2010; 7:23. [PubMed: 20492698]
- Kirker SG, Jenner JR, Simpson DS, Wing AM. Changing patterns of postural hip muscle activity during recovery from stroke. *Clin Rehabil.* 2000; 14:618–626. [PubMed: 11128737]
- Kirker SG, Simpson DS, Jenner JR, Wing AM. Stepping before standing: hip muscle function in stepping and standing balance after stroke. *J Neurol Neurosurg Psychiatry.* 2000; 68:458–464. [PubMed: 10727481]
- Lamontagne A, Richards CL, Malouin F. Coactivation during gait as an adaptive behavior after stroke. *J Electromyogr Kinesiol.* 2000; 10:407–415. [PubMed: 11102843]
- Lamontagne A, Stephenson JL, Fung J. Physiological evaluation of gait disturbances post stroke. *Clin Neurophysiol.* 2007; 118:717–729. [PubMed: 17307395]
- Lauziere S, Betschart M, Aissaoui R, Nadeau S. Understanding Spatial and Temporal Gait Asymmetries in Individuals Post Stroke. *Int J Phys Med Rehabil.* 2014; 2:201.
- Maclean N, Pound P, Wolfe C, Rudd A. Qualitative analysis of stroke patients' motivation for rehabilitation. *BMJ.* 2000; 321:1051–1054. [PubMed: 11053175]
- Mazzaro N, Nielsen JF, Grey MJ, Sinkjaer T. Decreased contribution from afferent feedback to the soleus muscle during walking in patients with spastic stroke. *J Stroke Cerebrovasc Dis.* 2007; 16:135–144. [PubMed: 17689409]
- Mercer VS, Chang SH, Williams CD, Noble K, Vance AW. Effects of an exercise program to increase hip abductor muscle strength and improve lateral stability following stroke: a single subject design. *J Geriatr Phys Ther.* 2009; 32:50–59. [PubMed: 20039583]
- Mohapatra S, Eviota AC, Ringquist KL, Muthukrishnan SR, Aruin AS. Compelled Body Weight Shift Technique to Facilitate Rehabilitation of Individuals with Acute Stroke, Compelled Body Weight Shift Technique to Facilitate Rehabilitation of Individuals with Acute Stroke. *ISRN Rehabil* 2012. 2012 pii: 328018.
- Neptune RR, Kautz SA, Zajac FE. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. *J Biomech.* 2001; 34:1387–1398. [PubMed: 11672713]
- Nielsen JB, Sinkjaer T. Afferent feedback in the control of human gait. *J Electromyogr Kinesiol.* 2002; 12:213–217. [PubMed: 12086816]
- Olney SJ, Richards C. Hemiparetic gait following stroke. Part I: Characteristics. *Gait Posture.* 1996; 4:136–148.
- Pai YC, Rogers MW, Hedman LD, Hanke TA. Alterations in weight-transfer capabilities in adults with hemiparesis. *Phys Ther.* 1994; 74:647–657. discussion 657–9. [PubMed: 8016197]
- Patterson KK, Gage WH, Brooks D, Black SE, McIlroy WE. Changes in gait symmetry and velocity after stroke: a cross-sectional study from weeks to years after stroke. *Neurorehabil Neural Repair.* 2010a; 24:783–790. [PubMed: 20841442]
- Patterson KK, Gage WH, Brooks D, Black SE, McIlroy WE. Evaluation of gait symmetry after stroke: A comparison of current methods and recommendations for standardization. *Gait Posture.* 2010b; 31:241–246. [PubMed: 19932621]
- Perry J., Burnfield, J. *Gait Analysis: Normal and Pathological Function.* Thorofare, NJ: Slack Inc; 1992.
- Perry J, Garrett M, Gronley JK, Mulroy SJ. Classification of Walking Handicap in the Stroke Population. *Stroke.* 1995; 26:982–989. [PubMed: 7762050]
- Roerdink M, Beek PJ. Understanding Inconsistent Step-Length Asymmetries Across Hemiplegic Stroke Patients Impairments and Compensatory Gait. *Neurorehabil Neural Repair.* 2011; 25:253–258. [PubMed: 21041500]
- Rosa MCN, Marques A, Demain S, Metcalf CD. Lower limb co-contraction during walking in subjects with stroke: A systematic review. *J Electromyogr Kinesiol.* 2014; 24:1–10. [PubMed: 24246405]

- Schaechter JD. Motor rehabilitation and brain plasticity after hemiparetic stroke. *Prog Neurobiol.* 2004; 73:61–72. [PubMed: 15193779]
- Schmid A, Duncan PW, Studenski S, Lai SM, Richards L, Perera S, Wu SS. Improvements in speed-based gait classifications are meaningful. *Stroke.* 2007; 38:2096–2100. [PubMed: 17510461]
- Stephens MJ, Yang JF. Loading during the stance phase of walking in humans increases the extensor EMG amplitude but does not change the duration of the step cycle. *Exp Brain Res.* 1999; 124:363–370. [PubMed: 9989442]
- Stephenson JL, Serres SJD, Lamontagne A. The effect of arm movements on the lower limb during gait after a stroke. *Gait Posture.* 2010; 31:109–115. [PubMed: 19854654]
- Taub E, Uswatte G, King DK, Morris D, Crago JE, Chatterjee A. A placebo-controlled trial of constraint-induced movement therapy for upper extremity after stroke. *Stroke.* 2006; 37:1045–1049. [PubMed: 16514097]
- Tyson SF. Trunk kinematics in hemiplegic gait and the effect of walking aids. *Clin Rehabil.* 1999; 13:295–300. [PubMed: 10460117]
- van Soest AJ, Haenen WP, Rozendaal LA. Stability of bipedal stance: the contribution of cocontraction and spindle feedback. *Biol Cybern.* 2003; 88:293–301. [PubMed: 12690488]
- Wolf SL, Winstein CJ, Miller JP, Thompson PA, Taub E, Uswatte G, Clark PC. The EXCITE Trial: Retention of Improved Upper Extremity Function Among Stroke Survivors Receiving CI Movement Therapy. *Lancet Neurol.* 2008; 7:33–40. [PubMed: 18077218]
- Wu M, Hornby TG, Landry JM, Roth H, Schmit BD. A cable-driven locomotor training system for restoration of gait in human SCI. *Gait Posture.* 2011; 33:256–260. [PubMed: 21232961]
- Yang JF, Stephens MJ, Vishram R. Transient disturbances to one limb produce coordinated, bilateral responses during infant stepping. *J Neurophysiol.* 1998; 79:2329–2337. [PubMed: 9582208]
- Yen SC, Schmit BD, Wu M. Using swing resistance and assistance to improve gait symmetry in individuals post-stroke. *Hum Mov Sci.* 2015; 42:212–224. [PubMed: 26066783]
- Zahradka, N. Effects of handrail use on healthy treadmill walking (Thesis). University of Delaware; 2010.
- Zeni JA Jr, Richards JG, Higginson JS. Two simple methods for determining gait events during treadmill and overground walking using kinematic data. *Gait Posture.* 2008; 27:710–714. [PubMed: 17723303]

Highlights

1. A mediolateral assistance force was applied to the pelvis toward the paretic side during walking.
2. Applying pelvic assistance may enhance paretic leg muscle activity and improve walking pattern.
3. Forceful weight shift to the paretic side could induce forced used of the paretic leg.

Figure 1A.

Illustration of the cable-driven robotic system and the application of pelvis assistance force.

Figure 1B.
Experimental setup and the locations of position sensors

Figure 2.

Pelvis displacement in the mediolateral direction from a representative subject. Data were retrieved from a gait cycle and normalized to % gait cycle. The dotted black line represents the beginning of swing phase, and the gray area indicates time period of applied pelvic force.

Figure 3.

A representative subject muscle activation patterns for baseline and each assistant force session. Data were averaged across 20 gait cycles and normalized to % gait cycle. The dotted black line represents the beginning of swing phase, and the gray area indicates timing of applied pelvic force.

Figure 4.
Average integrated EMG during stance phase in baseline and each assistant force session.
Error bars, ± 1 SE. * Significant difference between conditions ($p < 0.05$)

Figure 5.
Symmetry ratio of stance time under baseline and each assistant force session. Error bars, ± 1 SE.

Figure 6. Symmetry ratio of step length under baseline and each assistant force session. Error bars, ± 1 SE.

Figure 7.
Symmetry index of pelvis displacement under baseline and each assistant force session.
Error bars, ± 1 SE. * Significant difference between conditions ($p < 0.05$)

Table 1

Demographic information for the subjects

Subject	Gender	Age	Weight (kg)	Post injury (yr)	Paretic side	Assistive device	Test speed (m/s)
S1	F	65	66.7	9	R	Rollator/AFO	0.27
S2	M	40	90.7	8	L	AFO	0.62
S3	F	67	64.9	23	L	AFO	0.37
S4	F	54	68.0	5	R	AFO	0.59
S5	M	48	85.7	16	R	AFO	0.46
S6	M	46	67.1	11	L	AFO	0.74
S7	M	66	78.5	3	R	SPC/AFO	0.43
S8	F	46	92.1	9	L	None	0.53
S9	M	74	83.5	6	L	None	0.41
S10	M	67	62.6	2	L	SBQC/AFO	0.33
S11	F	46	51.3	2	L	AFO	0.47
S12	F	56	77.1	2	L	AFO	0.80
S13	F	61	75.8	3	L	SBQC/AFO	0.34
S14	F	65	67.6	1	L	None	0.37
S15	M	43	107.5	8	L	SPC/AFO	0.64

Abbreviations: AFO = ankle foot orthosis; SPC = single point cane; SBQC = small based quad cane.

Table 2

Symmetry ratio and absolute deviation from symmetry of stance time for each subject

Subject	Baseline (%)	Absolute deviation from symmetry (%)	Direction of symmetry	3% (%)	Absolute deviation from symmetry (%)	Direction of symmetry	6% (%)	Absolute deviation from symmetry (%)	Direction of symmetry (%)	9%	Absolute deviation from symmetry (%)	Direction of symmetry
S1	110.8	10.8	Paretic	109.2	9.2	Paretic	109.0	9.0	Paretic	106.9	6.9	Paretic
S2	94.1	5.9	Non-paretic	95.6	4.4	Symmetric	93.1	6.9	Non-paretic	94.5	5.5	Non-paretic
S3	94.5	5.5	Non-paretic	94.2	5.8	Non-paretic	94.0	6.0	Non-paretic	93.2	6.8	Non-paretic
S4	92.1	8.0	Non-paretic	99.2	0.8	Symmetric	99.2	0.8	Symmetric	99.2	0.8	Symmetric
S5	97.1	2.9	Symmetric	96.8	3.3	Symmetric	96.8	3.2	Symmetric	94.4	5.6	Non-paretic
S6	100.3	0.3	Symmetric	105.7	5.7	Paretic	104.4	4.4	Symmetric	104.6	4.6	Symmetric
S7	99.8	0.2	Symmetric	99.6	0.4	Symmetric	101.4	1.4	Symmetric	98.5	1.5	Symmetric
S8	97.3	2.7	Symmetric	94.5	5.5	Non-paretic	99.4	0.6	Symmetric	96.1	3.9	Symmetric
S9	100.4	0.4	Symmetric	99.5	0.5	Symmetric	100.8	0.8	Symmetric	101.2	1.2	Symmetric
S10	94.2	5.8	Non-paretic	94.0	6.0	Non-paretic	93.1	6.9	Non-paretic	94.8	5.2	Non-paretic
S11	90.4	9.6	Non-paretic	88.1	11.9	Non-paretic	88.2	11.8	Non-paretic	88.5	11.5	Non-paretic
S12	92.0	8.0	Non-paretic	92.0	8.0	Non-paretic	89.9	10.1	Non-paretic	89.1	10.9	Non-paretic
S13	90.6	9.4	Non-paretic	89.5	10.5	Non-paretic	92.8	7.2	Non-paretic	95.2	4.8	Symmetric
S14	100.8	0.8	Symmetric	102.5	2.5	Symmetric	104.9	4.9	Symmetric	97.1	2.9	Symmetric
S15	96.0	4.0	Symmetric	96.7	3.3	Symmetric	96.8	3.2	Symmetric	98.4	1.6	Symmetric

Paretic: longer paretic than non-paretic duration of stance, symmetry ratio $\gg$ 100%Non-paretic: longer non-paretic than paretic duration of stance, symmetry ratio $\ll$ 100%Symmetric: mostly symmetric duration of stance, symmetry ratio $> 95\%$ or $< 105\%$

Table 3
Symmetry ratio and absolute deviation from symmetry of step length for each subject

Subject	Baseline (%)	Absolute deviation from symmetry (%)	Direction of symmetry	3% (%)	Absolute deviation from symmetry (%)	Direction of symmetry	6% (%)	Absolute deviation from symmetry (%)	Direction of symmetry	9%	Absolute deviation from symmetry (%)	Direction of symmetry
S1	163.4	63.4	Paretic	180.4	80.4	Paretic	160.6	60.6	Paretic	226.9	126.9	Paretic
S2	107.7	7.7	Paretic	108.6	8.6	Paretic	104.3	4.3	Symmetric	112.4	12.4	Paretic
S3	86.4	13.6	Non-paretic	79.2	20.8	Non-paretic	77.0	23.0	Non-paretic	75.7	24.3	Non-paretic
S4	106.8	6.8	Paretic	100.0	0.0	Symmetric	91.6	8.4	Non-paretic	103.8	3.8	Symmetric
S5	82.7	17.3	Non-paretic	81.1	19.0	Non-paretic	82.5	17.5	Non-paretic	86.2	13.9	Non-paretic
S6	98.7	1.3	Symmetric	101.1	1.1	Symmetric	98.9	1.1	Symmetric	93.6	6.4	Non-paretic
S7	79.0	21.0	Non-paretic	63.3	36.7	Non-paretic	81.6	18.4	Non-paretic	68.2	31.8	Non-paretic
S8	149.9	49.9	Paretic	165.2	65.2	Paretic	166.7	66.7	Paretic	180.6	80.6	Paretic
S9	61.8	38.2	Non-paretic	58.4	41.6	Non-paretic	57.1	42.9	Non-paretic	57.6	42.4	Non-paretic
S10	97.6	2.4	Symmetric	87.1	12.9	Non-paretic	92.5	7.5	Non-paretic	102.6	2.6	Symmetric
S11	89.9	10.1	Non-paretic	93.8	6.2	Non-paretic	80.8	19.2	Non-paretic	85.8	14.2	Non-paretic
S12	80.6	19.4	Non-paretic	79.0	21.1	Non-paretic	80.4	19.6	Non-paretic	89.0	11.0	Non-paretic
S13	79.7	20.3	Non-paretic	86.3	13.7	Non-paretic	73.9	26.1	Non-paretic	98.8	1.2	Symmetric
S14	75.7	24.3	Non-paretic	82.2	17.9	Non-paretic	71.7	28.4	Non-paretic	74.1	25.9	Non-paretic
S15	114.7	14.7	Paretic	121.6	21.6	Paretic	121.2	21.2	Paretic	114.1	14.1	Paretic

Paretic: longer paretic than non-paretic steps, symmetry ratio $\gg$ 100%
Non-paretic: longer non-paretic than paretic steps, symmetry ratio $\ll$ 100%
Symmetric: mostly symmetric steps, symmetry ratio $> 95\%$ or $<105\%$