1-1-2015

Perception of Lower Extremity Loads in Stroke Survivors

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Accepted version. *Clinical Neurophysiology*, Vol 126, No. 2 (February 2015): 372-381. DOI.

Perception of Lower Extremity Loads in Stroke Survivors

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Abstract

Objective: This study aimed to improve our understanding of static and dynamic lower extremity sensory perception and the impact of sensory impairments on the control of walking in stroke survivors.
Methods: Using a custom, real-time unloading system, we tested load perception at heel strike, mid stance and push off in 10 stroke survivors and compared their performance to 10 age-matched and 5 young adult control subjects. Dynamic load perception was based on a judgment of which leg was bearing more load, which was altered on a step by step basis. We also examined lower extremity static load perception, coordination, proprioception, balance, and gait symmetry.

Clinical Neurophysiology, Vol 126, No. 2 (February 2015): pg. 372-381. DOI.

Results: The stroke survivors performed significantly worse than the control subjects in dynamic load perception, coordination, proprioception, balance and gait symmetry. Gait symmetry correlated with static and dynamic load perception measures but not with age, proprioception, coordination, and balance.
Conclusions: Sensory deficits related to load detection in the impaired limb could result in an increased uncertainty of limb load and a gait strategy in which stroke survivors minimize loading of the impaired limb.
Significance: This new method of measuring lower extremity dynamic load perception provides a framework for understanding gait-related sensory impairments in stroke survivors.
Keywords: Stroke; Load perception; Gait symmetry; Sensory deficits

1. Introduction

The loss of load perception in the impaired leg likely impacts control of walking in stroke survivors (Bohannon, 2003). While the gait impairments experienced by stroke survivors could result directly from damage to motor areas of the brain (Collen et al., 1990 and Enzinger et al., 2008), the lack of proper inputs from the environment (sensory information) clearly has an impact on the body's ability to control movement. In order to walk without losing balance, the motor control system needs to receive accurate sensory information from the limbs. Similarly, we would expect that a lack of accurate sensory information could lead to imbalance and asymmetries in gait. Both sensory impairments (Carey, 1995, Carey et al., 1996, Kim and Choi-Kwon, 1996 and Tyson et al., 2008) and gait asymmetry (Wall and Ashburn, 1979, Dettmann et al., 1987, Morita et al., 1995 and Titianova and Tarkka, 1995) have been well documented in stroke survivors but there has not been an attempt to study the relationship between the two.

Sensory dysfunction is estimated to be present in more than half of stroke survivors (Carey, 1995, Carey et al., 1996 and Tyson et al., 2008). This sensory dysfunction has been documented primarily as a loss of proprioception, with most proprioceptive tests in the post-stroke population involving limb position sense and the sensation of movement (Bohannon, 2003 and Sullivan and Hedman, 2008). About 36–54% of stroke survivors demonstrate some loss of limb position sense (Shah, 1978, Smith et al., 1983 and Carey, 1993). Other sensory impairments after stroke include deficits in tactile discrimination (Kim and Choi-Kwon, 1996), and impairments in vision, hearing, smell and taste (Bohannon, 2003). While these measurements of sensory loss are important, quantification of perception of limb loading has been extremely limited, despite the possible effects it could have on the control of standing or walking, as the significant role of limb loading in the regulation of gait has been previously illustrated in animal research (Duysens et al., 2000).

The effect of load perception on the control of walking can be appreciated by its likely relationship to gait asymmetry in stroke survivors. Gait asymmetry in stroke survivors has been reported in the temporal, spatial and kinetic domains. The step-length ratio between the paretic and non-paretic limb is approximately 1.13 (Dettmann et al., 1987). The paretic limb also has a shorter stance time, prolonged swing time and decreased ground reaction forces relative to the non-paretic limb (Wall and Ashburn, 1979, Morita et al., 1995, Titianova and Tarkka, 1995 and Bohannon, 2003). An asymmetrical gait is poor for balance and energetically inefficient (Winter, 1978, Lowery, 1980, Olney et al., 1986, Iida and Yamamuro, 1987 and Olney and Richards, 1996), making it an important target for rehabilitation training. Researchers have proposed various factors as the cause for post-stroke gait asymmetries, including spasticity (Dietz and Berger, 1984; Bohannon et al., 1987; Hsu et al., 2003), muscle weakness (Tang and Rymer, 1981, Bourbonnais and Vanden Noven, 1989 and Olney et al., 1991), inappropriate co-contraction (Knutsson and Richards, 1979 and Conrad et al., 1985) and reduced voluntary drive from the central nervous system (McComas et al., 1973). However, these factors do not fully explain the asymmetries observed in post-stroke gait (Hsu et al., 2003). We believe that limb load perception also has an important role in maintaining gait symmetry, and has been left out of previous studies of gait symmetry.

This study is the first to specifically examine load perception during walking in stroke survivors. We examined both static load perception and dynamic load perception (i.e. during walking). We recruited 10 stroke survivors, 10 age-matched neurologically-intact controls and 5 young adult controls in order to test the effects of stroke and age on lower extremity load perception. We used a motorized body weight support system to manipulate the weight bore by each leg during walking to test dynamic load perception. Further, we examined lower extremity coordination, proprioception, force detection, balance, static standing weight distribution and loading symmetry during gait. In the stroke survivors, we also examined their knee strength, and administered the sensory and motor subsections of the Fugl-Meyer Test (Fugl-Meyer et al., 1975) for the lower extremities and the Modified Ashworth scale (Bohannon and Smith, 1987) to measure spasticity. We hypothesized that sensory deficits in stroke survivors would affect load perception and the severity of this impairment would correlate with gait symmetry.

2. Methods

2.1. Participants

Ten participants with chronic stroke were recruited to participate in this study (characteristics shown in Table 1). The mean age of the stroke participants was 57.27 years (standard deviation (S.D.) = 7.62 years). Two of the 10 stroke participants were female. All 10 participants had a cerebrovascular accident (CVA) more than 6 months before the test date. Due to the treadmill walking requirement of the test, we only recruited participants who were able to take steps independently. Participants were medically stable, with no concurrent medical illnesses. Participants were excluded for unhealed decubiti, bladder or other infection, severe contracture or osteoporosis, heterotopic ossification, cardiac arrhythmia or inability to give informed consent.

Table 1. Participant characteristics.

ID	**Stroke**					**Age-matched**	
	Age (years)	**Gender**	**Years post stroke**	**Diagnosis**	**Paretic side**	**Age (years)**	**Gender**
1	57.48	M	8.65	Left CVA	Right	59.56	M
2	66.61	M	17.26	Right hippocampus CVA	Left	66.67	M
3	53.75	M	1.40	Left thalamic CVA	Right	56.88	F
4	51.08	M	0.92	Right CVA	Left	50.72	M
5	66.74	M	1.21	Left CVA	Right	68.13	M
6	51.39	F	2.40	Left basal ganglia CVA	Right	48.74	F
7	47.23	M	1.63	Left frontal parietal CVA	Right	45.34	M
8	62.58	F	2.57	Left CVA	Right	62.40	M
9	65.97	M	1.95	Right CVA	Left	65.50	M

ID	Stroke					Age-matched	
	Age (years)	**Gender**	**Years post stroke**	**Diagnosis**	**Paretic side**	**Age (years)**	**Gender**
10	49.85	M	1.02	Right CVA	Left	50.27	M

We also recruited 10 age-matched controls with no history of neurological disorder. The mean age of the age-matched controls was 57.42 years (S.D. = 8.25 years). Each control participant recruited in the study was within 3 years in age of one of the participants in the stroke group. The age of the stroke group and the age-matched control group was not significantly different (p = 0.96). There were 2 females in the age-matched control group. A third group of five young controls, mean age 25.88 years old (S.D. = 3.6795 years), were recruited into the study. In this group, all participants were female. Informed consent was obtained in writing from all participants before enrollment and participation in the study. All study procedures were conducted in accordance with the Declaration of Helsinki and with approval from the Northwestern University. All tests were conducted in research laboratories at the Rehabilitation Institute of Chicago (RIC).

2.2. Clinical measures

Clinical measures of sensory and motor function, and spasticity were measured in the stroke participants. The results are presented in Table 2. Sensory and motor function was measured using the Fugl-Meyer sensory and motor subtests for the lower extremities (Fugl-Meyer et al., 1975). Spasticity was assessed using the Modified Ashworth scale (Bohannon and Smith, 1987) on the ankle plantarflexors, knee flexors and extensors, and hip flexors, extensors and adductors.

Table 2. Clinical measures for the stroke participants, FMS – Fugl-Meyer sensory score for the lower extremities; FMM – Fugl-Meyer motor score for the lower extremities; MAS – modified Ashworth Score; NET – Normalized Extension Torque calculated based on body weight.

ID	Fugl-Meyer		MAS-quadriceps		MAS-hamstrings		MAS-plantarflexors		Knee NET (Nm/kg)	
	FMS	**FMM**	**Paretic**	**Non-paretic**	**Paretic**	**Non-paretic**	**Paretic**	**Non-paretic**	**Paretic**	**Non-paretic**
1	11	29	2	1	1	0	2	0	0.86	1.40
2	2	20	0	0	0	0	0	0	0.36	0.84

ID	Fugl-Meyer		MAS-quadriceps		MAS-hamstrings		MAS-plantarflexors		Knee NET (Nm/kg)	
	FMS	FMM	Paretic	Non-paretic	Paretic	Non-paretic	Paretic	Non-paretic	Paretic	Non-paretic
3	11	22	0	1+	0	1	0	0	0.58	1.20
4	6	19	0	0	0	0	0	0	0.89	1.56
5	11	25	1+	0	1+	0	2	0	0.65	1.18
6	3	27	0	0	0	0	0	0	1.05	1.42
7	12	28	0	0	0	0	0	0	1.04	2.85
8	12	15	1+	0	0	0	0	0	0.44	1.43
9	11	22	3	2	1	1	4	0	0.79	1.83
10	8	28	0	0	0	0	0	0	2.03	2.44

Clinical observations suggest that a common barrier to successful walking is buckling at the knee, which affects the ability to support body weight during stance. Specifically, sufficient knee extension strength is needed to prevent knee buckling. Therefore, we assessed the isometric maximum voluntary contraction (MVC) torque for knee extension in the stroke group using the protocol described in previous studies (Hornby et al., 2009). Neckel and colleagues observed that the sagittal ankle and hip torques do not change during walking in stroke survivors; however, the sagittal knee torques differ significantly (Neckel et al., 2008). In order to examine strength in relation to walking, we compared the isometric knee torques to the maximum knee torque during normal walking as reported by Neckel et al. (2008). In normal walking, the highest knee extension torque occurs during early stance, and peaks at 0.3 Nm/kg. We normalized the maximum knee extension torque by each participant's body weight and all the stroke participants had a knee extension MVC that was higher than the knee extension torque needed during a gait cycle.

2.3. Experimental setup

An eight-camera motion capture system (Motion Analysis Corp, Santa Rosa, CA) was used to record three-dimensional movement of retroreflective markers placed on boney landmarks on both legs (Lewek et al., 2009). The 1 inch retroreflective markers were placed on the posterior sacrum, bilateral anterior–superior iliac spine, medial and lateral femoral condyles, medial and lateral malleoli, and posterior heel of the shoe and dorsally over the second and fifth metatarsal heads to identify the boney landmarks. Three markers were rigidly affixed on thermoplastic casts that were secured on the thighs and the shanks.

Body weight support was provided through a custom motorized body weight support system. This is a modified design based on the body weight support (BWS) system of Grabowski et al. (2005), where we replaced elastic bands with a motor/spindle. These components are shown in Fig. 1. The system includes an overhead actuator consisting of a DC servomotor (Kollmorgen, Northampton, MA) coupled to a cable–pulley system. The BWS system provides a controlled vertical upward force (up to 3500 N) to the participant through a harness. The motor and pulley system are mounted on a trolley that allows movements in the horizontal plane, thus allowing sideways and forward/backward movement of the participant and eliminating any propulsion or corrective forces. The motorized system also allows for real time control of the amount of BWS through a computer program with a clock cycle of 30 Hz. The BWS system is mounted over an instrumented split-belt treadmill (Bertec, Columbus, OH).

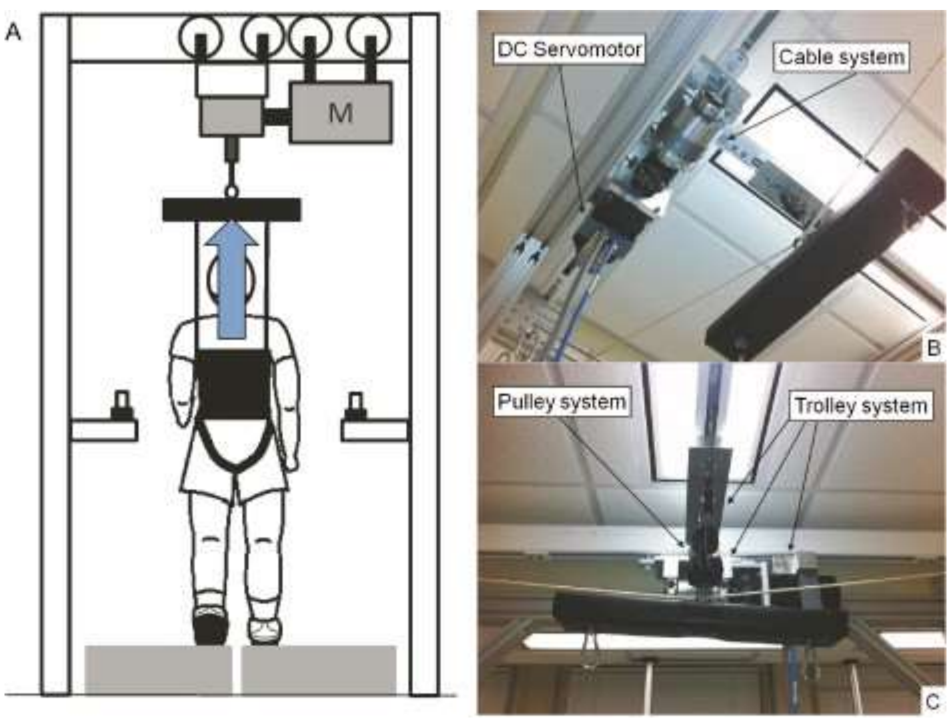

Fig. 1. Custom body weight support system for controlling load during treadmill stepping. A controlled vertical force was applied by the motor system using a cable that passes through a trolley. This system is a modified design based on the BWS system of Grabowski et al. (2005).

The instrumented split-belt treadmill, equipped with independent six-dimensional force plates beneath each belt, was used to measure static and dynamic (during stepping) loading. The three-dimensional position of the markers and force plate data were collected using Cortex software (Motion Analysis, Santa Rosa, CA). Customized LabVIEW software was written to control the motor that provided the body weight support, with weight support altered according to the gait cycle of the participant. The gait cycle was identified in real time based on the center of pressure (COP) measured by the force plates (Roerdink et al., 2008). Heel strike was detected by monitoring large changes in the medial–lateral axis of COP in real time. Timing of the body weight support changes were calculated based on percentage of the period of a baseline gait cycle.

2.4. Dynamic load perception

Prior to testing dynamic load perception, participants walked on the treadmill with symmetrical BWS. The treadmill was set at a self-selected comfortable walking speed for the participant. Kinetics and kinematics were recorded using the force plates and motion capture system.

Dynamic load perception was tested in the lower extremities, using a newly developed technique. Note that changes in BWS have been used as a method to study gait characteristics (Stephens and Yang, 1999). In contrast, in the present study, we combined the use of asymmetrical BWS, where the amount of BWS changed depending on the foot (right or left) on the treadmill, with a parameter estimation algorithm to test dynamic load perception. The iterative algorithm known as parameter estimation by sequential testing (PEST) (Taylor and Creelman, 1967) was adapted for this test, along with manipulation of the difference between the BWS provided when each leg was on the ground. PEST is often used in the estimation of psychophysical thresholds and this algorithm has been used previously in an upper limb position perception task (Ostry et al., 2010). Each PEST trial began with a suprathreshold difference in loading in each leg through the manipulation of the BWS. Then, based on the participant's response, the algorithm progressively decreased the loading difference until the threshold of detection.

Based on pilot testing, we used 30% body weight (BW) as the initial load difference between the two legs (20% BWS on one leg and 50% BWS on the other). Every few steps, the participant was asked which leg was bearing more weight. After each answer, we adjusted the BWS on the leg that began with 50% BWS, such that the amount and direction of change in the BWS reduced the perceived difference in load between the two legs. The initial step size was set at 7% BW. Each time the subject reported a change in the leg that bore more weight, the step size was reduced by half. The PEST trial terminated when the upcoming step size fell below 0.5% BW. An example of the PEST algorithm is provided in Table 3.

Table 3. PEST algorithm example. BWS setting for each leg, the subject's answer to the question: "Which leg is bearing more weight?" and the upcoming change step size for a sample trial. The BWS for the left leg was adjusted according to the subject's answer such that the perceived load difference was reduced. The change step size was reduced in half each time the subject changed their answer.

	1	**2**	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**10**
Right BWS	20%	20%	20%	20%	20%	20%	20%	20%	20%	20%
Left BWS	50%	43%	36%	29%	32.5%	30.75%	29%	27.25%	28.125%	29%
Subject answer	R	R	R	L	R	R	R	L	L	R
Change step size	7%	7%	7%	3.5%	1.75%	1.75%	1.75%	0.875%	0.875%	0.4375%

Three conditions (heel strike, mid stance, push off) were tested. In each condition, the change of BWS between steps was triggered at different times to ensure that the BWS was constant during the test phase of the gait cycle (see Fig. 2). In the heel strike condition, the participants were asked to focus on the time when the foot strikes the treadmill surface. They were asked to determine which foot had a harder impact with the treadmill. In the mid stance condition, the subjects were asked to focus on which leg was bearing more weight when that leg was fully planted on the ground. In the push off condition, the subjects were asked to determine which leg they felt had to push harder to lift off. The three conditions were tested in random order. For all conditions, participants were asked to maintain a normal gait at their self-selected comfortable speed. Note that heel strike and push off conditions occurred during double leg stance. Although the amount of weight bore by each foot during double stance was not experimentally controlled, assuming that the participants were maintaining a normal gait, the differences in the amount of weight carried by each foot during heel strike or push off should reflect the difference in the BWS levels. The actual ground reaction forces for each foot was measured and used in load perception analysis.

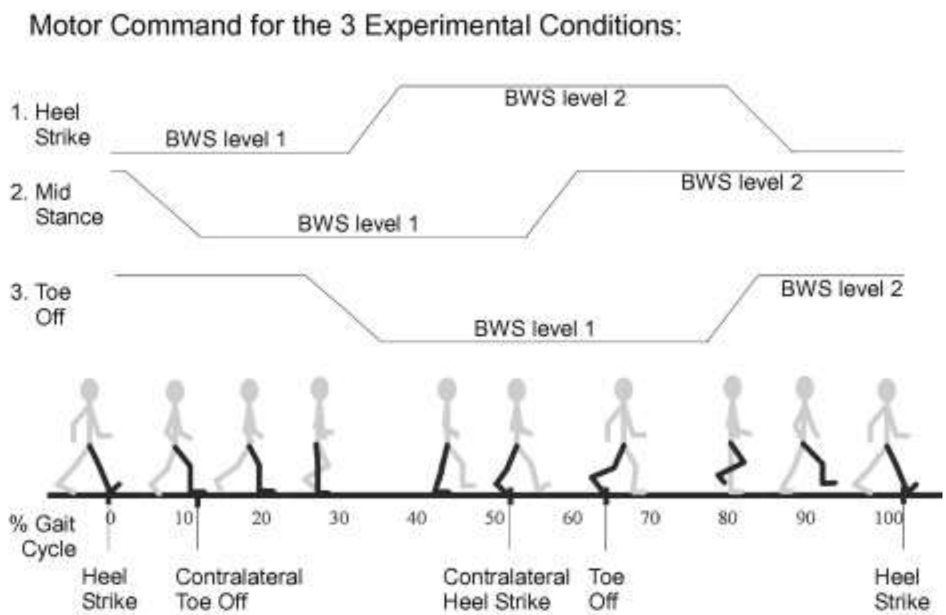

Fig. 2. Experimental design for the dynamic load perception task. The lines indicate the changes between two BWS levels for the three experimental conditions. The stick figures represent the different phases of the gait cycle.

For each condition, we conducted 4 PEST trials. In each trial, one leg would start bearing the higher load (BWS 20%), and the other leg would bear the smaller load (BWS 50%). During the PEST trial, the BWS for one leg would be changed to reduce the perceived loading difference while the BWS for the other leg would be kept constant. In two of the four trials, we tested the right leg with the higher initial load, where in one trial the BWS for the left leg was held constant, and in the other trial with the BWS of the right leg was held constant. Similarly, in the other two trials we started with the left leg on the higher initial load. The order of the trials was assigned randomly. In each trial, we computed the response accuracy as the percentage of correct responses based on comparing the verbal response to the actual ground reaction force recorded from the force plates. In the event the participant's response would result in unsafe operation of the BWS system (e.g. dropping below zero), the trial was prematurely terminated.

The PEST algorithm allowed us to develop a model of the participant's load perception by calculating a decision curve (see Fig. 3). Based on the subject's answers and the recorded forces from the force plates beneath each foot, we were able to model the decision curve for the subject (probability of answering "Right foot is bearing more weight") using a regression fit with a sigmoidal curve. Load perception was then quantified by computing the perception error, which was defined as the difference in load between the two limbs when the decision curve was at chance (probability = 0.5).

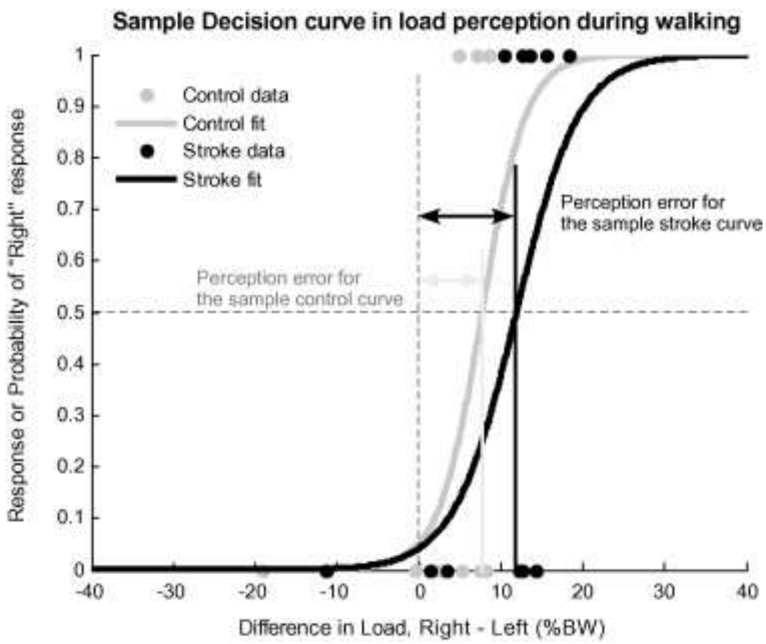

Fig. 3. Example PEST decision curve. A sample decision curve of a trial from a stroke subject and a sample decision curve for a control subject. The perception error was defined as the absolute load difference between the two legs at the decision curve = 0.5.

2.5. Sensory and motor outcome measures

We measured static load perception in the lower extremities while standing using a modified single-leg force proprioception test (Murtaugh and Costigan, 2003). Participants were asked to stand on the stationary treadmill, with one foot on each force plate (underneath each treadmill belt), shifting their weight from one side to the other to match target distributions of body weight between their two feet. Visual feedback of the load distribution in the form of a moving bar and a target was given to the participants, and they were asked to place the moving bar onto the target line. The participants were asked to hold the force for 5 s and remember how the load distribution felt. Participants were then asked to step in place on the force plates, such that each foot was lifted off the force plates at least once. Then, participants were asked to reposition their feet and reproduce the same load distribution while the visual feedback was turned off. Static load perception was tested for 5 load distributions, evenly spread across the range of loads that the participant was able to put on the paretic or test leg while maintaining balance. In the controls, the range of load on the test leg was 10–90% BW. In the stroke survivors, the range of load on the paretic leg was 10–80%, depending on the participant's ability to maintain balance. The order of load distribution was randomly assigned. The outcome measure was the mean absolute difference between the target distribution (with visual feedback) and the matching distribution (without visual feedback) as a percent of body weight.

To measure force detection threshold, we used Semmes–Weinstein monofilaments to find the perception threshold force on the sole of the feet (both the paretic and non-paretic limbs). This test allowed for measurement of small forces near the perception threshold. With the participant's eyes closed, we tested three locations on the sole of the feet that typically load-bear during walking: the plantar side of the 1st distal phalanx, the lateral arch, and the heel. A Touch-Test 20 Piece Full Kit (North Coast Medical, Morgan Hill, CA) was used, starting with the thinnest filament (size 1.65 mm) and gradually increasing size. At each test site, we pressed the filaments at 90° against the skin until the filament bowed. Each filament was applied to the test site three times to elicit a response. Participants were instructed to say "yes" every time they felt their skin touched by the monofilament. The first (smallest) filament to elicit two correct responses (out of three) was noted and its calibrated force was recorded as the force detection threshold.

In order to examine kinesthesia and proprioception in the lower limbs, we adapted the conventional "finger-to-nose" test for use with the legs. The participants were seated on a chair and instructed to plant one foot on the floor, close to the body midline, and 'reach' with the other foot. The participants were asked to reach out as far as possible with their big toe, without moving their trunk, and then reach into touch the big toe of the foot planted on the floor, as fast and as accurate as possible. Participants were asked to perform the movement sequence 5 times with eyes open, and repeat the movement with eyes closed. The movement sequence was performed by the dominant limb in the control groups and repeated with both feet for the stroke group. The movements were captured using the motion capture system with a 1-cm reflective marker placed on the tip of each big toe. The average movement time and average minimum distance between the two toes were calculated for each condition (eyes opened and eyes closed). The trials completed with eyes open were associated with coordination of lower extremity, whereas the difference between the eyes opened and eyes closed trials was associated with proprioception error.

We assessed standing balance in the participants using the Romberg test (Khasnis and Gokula, 2003). Participants were asked to stand upright on a force plate as still as possible, first with eyes open for 60 s, followed by another 60 s with eyes closed. Ground reaction forces were measured and the center of pressure map was calculated. Changes in the size of the center of pressure (COP) map from the eyes open to the eyes closed condition was used as a measure of balance. Size of the COP map was calculated as the maximum distance from the center of the COP map.

2.6. Static and gait symmetry

We assessed load symmetry during standing and walking. Participants were instructed to stand as still as possible for 1 min, with one foot on each force plate without external support. We calculated the difference in percentage of body weight supported by the test leg (paretic side in the stroke participants) and the other leg. During walking, we have to consider gait timing in addition to loading differences when examining load symmetry. We calculated the proportion of body weight that was supported by each leg, averaged across the entire gait cycle. This allowed us to capture both the timing and force information in a measure of dynamic load symmetry.

2.7. Relationship between load symmetry and study parameters

We examined the relationship between load symmetry (during standing and walking) and the sensory and motor parameters we measured in this study. Linear regressions were calculated between the two load symmetry measures and the three load perception parameters (static load perception, dynamic load response accuracy and dynamic load perception error). For the dynamic load perception measures, we focused on the heel strike condition due to the overall best performance in this condition. Correlations between load symmetry and force detection, proprioception, and balance were also calculated.

2.8. Statistical analysis

Data analysis to calculate the measures for each test was carried out in MATLAB. Statistical analysis was performed in the statistical software STATVIEW. One-way ANOVAs were used to examine the differences between the stroke survivors and the two control groups. Two-way ANOVAs were used to examine the difference between the three subject groups and the three experimental conditions in the dynamic load perception test. A post hoc test, Fisher's Protected Least Significant Difference (PLSD) test, was used to conduct pairwise comparisons between subject groups and experimental conditions. Pearson's correlation coefficients from the linear regressions were used to examine the correlations between load symmetry and load perception. Other sensory measures and measures of motor deficits were also examined in relation to load symmetry. In all statistical tests, the significance level was set at $\alpha = 0.05$.

3. Results

3.1. Dynamic load perception

During the dynamic load perception test, participants walked at a self-selected comfortable walking speed. The speed of the treadmill for the stroke group ranged from 0.1 m/s to 0.75 m/s, with a mean of 0.36 m/s (S.D. = 0.16 m/s). The treadmill speed for the age-matched control group ranged from 0.4 m/s to 0.9 m/s, with a mean of 0.62 m/s (S.D. = 0.14 m/s). The treadmill speed for the young control group ranged from 0.6 m/s to 0.87 m/s, with a mean of 0.71 m/s (S.D. = 0.12 m/s). We confirmed that the difference between the ground reaction forces recorded for each leg were not significantly different from the difference between the amounts of BWS provided when each leg was on the ground ($p = 0.89$), indicating that our BWS perturbations and measurements were consistent.

One participant (Participant 2) in the stroke group was unable to judge the loading on his legs and the test had to be prematurely aborted. Although the participant was able to walk on the treadmill with BWS, whenever BWS was applied, the participant claimed that his paretic leg was not touching the treadmill and not bearing any weight. After multiple attempts of adjusting to different amounts of body weight support, the participant was still adamant that his leg was not touching the treadmill whenever the BWS was turned on. For this reason, we aborted the test and recorded the participant being unable to perform the task.

For each condition, we calculated the mean response accuracy (% of correct response) for each group. For stroke participant 2, due to his inability to sense dynamic forces, we assumed a response accuracy of 0%. The results are graphically presented in Fig. 4a. A 2-way ANOVA with group (stroke, age-matched, young) and experimental condition (heel strike, mid stance, push off) as independent factors showed that both independent factors were significant (group: $p < 0.0001$, experimental condition: $p = 0.011$) but the interaction between factors were not significant. Response accuracy for all three groups were significantly lower in the push off condition compared to the other two conditions (heel strike: $p = 0.013$, mid stance: $p = 0.0075$), whereas the heel strike and mid stance conditions were not significantly different ($p = 0.8359$). The stroke group had the lowest response accuracy ($p < 0.0001$) compared to the other two control groups. The two control groups did not significantly differ in their response accuracy ($p = 0.51$).

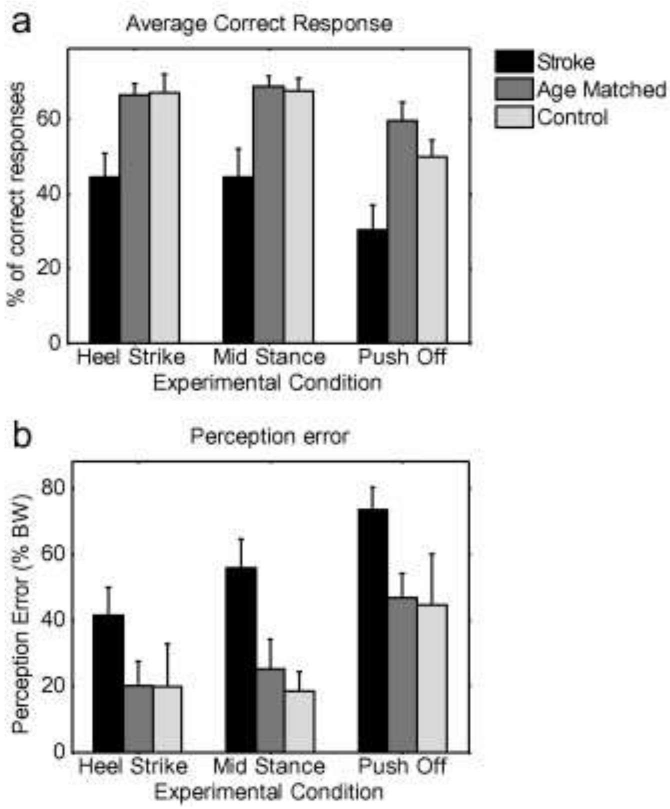

Fig. 4. Average results in the dynamic load perception task. a. Average response accuracy, averaged by subject group and experimental condition. b. Average perception error for subject group and experimental condition. The error bars indicate standard error.

Decision curves were calculated for each PEST trial; sample curves are shown in Fig. 3. Perception error was defined as the difference in load between the two limbs when the perception of either leg bearing more weight was at chance (decision curve probability = 0.5). Perception error provided another measure of the participants' dynamic load perception. For stroke participant 2, due to his inability to sense dynamic forces, we assumed a perception error of 100%BW. Group averages are presented graphically in Fig. 4b. A 2-way ANOVA with group (stroke, age-matched, young) and experimental condition (heel strike, mid stance, push off) as independent factors showed that both independent factors were significant (group: $p = 0.0001$, experimental condition: $p = 0.0014$) but the interaction between factors was not significant. Perception error for all three groups was significantly higher in the push off condition compared to the other two conditions (heel strike: $p = 0.002$, mid stance: $p = 0.0053$), whereas the heel strike and mid stance conditions were not significantly different ($p = 0.31$). The stroke group had the highest perception error ($p < 0.001$) compared to the other two control groups. The two control groups did not significantly differ in their perception error ($p = 0.70$).

The response accuracy (Fig. 4a) captured the percentage of correct response for each trial. It was noted that the percentage of correct response in some groups was lower than 50% (chance). This result was due to prematurely stopped trials, which occurred when the subjects' response would have resulted in unsafe operation of the BWS system, resulting in trials with 0% or very low correct responses. Therefore, the perception error calculated from the decision curve fits, shown in Fig. 4b, complemented the response accuracy measure to give a more complete picture of the participants' dynamic load perception.

Both the response accuracy and perception error revealed similar results, where the stroke participants had worse dynamic load perception than controls. It is interesting to note that the two control groups were not significantly different in both measures, showing that age was not a factor that significantly affected dynamic load perception in the lower extremities. The phase of the gait cycle during which we asked the subjects to perceive loads resulted in different response accuracy and perception error. It was easier for participants to perceive loads during heel strike and mid stance, and this was similar in both control groups and the stroke group.

3.2. Other sensorimotor outcome measures

The group averages and statistics are presented in Table 4. Although the stroke participants had higher force detection threshold and larger static load perception errors, the differences between groups were not significantly different. The stroke participants had larger errors and performed slower in the lower extremity coordination test. The reach error was significantly higher in the stroke participants compared to both control groups during eyes closed condition, indicating poor proprioception and kinesthesia. Balance, as measured by the change in COP map with and without vision, was not significantly different between the three subject groups.

Table 4. Statistical table for other sensorimotor outcome measures.

Outcome measures	Stroke		Age-matched control	Young control	*F* statistics	*p*-Value
	Paretic	Non-paretic				
Force detection threshold (g)	78.88 ± 124.02	40.81 ± 93.20	5.451 ± 5.721	0.598 ± 0.494	$F(3, 31) = 1.63$	0.20
Static load perception error (%BW)	6.32 ± 4.40		5.63 ± 2.38	5.10 ± 0.77	$F(2, 22) = 0.26$	0.77
Reach error (mm)						
Eyes open	91.43 ± 48.61	30.85 ± 10.51	29.61 ± 12.00	12.79 ± 5.10	$F(3, 31) = 13.59$	<0.0001
Difference (eyes closed − open)	52.53 ± 74.74	11.63 ± 12.82	1.26 ± 6.44	2.05 ± 10.60	$F(3, 31) = 3.22$	0.0362
Reach time (s)						
Eyes open	1.86 ± 0.76	0.93 ± 0.61	0.82 ± 0.30	0.69 ± 0.19	$F(3, 31) = 8.32$	0.0003
Difference (eyes closed − open)	−0.22 ± 0.57	−0.08 ± 0.41	0.052 ± 0.17	-0.05 ± 0.05	$F(3, 31) = 0.85$	0.48
Width of COP map (cm)						
Difference (eyes closed − open)	0.20 ± 0.36	0.071 ± 0.22	0.070 ± 0.084	F(2, 22) = 0.65	0.53	

Mean (±SD) values of reach error and reach time during the eyes opened condition of the lower extremity "big-toe-to-big-toe" test, and the difference between the eyes closed and eyes opened conditions in reach error, time and center of pressure (COP) map during the balance test.

3.3. Static and gait symmetry

We examined the load symmetry for all participants both in standing and walking (Table 5). The difference between the amount of weight supported by the two legs (non-paretic – paretic) was significantly different between groups ($p = 0.044$) during standing. The stroke survivors put significantly more weight on their non-paretic leg ($p = 0.0237$) when compared to their age-matched controls. When looking at gait symmetry during walking, we combined both the loading and timing by examining the ground reaction forces through the entire gait cycle. The difference between groups was significant in a one-way ANOVA ($p = 0.0003$). The stroke participants had a significant asymmetry compared to both control groups (age-matched $p = 0.0002$; young $p = 0.0022$).

Table 5. Statistical table for symmetry measures.

Symmetry measures	Stroke	Age-matched control	Young control	*F* statistics	*p*-Value
Standing weight asymmetry (%BW)	19.34 ± 22.37	2.31 ± 9.34	1.78 ± 5.43	$F(2, 22) = 3.62$	0.044
Walking weight asymmetry (%BW)	8.12 ± 6.36	−0.21 ± 0.77	0.32 ± 078	$F(2, 22) = 11.84$	0.0003

Mean (±SD) values of force detection threshold measured with monofilaments, static load perception error, standing weight bias and walking weight bias as measures of symmetry. The weight bias was calculated as the difference between the non-paretic and the paretic legs for the stroke group, and the difference between the dominant and non-dominant legs for the control groups.

3.4. Relationship between load symmetry and study parameters

Statistics for the regressions of the static load asymmetry (standing) and dynamic load asymmetry (walking) are reported in Table 6. In the correlations with the dynamic load perception error and response accuracy, data for the heel strike condition was reported. Static force perception, dynamic load response accuracy, and force detection threshold significantly correlated with load symmetry during standing ($p < 0.05$). Both dynamic load perception error and accuracy, and force detection threshold correlated with dynamic load asymmetry. Load asymmetry during standing and walking were correlated ($p < 0.0001$). Scatter plots of the two load asymmetry measures against dynamic load response accuracy are shown in Fig. 5. Both load asymmetry measurements were not correlated with subject age ($p = 0.36$ and $p = 0.21$).

Table 6. Statistical results for regressions between load asymmetry and experimental measures.

	Static load asymmetry				Dynamic load asymmetry			
	F-stats	*p*-Value	R^2	Coefficient	*F*-stats	*p*-Value	R^2	Coefficient
Dynamic load response accuracy	$F(1, 23) = 16.04$	0.0006	0.411	−0.30	$F(1, 23) = 20.74$	0.0001	0.474	−0.21
Dynamic load perception error	$F(1, 23) = 3.94$	0.060	0.146	0.12	$F(1, 23) = 5.35$	0.03	0.189	0.09
Static load perception	$F(1, 23) = 7.13$	0.014	0.237	1.35	$F(1, 23) = 1.95$	0.18	0.078	0.51
Force detection threshold	$F(1, 23) = 7.63$	0.011	0.249	0.10	$F(1, 23) = 37.08$	<0.0001	0.617	0.053

	Static load asymmetry				Dynamic load asymmetry			
	***F*-stats**	***p*-Value**	**R^2**	**Coefficient**	***F*-stats**	***p*-Value**	**R^2**	**Coefficient**
Proprioception error	$F(1, 23) = 2.38$	0.14	0.094	0.050	$F(1, 23) = 1.32$	0.26	0.054	0.025
Balance	$F(1, 23) = 0.024$	0.88	0.001	−209.00	$F(1, 23) = 3.46$	0.076	0.131	768.08
Age	$F(1, 23) = 0.87$	0.36	0.069	0.074	$F(1, 23) = 1.70$	0.21	0.036	0.31

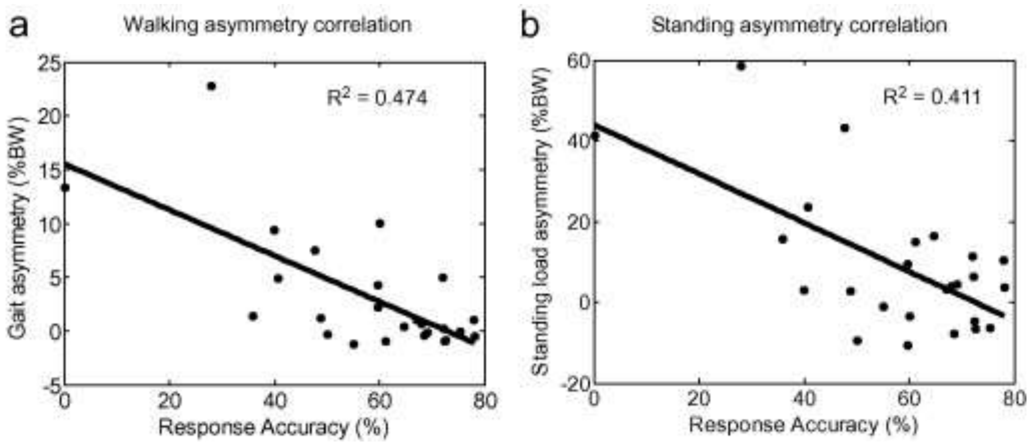

Fig. 5. Correlations of walking (a) and standing (b) asymmetry with response accuracy in the dynamic load perception task.

4. Discussion

Load perception deficits in the lower extremity were observed in stroke survivors. When we examined static load perception, although the post-stroke group performed slightly worse, the differences between the three groups of subjects (post-stroke, age-matched adults, and young adults) were not statistically significant. One possible reason for the small effect was the involvement of both legs in the task. Since the participants only need to replicate a load distribution, the participants can compensate with intact (or better) sensation from the non-paretic leg. The deficits in load perception became apparent when we examined the dynamic load perception. The stroke subjects had lower response accuracy and higher perception errors when compared to both control groups. No difference was observed between the age-matched and young adult control subjects, suggesting that the deficits in dynamic load perception were not part of normal aging. It is interesting to note that the deficits in load perception were not significant in the static and simple tasks, but were significant during the dynamic walking. This showed that the stroke survivors may be able to sense loads when they have full attention on the sensory task, but unable to coordinate multiple sensory and motor cues when performing a functional task. However, isolated sensory and motor tasks are rare in the daily life. Our results highlight the importance of testing perception in a functional context rather than in an isolated test.

The beginning phase of the gait cycle appeared to be easier for the subjects to perceive loads. All subjects groups had higher response accuracy and lower perception error during heel strike and mid stance than in the push off phase. In particular, the performance of the stroke subjects during heel strike was similar to the performance of the control subjects during push off. Load during heel strike and mid stance is likely to be easier to sense as the load is higher and more distinct than during push off. The load during walking peaks at heel strike with dominant high frequency components (Simon et al., 1981). Also, during heel strike, the participants had to judge the difference between two impact forces, while during push off, the subjects had to judge the difference between two self-generated forces. The perception of an impact force is primarily sensory in nature, whereas the perception of a self-generated force also involves the motor system, thus convoluting perception (Shergill et al., 2003). Increased sensitivity to evoked potentials at the end of swing in anticipation of heel strike and decreased sensitivity after footfall during stance has been observed during walking (Duysens et al., 1995). This phasing of sensitivity has been attributed to the gating and facilitation of sensory signals in anticipation of gait events. Efference copy of motor commands suppresses sensations resulting from the voluntary actions (Crapse and Sommer, 2008). During the push off phase of the gait cycle, the increase in sensory gating associated with volition may contribute to the poorer performance in load perception.

Aside from load perception deficits, other sensory deficits are common after a stroke (Carey, 1995, Carey et al., 1996 and Tyson et al., 2008). Previous studies showed that common deficits include tactile discrimination, tactile detection and proprioception (Shah, 1978, Smith et al., 1983, Carey et al., 1993, Kim and Choi-Kwon, 1996, Bohannon, 2003, Leibowitz et al., 2008 and Sullivan and Hedman, 2008). Similar to these studies, we observed impaired proprioception in the stroke survivors with significantly increased error when they closed their eyes. Lower extremity force detection as measured by the Semmes Weinstein monofilaments was worse in particular stroke participants, but overall as a group, the stroke participants were not significantly different from the control groups.

Our results supported the hypothesis that gait asymmetry may be due, in part, to poor load perception in stroke survivors. Similar to previous studies, the stroke survivors in this study had increased gait and standing asymmetry (Wall and Ashburn, 1979, Dettmann et al., 1987, Morita et al., 1995 and Titianova and Tarkka, 1995). Both static and dynamic load perception measures correlated with load asymmetry measures during both standing and walking. Force detection threshold also appeared to be a factor that correlated with gait asymmetry. Significant correlations between load perception and the asymmetry measures suggest that people with poor load perception in their lower limbs presented with worse asymmetry. Sensory impairments in stroke survivors have been associated with motor deficits such as poor balance (Tyson et al., 2006) and with increased incidence of falls (Sorock and Labiner, 1992 and Yates et al., 2002). Our study adds to our understanding of how sensory deficits affect motor tasks, showing that the ability to perceive loads accurately, especially dynamic loads, might play a role in the control of walking.

Other possible factors contributing to gait asymmetries in stroke survivors have been investigated and found to not correlate with the asymmetry measures. Although poor balance perception has been previously correlated with slower walking speed and fewer walking activities (Talkowski et al., 2008) and balance training has been shown to improve stance symmetry (Shumway-Cook et al., 1988, Winstein et al., 1989, Nichols, 1997 and Sackley and Lincoln, 1997), our results showed that balance deficits, as measured as sway during eyes-closed standing, did not correlate with gait asymmetry. It is possible that stroke survivors do not have an accurate perception of their balance, which could discourage them from participating in walking activities. However, our results showed that having poor balance is not directly linked to gait asymmetries. Studies have also related plantarflexor spasticity and muscle strength to gait symmetry (Hsu et al., 2003, Lin et al., 2006 and Laroche et al., 2012). Although we did not directly correlate strength and spasticity measures with gait symmetry, our results showed that the stroke participants have sufficient joint strength needed for walking and did not have significant spasticity in their plantarflexors. Poor proprioception has been linked to deficits in upper extremity coordination in stroke survivors (Sainburg et al., 1993); however, our results showed that poor lower limb proprioception did not affect gait symmetry.

Gait training has been used to improve gait symmetry in the stroke population with mixed results. Spatial symmetry has been shown to improve with various types of gait training, such as body weight supported treadmill training, split belt treadmill training, rhythmic facilitation, and traditional physical therapy such as neurodevelopmental treatment (i.e. NDT) (Hassid et al., 1997, Thaut et al., 1997, Patterson et al., 2010 and Reisman et al., 2013). Some studies have shown temporary improvements in temporal and kinetic symmetry (Hassid et al., 1997 and Reisman et al., 2007), however, others show temporal symmetry remains unchanged after treadmill training or traditional gait rehabilitation (Silver et al., 2000, Den Otter et al., 2006 and Reisman et al., 2013). Since gait symmetry has been linked to gait velocity and motor recovery (Kim and Eng, 2003), it is important to explore other strategies that can improve gait symmetry.

Our results support the use of sensory retraining for the purpose of improving motor function. Sensory retraining has been done to improve sensorimotor function in stroke survivors. Promising results have been reported in sensory retraining in the upper extremities, with improvements in joint position sense, object recognition, discrimination and detection of touch (Carey et al., 1993, Yekutiel and Guttman, 1993, Byl et al., 2003 and Smania et al., 2003). A few studies focusing on the legs have shown some evidence of sensory retraining for postural control (Morioka and Yagi, 2003, Van Peppen et al., 2004 and Hillier and Dunsford, 2006), but the results are far from conclusive. We believe that load perception can be improved with training, and when combined with education, has the potential to improve gait performance post stroke. The potential for load perception training has yet to be explored, and this study provides a basis for such training in stroke survivors.

4.1. Study limitations

A major limitation of the dynamic load perception measure is the confounding factor of sensory cues from the trunk and pelvis. The body weight support is provided through a trunk harness with leg straps. Pressure or movement from the harness on the skin in those areas can provide sensory cues to the participant regarding the amount of body weight support. Caution was taken to strap the harness on as tight as possible to minimize movement in the harness. Also, clear instructions were given to the participants, asking them to focus on different parts of the gait cycle to direct their attention away from the pressure from the harness. Another limitation of the study is the cognitive demand on the participants throughout the task. The task of consciously identifying load on their legs during walking is cognitively demanding. Cognitive ability was not specifically tested in the post-stroke participant group and could have been a factor in the task performance. Furthermore, the gait speeds of the post-stroke participant group were significantly slower than the control group. While the difference in gait speed complicates the comparison between the two groups, we believe that this only biased the comparison such that the differences between groups are smaller. The post-stroke group had more time with each step, given the slower gait speed, potentially allowing a more accurate perception. Finally, a correlation study can only imply that gait asymmetry is related to load perception deficits, and does not prove causation. Further study is needed to determine cause-and-effect relationships between the two measures.

5. Conclusions

Very little is known about the relationship between sensory perception and the motor control of walking. This understanding is further convoluted by the complicated nature of the sensory impairments in stroke survivors. This study specifically isolated the load perception of the lower limbs and studied its effect on gait asymmetry in stroke survivors. We developed a method for measuring dynamic load perception during walking and found that poor load perception correlated with loading asymmetry during standing and walking. Also, load perception during heel strike was found to be more accurate than during push off. The knowledge gained from this study provides a framework for understanding gait-related sensory and motor impairments in stroke survivors.

Acknowledgements

The authors would like to acknowledge financial support for this project from the American Heart Association, MidWest Affiliate Postdoctoral Fellowship (11POST7370016).

Conflict of interest statement: None.

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