Published in final edited form as:

Clin Biomech (Bristol, Avon). 2017 May ; 44: 14–20. doi:10.1016/j.clinbiomech.2017.02.013.

Effects of hip abduction and adduction accuracy on post-stroke gait

Jesse C. Dean^{a,b,c}, Aaron E. Embry^{a,b,c}, Katy H. Stimpson^{a,c}, Lindsay A. Perry^{a,c,d}, and Steven A. Kautz^{a,b,c}

^aRalph H. Johnson VAMC, Charleston, SC, USA

^bDivision of Physical Therapy, Medical University of South Carolina (MUSC), Charleston, SC, USA

^cDepartment of Health Sciences and Research, MUSC, Charleston, SC, USA

Abstract

Background—Gait instability often limits post-stroke function, although the mechanisms underlying this instability are not entirely clear. Our recent work has suggested that one possible factor contributing to post-stroke gait instability is a reduced ability to accurately control foot placement. The purpose of the present experiments was to investigate whether post-stroke gait function is related to the ability to accurately abduct and adduct the hip, as required for accurate foot placement.

Methods—35 chronic stroke survivors and 12 age-matched controls participated in this experiment. Participants performed hip oscillation trials designed to quantify hip abduction/adduction accuracy, in which they lay supine and moved their leg through a prescribed range of motion in time with a metronome. Stroke survivors also performed overground walking trials at their self-selected speed.

Findings—28 of the 35 stroke survivors had sufficient active range of motion to perform the prescribed hip oscillation task. In comparison to controls, these 28 stroke survivors were significantly less accurate at matching the abduction target, matching the adduction target, and moving in time with the metronome. Across these stroke survivors, a multiple regression revealed that only paretic hip abduction accuracy made a unique contribution to predicting paretic step width and paretic step period, metrics of gait performance.

Corresponding author: Jesse C. Dean, deaje@musc.edu, Address: 77 President St., MSC 700, Charleston, SC 29425-5712, USA.

^dLindsay Perry is currently at the University of St. Augustine for Health Sciences., Address: 1 University Blvd., St., Augustine, FL 32806, USA

Conflicts of Interest: None

Author Contributions

Jesse Dean contributed to the conception and design of this study, analysis and interpretation of data, and drafting the manuscript. Aaron Embry, Katy Stimpson, and Lindsay Perry contributed to acquisition of data, and analysis and interpretation of data. Steven Kautz contributed to conception and design of the study, and interpretation of data. All authors revised the manuscript critically for important intellectual content, and approved the version to be submitted.

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.

Interpretation—The present results demonstrate that the ability to accurately abduct the hip is related to post-stroke gait performance, as predicted from a model-based gait stabilization strategy. Therefore, interventions designed to improve lower limb movement accuracy may hold promise for restoring post-stroke gait stability.

Keywords

balance; foot placement; gait stability; hip abduction; motor control; stroke

1. Introduction

Gait instability commonly limits post-stroke functional mobility. Such instability is reflected by an increased fall risk among chronic stroke survivors in comparison to the healthy elderly, with falls often occurring during walking [Weerdesteyn et al. 2008; Batchelor et al. 2012]. Mobility can also be limited by fear of falling, even without a preceding fall [Hyndman et al. 2002; Schmid et al. 2012]. The prevalence of gait instability has motivated the development of clinical tests (e.g. Berg Balance Scale, Dynamic Gait Index) intended to identify individuals with increased fall risk [Blum and Korner-Bitensky 2008; Jonsdottir and Cattaneo 2007]. Similarly, many lab-based measures (92 reported in a recent review) have been proposed as predictive of falls [Hamacher et al. 2011]. These measures vary widely in terms of complexity, ranging from simple linear metrics (e.g. step width variability, margin of stability) [Brach et al. 2005; Hof 2008; Hak et al. 2013] to more complex nonlinear metrics (e.g. Lyapunov exponents, Floquet multipliers, long-term correlations) [Herman et al. 2005; Granata et al. 2008; Lockhart and Liu 2008]. However, neither the clinical tests nor the lab-based measures typically provide insight into the mechanistic causes of instability.

Rather than simply classifying a gait pattern as unstable, it would be beneficial to identify specific limitations that contribute to an individual's instability. Interventions could then be targeted to the underlying causes, allowing both a mechanistic approach and patient-specific treatment. As an example, lower extremity muscular weakness has been linked to fall risk and fear of falling [Mackintosh et al. 2006; Kim et al. 2012]. Therefore, strengthening interventions may improve gait stability in stroke survivors with this limitation. However, strengthening alone is likely insufficient, as a large-scale intervention including task-related strengthening did not reduce post-stroke fall incidence [Dean et al. 2012]. Beyond strength, post-stroke fall risk has been linked to many other factors, including general balance deficits, neglect, and altered sensation or perception [Batchelor et al. 2012]. Despite this complexity, appropriate mechanical interactions with the environment are in every case a necessary component of a stable gait pattern. Therefore, identifying limitations in mechanical behavior that contribute to post-stroke gait instability may be a valuable step in the development of more robust interventions.

We have taken the approach of first investigating how neurologically intact controls maintain gait stability. As mediolateral stability is particularly challenging during bipedal gait [Kuo 1999], we have sought to identify the typical strategy for controlling mediolateral foot placement, an efficient stabilization mechanism [MacKinnon and Winter 1993; Kuo 1999; Bauby and Kuo 2000]. Uninjured controls use a consistent strategy of modulating their

swing leg hip abductor activity based on their body's mechanical state, with this abductor activity influencing foot placement [Rankin et al. 2014]. Briefly, the mediolateral distance between the stance foot and center of mass [CoM] at the start of a step is positively associated with step width [Hurt et al. 2010; Wang and Srinivasan 2014]. Among controls, this stabilization strategy is used both during unperturbed walking and when mediolateral perturbations are applied to the swing leg or CoM [Hof and Duysens 2013; Rankin et al. 2014]. However, a subset of stroke survivors classified as unstable does not exhibit this foot placement strategy for paretic steps [Dean and Kautz 2015], providing indirect evidence for the functional importance of this strategy.

Given the importance of foot placement location and timing for ensuring mediolateral gait stability [Hof 2008; Hak et al. 2013], an inability to accurately control the magnitude and timing of hip abduction and adduction could prevent use of the typical stabilization strategy. The most severe consequence of improper foot placement is an increased risk of overly medial swing foot placement, causing the CoM to pass laterally to the base of support during the subsequent stance phase and requiring a cross-over step to prevent a fall. Stroke survivors may adjust their gait pattern to avoid such losses of balance; specifically, improper paretic foot placement may be compensated for by walking with wider steps, while improper step timing may be compensated for by shortening the stride period [Hof 2008; Hak et al. 2013].

The purpose of the present work was to investigate whether the ability to accurately abduct and adduct the hip (as required to adjust the location and timing of foot placement during the swing phase of gait) is related to post-stroke gait function. Specifically, we tested whether performance of a dynamic hip positioning task with low strength demands was related to paretic step width and step period, as changes in these metrics are often associated with a more conservative gait pattern [Wall et al. 1991; Turnbull et al. 1995]. Secondly, we tested whether hip movement accuracy was related to paretic step width variability (a simple metric of mediolateral stability) and gait speed (a general metric of gait function). The results could motivate interventions focused on improving post-stroke movement accuracy. We hypothesized that stroke survivors would exhibit a reduced ability to accurately abduct and adduct their paretic limb in comparison to controls. Across individual stroke survivors, we hypothesized that poorer paretic hip abduction/adduction accuracy would be associated with wider paretic steps, shorter paretic step periods, increased step width variability, and slower gait speeds.

2. Methods

2.1. Participants

We recruited 35 chronic stroke survivors and 12 neurologically intact controls to participate in this experiment. As this was an exploratory study, our sample sizes were based on previous work that identified significant associations between gait performance and isolated lower extremity function among stroke survivors ($n = 21$) [Lee et al. 2005; Lin et al. 2005], and that reported significant post-stroke deficits in joint motion tracking relative to controls ($n = 11$ per group) [Chung et al. 2006; Halaney et al. 1989; Patten et al. 2003]. All participants were recruited using convenience sampling from a database housed at the

Medical University of South Carolina, contacting individuals who met the study's basic inclusion and exclusion criteria. Stroke survivor inclusion criteria were: a stroke resulting in unilateral paresis at least 6 months prior to recruitment; ability to walk independently without external bracing or an assistive device; bilateral active frontal plane hip range of motion of at least 20°. Exclusion criteria were: legal blindness or severe visual impairment; inability to follow 3-step commands; depressive symptomology (PHQ-9>5); cardiac instability or significant medical complications limiting functional capabilities. Control participants were recruited to approximately match the age distribution of the stroke survivors. All participants provided informed consent using a form approved by the MUSC Institutional Review Board, and consistent with the Declaration of Helsinki.

2.2. Hip oscillation task: experimental procedure

A reclined hip oscillation task was used to quantify participants' ability to accurately abduct and adduct their hip during dynamic, active movement. The task was designed to minimize the muscle forces required to produce the movement (reducing the effects of potential strength deficits) and mimic the sources of sensory feedback typically available from the swing leg during walking. Participants lay supine on a padded table with one leg supported by a sling hanging from a low-friction track, such that minimal hip torques were required to move the leg. The knee of the other leg was bent to 90°, and supported on a lower surface. To prevent participants from using cutaneous feedback to sense their movement, they were positioned so the sling-supported leg was entirely off the table and would not make contact with the contralateral leg during hip oscillation. Participants' vision of their legs was blocked by a monitor and curtain. Participants wore noise-cancelling headphones playing brown noise to prevent them from using auditory feedback to monitor leg motion. Participants completed at least two practice trials (30 seconds each) of the hip oscillation task before data collection. Additional practice trials were repeated as necessary until the experimenter was confident that the participant understood the requirements of the task.

A 12-camera motion capture system (PhaseSpace; San Leandro, CA, USA) with spatial resolution of 0.25 mm was used to record leg motion. Six active LED markers were placed bilaterally on the ASIS, the anterior aspect of the proximal tibia of each leg, and the anterior aspect of the distal tibia of each leg. Marker positions were sampled at 60 Hz, and used to calculate the hip abduction/adduction angle of the supported leg in near real-time, by calculating the horizontal plane angle between a vector connecting the two ASIS markers and a vector connecting the proximal and distal tibia markers (effective angular resolution of 0.1°). Real-time hip angle values were displayed graphically on the monitor (Fig 1A).

Participants performed nine 30-second hip oscillation trials with each leg. In all trials, participants were instructed to move in time with a metronome played through the headphones at 40 BPM, just reaching the peak abduction and adduction angles at each tone (cycle period = 3 seconds). Movement amplitude was prescribed to one of three levels (10°, 15°, 20° peak-to-peak) using target lines displayed on the monitor, with three trials performed at each amplitude. The midpoint of the prescribed motion was always 5° hip abduction, and scaling on the monitor was varied so the targets for each prescribed amplitude appeared the same to participants. Trial order was assigned using a randomized

block design (i.e. each “block” of three trials included one trial at each prescribed amplitude, in random order). For the largest prescribed amplitude, following the metronome in a sinusoidal pattern would result in a maximum angular velocity of ~20 degrees/second, approximately matching the peak swing phase hip abduction velocity from published normative gait data [Vaughan et al. 1999]. Participants were provided with at least 30-seconds of rest between trials.

2.3. Hip oscillation task: data analysis

The calculated hip angle was low-pass filtered at 10 Hz with a 3rd order Butterworth filter. We identified the peak abduction and adduction angles using the MATLAB findpeaks function (separated by at least 1.5 seconds), as well as the absolute error between these peak angles and the matching target angle. We calculated movement period as the time between consecutive peak abduction angles (Fig. 1B), as well as the absolute error in comparison to the prescribed 3 second period. We did not include the first 6-seconds of the trial, allowing participants to perform two complete movement cycles of practice with the prescribed task. Therefore, a total of 24 seconds (~8 movement cycles) were included in our analyses.

While all participants had sufficient range of motion to reach the hip positions required for the hip oscillation task under static conditions (see inclusion criteria), we determined in offline analysis whether each participant had sufficient range of motion to perform the more challenging prescribed dynamic task. Any participant consistently unable to achieve the largest prescribed hip abduction or adduction angle was omitted from subsequent analyses. For other participants, we calculated the following average metrics for each prescribed range of motion: peak abduction angle; peak adduction angle; absolute error for peak abduction angle; absolute error for peak adduction angle; movement period; and absolute error in movement period.

2.4. Walking: experimental procedure

Stroke survivors performed five trials in which they walked at self-selected speed over a 7 m instrumented mat (GAITRite; Franklin, NJ, USA), which recorded foot pressure location at a sampling rate of 80 Hz. Participants had sufficient space (2 m) before and after the mat to accelerate and decelerate (steps were not measured), and wore a safety harness attached to an overhead rail to prevent a fall in case of a loss of balance.

2.5. Walking: data analysis

For walking trials, we calculated each stroke survivor’s average paretic step width, average paretic step period, paretic step width variability (standard deviation across all trials), and gait speed using data collected by the instrumented mat. These measures correspond to the gait characteristics that we hypothesized would be affected by an inability to accurately abduct and adduct the hip.

2.6. Statistics

We first tested whether stroke survivors and controls differed in their ability to accomplish the prescribed hip oscillation task, using a series of 2-way ANOVAs with interactions (type 3 sum of squares). The independent variables were group (control legs vs. non-paretic legs

vs. paretic legs) and prescribed amplitude (10° vs. 15° vs. 20°), while the dependent variables were peak abduction angle, peak adduction angle, and movement period. We then tested whether stroke survivors and controls differed in their ability to move their hip accurately on a cycle-by-cycle basis (independent of prescribed amplitude), using a series of 1-way ANOVAs. Here, the independent variable was group (control legs vs. non-paretic legs vs. paretic legs), while the dependent variables were absolute abduction error, absolute adduction error, and absolute movement period error. For all ANOVAs, we performed Tukey-Kramer post-hoc tests when appropriate to identify significant differences across the independent variables.

We also tested whether a stroke survivor's ability to accurately perform the hip oscillation task with their paretic leg was related to their gait characteristics. As an initial indication of the direction and strength of these potential relationships, we first performed Pearson correlations between the paretic hip oscillation error measures (abduction, adduction, and movement period) and gait measures (paretic step width, paretic step period, paretic step width variability, and gait speed). Correlation magnitudes were classified as absent (<0.25), fair (0.25 – 0.50), moderate to good (0.50 – 0.75), or good to excellent (>0.75) [Portney and Watkins 2015]. To more clearly test whether individual hip oscillation measures made a unique contribution to predicting gait performance, we performed a series of multiple regressions using the same independent and dependent variables. All statistical tests were performed using the MATLAB Statistics and Machine Learning Toolbox. P-values less than 0.05 were interpreted as significant.

3. Results

3.1. Hip oscillation task

All 12 control participants were able to successfully perform the hip oscillation task. Of the 35 tested stroke survivors, 28 were able to successfully perform this task. 7 had insufficient dynamic range of motion, and are thus not included in subsequent analyses. Basic demographics of all included participants are provided in Table 1.

During hip oscillation, the average movement pattern was quite similar between controls and stroke survivors. Peak abduction angle scaled as expected with the prescribed movement amplitude ($p < 0.001$), but did not vary significantly between groups ($p = 0.20$) and was not influenced by a significant interaction between amplitude and group ($p = 0.07$) (Fig. 2A). Similarly, peak adduction angle also scaled with the prescribed amplitude ($p < 0.001$), but was not significantly affected by group ($p = 0.55$) or an interaction between amplitude and group ($p = 0.31$) (Fig. 2B). Movement period was consistent across conditions, with no significant effect of prescribed amplitude ($p = 0.85$), group ($p = 0.39$), or an interaction between amplitude and group ($p = 0.72$) (Fig. 2C).

While the average movement patterns were similar across groups, stroke survivors were less accurate on a cycle-by-cycle basis. The absolute error in peak abduction angle was significantly influenced by group ($p = 0.002$), and was significantly larger in paretic legs than control legs (Fig. 3A). Absolute error in peak adduction angle was also significantly influenced by group ($p = 0.002$), with significantly larger errors for both paretic and non-

paretic legs compared to controls (Fig. 3B). Finally, absolute error in movement period was significantly influenced by group ($p=0.005$), and was significantly larger in paretic legs than control legs (Fig. 3C).

3.2. Relationships between post-stroke hip oscillation and walking performance

A stroke survivor's ability to accurately perform the hip oscillation task with their paretic leg was related to their gait characteristics, with the results of linear correlations presented in Table 2. Briefly, larger errors in peak abduction angle, peak adduction angle, and movement period were all associated with wider paretic steps, longer paretic step periods, and slower gait speeds (with fair to moderate correlation magnitudes ranging from 0.39 to 0.59). In contrast, none of our measures of hip oscillation performance were significantly associated with paretic step width variability.

Regression analyses revealed a unique contribution of hip abduction error to predicting several gait characteristics, with results presented in Table 3. The combination of abduction error, adduction error, and movement period error predicted 40% of the variation in average paretic step width, but only abduction error made a statistically significant independent contribution. Similarly, the combination of these three error metrics predicted 38% of the variation in average paretic step period, but again only abduction error made an independent contribution. In contrast, intersubject variation in paretic step width variability was not significantly predicted by the combination of our error metrics. Finally, 37% of the variation in gait speed was predicted by our three error metrics, although none of these metrics made an independent contribution.

4. Discussion

The present work investigated the relationship between hip abduction/adduction accuracy and gait function in chronic stroke survivors. As hypothesized, stroke survivors as a group were less accurate than controls at performing the hip oscillation task, in terms of peak hip abduction, peak hip adduction, and movement period errors. Our hypotheses relating hip oscillation performance to gait performance were only partially supported, as increased paretic hip abduction errors were predictive of wider paretic steps but also of longer paretic step periods.

Stroke survivors were less accurate than age-matched controls at abducting and adducting their hip, particularly for the paretic leg. However, these errors were relatively small, raising the question of their functional relevance. Across the tested stroke survivors, average paretic hip abduction error varied from 0.5° in the most accurate quartile to 1.3° in the least accurate quartile. For an individual of average leg length, we can estimate that such hip abduction errors would correspond to mediolateral foot placement errors of 0.8 cm and 2.1 cm, respectively. As comparison, young controls have been reported to walk with an average mediolateral margin of stability of only 1.6 cm, a metric relating the foot center of pressure location to the location and velocity of the CoM [Hof et al. 2007]. In this context, a foot placement error of over 2 cm could have a major impact on gait stability.

Performance of our hip oscillation task required participants to sense the position of their leg (likely using visual and proprioceptive feedback), process this multisensory feedback, and generate an appropriate descending motor command. Therefore, post-stroke deficits in sensation, sensorimotor processing, or motor coordination could contribute to the decreased accuracy. Our results are thus unable to isolate specific sensory deficits (e.g. proprioception), which are typically quantified using detection or matching of passively applied movements [Suetterlin and Sayer 2014; Hillier et al. 2015]. However, even these tests of conscious body position or movement awareness may provide only limited insight into the subconscious use of sensory feedback during active, dynamic movements [Bhanpuri et al. 2013; Aman et al. 2015; Chu et al. 2015], when such feedback is likely to be most important functionally [Suetterlin and Sayer 2014; Hillier et al. 2015].

Across individual stroke survivors, the ability to accurately abduct the paretic leg was related to some aspects of gait performance. While we found many significant correlations between our metrics of hip oscillation accuracy and gait, regressions found only hip abduction accuracy to make an independent contribution. This apparent relative importance of hip abduction accuracy is likely a result of the primary role played by the hip abductors in controlling mediolateral foot placement [Hof and Duysens 2013; Rankin et al. 2014]. Larger hip abduction errors predicted increased step width, likely a compensatory response to the perceived risk of lateral losses of balance [Maki 1997; IJmker et al. 2015]. However, these wider steps may not be an entirely beneficial response to instability, as increased step widths appear to contribute to the large fluctuations in frontal plane angular momentum once unstable stroke survivors transition into paretic single leg stance [Nott et al. 2014]. Contrary to our expectations, larger hip abduction errors actually predicted longer step periods. While we hypothesized that individuals with poor accuracy would use faster steps to help ensure their stability (as observed previously; [Hak et al. 2013]), they instead slowed their steps along with their overall gait speed, possibly to allow more time to achieve a safe paretic foot placement location. Surprisingly, none of the quantified measures of hip movement accuracy were significantly associated with step width variability, a simple measure of mediolateral gait stability. This negative result could be attributed to either the non-monotonic relationship between step width variability and fall-risk [Brach et al. 2005], or the need for substantially more steps (~400) to accurately quantify this measure [Owings and Grabiner 2003].

While the design of our hip oscillation task allowed us to focus on post-stroke deficits in hip abduction/adduction accuracy, other limitations could clearly also influence gait function. Indeed, our regressions suggest that ~60% of the intersubject variability in paretic step width and step period is attributable to other causes. For example, motor dyscoordination causing the hip circumduction commonly observed in low-functioning stroke survivors [Chen et al. 2005; Stanhope et al. 2014] could lead to wider steps. Alternatively, post-stroke deficits in accurately perceiving the body's mechanical state - such as the perception of CoM mechanics [de Oliveira et al. 2008] or lower extremity loading [Chu et al. 2015] - could indirectly contribute to the preference of a more conservative gait pattern with slower, wider steps. To optimally improve post-stroke gait stability and overall function, interventions must be developed to address each of these potential deficits.

While the present results demonstrate a link between hip abduction accuracy and post-stroke gait, several limitations must be addressed. Most notably, we did not directly quantify gait stability. Future work should relate hip abduction accuracy to use of the typical gait stabilization strategy and clinical measures of stability. We quantified mediolateral gait motion using step width, due to its relative ease of measurement and link to gait instability and fear of falling [Maki 1997]. However, accurate leg positioning could be more closely related to mediolateral foot placement, which quantifies leg position relative to the CoM during walking and can exhibit trends distinct from step width [Balasubramanian et al. 2010; Roden-Reynolds et al. 2015]. More detailed kinematic measurements would allow us to test whether hip abduction accuracy is related to mediolateral foot placement accuracy, possibly while stepping to visually presented targets [Zissimopoulos et al. 2014; Reissman and Dhaer 2015]. This initial study investigated unperturbed forward walking at a self-selected speed, as even this seemingly easy task is the cause of many falls [Robinovitch et al. 2013]. Our results could be extended by testing whether appropriate foot positioning also influences post-stroke gait performance with more challenging tasks, such as changes in direction or overcoming external perturbations [Hof 2008]. We quantified hip abduction and adduction accuracy using a relatively simple task in which visual feedback of leg position was available to participants, and movement period remained constant across trials. Further insight may be possible by reducing the available visual feedback [Vidoni and Boyd 2009] to investigate individuals' use of proprioceptive feedback, and by varying the mechanical parameters of the task to isolate displacement from velocity effects [Steyvers et al. 2001]. Finally, we focused on relationships between behavioral measures that could conceivably be monitored in a physical therapy clinic, and did not control for factors such as stroke location that can influence sensorimotor function [Collins 2007]. Ideally, the type of behavioral measurements described here can be combined with neurophysiological and clinical measurements in order to identify the cause of the altered behavior limiting gait function, and select the best treatment option for an individual patient.

Despite these limitations, the present results have implications for future clinical studies. By minimizing the active torque demands of the hip oscillation task, we were able to identify deficits in movement accuracy that are not solely due to weakness, and relate these deficits to gait performance. Therefore, functional mobility may be improved by training methods focused on post-stroke movement accuracy, including step accuracy training currently under investigation [Heeren et al. 2013; Hollands et al. 2015]. In addition, interventions designed to enhance available sensory feedback could contribute to improvements in movement accuracy. Specific to hip abduction, vibratory stimuli may hold promise for modulating the sensory feedback used to control foot placement location [Walker et al. 2014; Roden-Reynolds et al. 2015]. An appropriate combination of behavioral training and enhanced sensation may allow improvement of post-stroke hip abduction accuracy and gait function.

Acknowledgments

Role of the Funding Sources

This work was partially funded by a grant from the Department of Veterans Affairs under grant number IK2 RX000750. This research was also supported by the COBRE for Stroke Recovery, through an Institutional Development Award (IDeA) from the National Institute of General Medical Sciences of the National Institutes of Health.

Health under grant number P20GM109040. This work does not represent the view of the Department of Veterans Affairs or the United States Government. The funding sources had no involvement in study design; collection, analysis, and interpretation of data; writing of the report; or decision to submit the manuscript for publication.

References

- Aman JE, Elangovan N, Yeh IL, Konczak J. The effectiveness of proprioceptive training for improving motor function: a systematic review. *Front Hum Neurosci*. 2015; 8:1075. [PubMed: 25674059]
- Balasubramanian CK, Neptune RR, Kautz SA. Foot placement in a body reference frame during walking and its relationship to hemiparetic walking performance. *Clin Biomech*. 2010; 25:483–490.
- Batchelor FA, Mackintosh SF, Said CM, Hill KD. Falls after stroke. *Int J Stroke*. 2012; 7:482–490. [PubMed: 22494388]
- Bauby CE, Kuo AD. Active control of lateral balance in human walking. *J Biomech*. 2000; 33:1433–1440. [PubMed: 10940402]
- Bhanpuri NH, Okamura AM, Bastian AJ. Predictive modeling by the cerebellum improves proprioception. *J Neurosci*. 2013; 33:14301–14306. [PubMed: 24005283]
- Blum L, Korner-Bitensky N. Usefulness of the Berg Balance Scale in stroke rehabilitation: a systematic review. *Phys Ther*. 2008; 88:559–556. [PubMed: 18292215]
- Brach JS, Berlin JE, VanSwearingen JM, Newman AB, Studenski SA. Too much or too little step width variability is associated with a fall history in older persons who walk at or near normal gait speed. *J NeuroEng Rehabil*. 2005; 2:21. [PubMed: 16042812]
- Chen G, Pattern C, Kothari DH, Zajac FE. Gait differences between individuals with post-stroke hemiparesis and non-disabled controls at matched speeds. *Gait Posture*. 2005; 22:51–56. [PubMed: 15996592]
- Chu VW, Hornby TG, Schmit BD. Perception of lower extremity loads in stroke survivors. *Clin Neurophysiol*. 2015; 126:372–381. [PubMed: 25097091]
- Chung YJ, Cho SH, Lee YH. Effect of the knee joint tracking training in closed chain kinetic condition for stroke patients. *Rest Neurol Neurosci*. 2006; 24:173–180.
- Collins C. Pathophysiology and classification of stroke. *Nurs Stand*. 2007; 21:35–39.
- de Oliveira CB, de Medeiros IRT, Frota NAF, Greters ME, Conforto AB. Balance control in hemiparetic stroke patients: main tools for evaluation. *J Rehab Res Dev*. 2008; 45:1215–1226.
- Dean CM, Rissel C, Sherrington C, Sharkey M, Cumming RG, Lord SR, Barker RN, Kirkham C, O'Rourke S. Exercise to enhance mobility and prevent falls after stroke: the community stroke club randomized trial. *Neurorehab Neur Rep*. 2012; 26:1046–1057.
- Dean JC, Kautz SA. Foot placement control and gait instability among people with stroke. *J Rehabil Res Dev*. 2015; 52:577–590. [PubMed: 26437301]
- Granata KP, Lockhart TE. Dynamic stability differences in fall-prone and healthy adults. *J Electromyogr Kinesiol*. 2008; 18:172–178. [PubMed: 17686633]
- Hak L, Houdijk H, van der Wurff P, Prins MR, Mert A, Beek PJ, van Dieen JH. Stepping strategies used by post-stroke individuals to maintain margins of stability during walking. *Clin Biomech*. 2013; 28:1041–1048.
- Halaney ME, Carey JR. Tracking ability of hemiparetic and healthy subjects. *Phys Ther*. 1989; 69:342–348. [PubMed: 2710816]
- Hamacher D, Singh NB, Van Dieen JH, Heller MO, Taylor WR. Kinematic measures for assessing gait stability in elderly individuals: a systematic review. *J R Soc Interface*. 2011; 8:1682–1698. [PubMed: 21880615]
- Heeren A, van Ooijen MW, Geurts ACH, Day BL, Janssen TWJ, Beek PJ, Roerdink M, Weerdesteyn V. Step by step: a proof of concept study of C-Mill gait adaptability training in the chronic phase after stroke. *J Rehabil Med*. 2013; 45:616–622. [PubMed: 23811818]
- Herman T, Giladi N, Gurevich T, Hausdorff JM. Gait instability and fractal dynamics of older adults with a “cautious” gait: why do certain older adults walk fearfully? *Gait Posture*. 2005; 21:178–185. [PubMed: 15639397]
- Hillier S, Immink M, Thewlis D. Assessing proprioception: a systematic review of possibilities. *Neurorehab Neur Rep*. 2015; 29:933–949.

- Hof AL. The 'extrapolated center of mass' concept suggests a simple control of balance in walking. *Hum Mov Sci.* 2008; 27:112–125. [PubMed: 17935808]
- Hof AL, Duysens J. Responses of human hip abductor muscles to lateral balance perturbations during walking. *Exp Brain Res.* 2013; 230:301–310. [PubMed: 23934442]
- Hof AL, van Bockel RM, Schoppen T, Postema K. Control of lateral balance in walking. Experimental findings in normal subjects and above-knee amputees. *Gait Posture.* 2007; 25:250–258. [PubMed: 16740390]
- Hollands KL, Pelton TA, Wimperis A, Whitham D, Tan W, Jowett S, Sackley CM, Wing AM, Tyson SF, Mathias J, Hensman M, van Vliet PM. Feasibility and preliminary efficacy of visual due training to improve adaptability of walking after stroke: multi-centre, single-blind randomised control pilot trial. *PLOS ONE.* 2015; 10:e0139251. [PubMed: 26406237]
- Hurt CP, Rosenblatt N, Crenshaw JR, Grabiner MD. Variation in trunk kinematics influences variation in step width during treadmill walking by older and younger adults. *Gait Posture.* 2010; 31:461–464. [PubMed: 20185314]
- Hyndman D, Ashburn A, Stack E. Fall events among people with stroke living in the community: circumstances of falls and characteristics of fallers. *Arch Phys Med Rehabil.* 2002; 83:165–170. [PubMed: 11833018]
- IJmker T, Lamoth CJ, Houdijk H, Tolsma M, van der Woude LHV, Daffertshofer A, Beek PJ. Effects of handrail hold and light touch on energetics, step parameters, and neuromuscular activity during walking after stroke. *J NeuroEng Rehabil.* 2015; 12:70. [PubMed: 26298647]
- Jonsdottir J, Cattaneo D. Reliability and validity of the dynamic gait index in persons with chronic stroke. *Arch Phys Med Rehabil.* 2007; 88:1410–1415. [PubMed: 17964880]
- Kim EJ, Kim DY, Kim WH, Lee KL, Yoon YH, Park JM, Shin JI, Kim SK, Kim DG. Fear of falling in subacute hemiplegic stroke patients: associating factors and correlations with quality of life. *Ann Rehabil Med.* 2012; 36:797–803. [PubMed: 23342312]
- Kuo AD. Stabilization of lateral motion in passive dynamic walking. *Int J Rob Res.* 1999; 18:917–930.
- Lee MJ, Kilbreath S, Refshauge KM. Movement detection at the ankle following stroke is poor. *Aus J Physiother.* 2005; 51:19–24.
- Lin SI. Motor function and joint position sense in relation to gait performance in chronic stroke patients. *Arch Phys Med Rehabil.* 2005; 86:197–203. [PubMed: 15706543]
- Lockhart TE, Liu J. Differentiating fall-prone and healthy adults using local dynamic stability. *Ergonomics.* 2008; 51:1860–1872. [PubMed: 19034782]
- Mackintosh SF, Hill KD, Dodd KJ, Goldie PA, Culham EG. Balance score and a history of falls in hospital predict recurrent falls in the 6 months following stroke rehabilitation. *Arch Phys Med Rehabil.* 2006; 87:1583–1589. [PubMed: 17141637]
- MacKinnon CD, Winter DA. Control of whole body balance in the frontal plane during human walking. *J Biomech.* 1993; 26:633–644. [PubMed: 8514809]
- Maki BE. Gait changes in older adults: predictors of falls or indicators of fear? *J Am Geriatr Soc.* 1997; 45:313–320. [PubMed: 9063277]
- Nott CR, Neptune RR, Kautz SA. Relationships between frontal plane angular momentum and clinical balance measures during post-stroke hemiparetic walking. *Gait Posture.* 2014; 39:129–134. [PubMed: 23820449]
- Owings TM, Grabiner MD. Measuring step kinematic variability on an instrumented treadmill: how many steps are enough? *J Biomech.* 2003; 36:1215–1218. [PubMed: 12831749]
- Patten C, Kothari D, Whitney J, Lexell J, Lum PS. Reliability and responsiveness of elbow trajectory tracking in chronic poststroke hemiparesis. *J Rehabil Res Dev.* 2003; 40:487–500. [PubMed: 15077661]
- Portney, LG., Watkins, MP. Foundations of clinical research: applications to practice. 3. F.A. Davis Company; Philadelphia, Pennsylvania: 2015.
- Rankin BL, Buffo SK, Dean JC. A neuromechanical strategy for mediolateral foot placement in walking humans. *J Neurophysiol.* 2014; 112:374–383. [PubMed: 24790168]
- Reissman ME, Dhafer YY. A functional tracking task to assess frontal plane motor control in post stroke gait. *J Biomech.* 2015; 48:1782–1788. [PubMed: 26037229]

- Robinovitch SN, Feldman F, Yang Y, Schonnop R, Leung PM, Sarraf T, Sims-Gould J, Loughin M. Video capture of the circumstances of falls in elderly people residing in long-term care: an observational study. *Lancet*. 2013; 381:47–54. [PubMed: 23083889]
- Roden-Reynolds DC, Walker MH, Wasserman CR, Dean JC. Hip proprioceptive feedback influences the control of mediolateral stability during human walking. *J Neurophysiol*. 2015; 114:2220–2229. [PubMed: 26289467]
- Schmid AA, Van Puymbroeck M, Altenburger PA, Dierks TA, Miller KK, Damush TM, Williams LS. Balance and balance self-efficacy are associated with activity and participation after stroke: a cross-sectional study in people with chronic stroke. *Arch Phys Med Rehabil*. 2012; 93:1101–1107. [PubMed: 22502804]
- Stanhope VA, Knarr BA, Reisman DS, Higginson JS. Frontal plane compensatory strategies associated with self-selected walking speed in individuals post-stroke. *Clin Biomech*. 2014; 29:518–522.
- Steyvers M, Verschueren SMP, Levin O, Ouamer M, Swinnen SP. Proprioceptive control of cyclical bimanual forearm movements across different movement frequencies as revealed by means of tendon vibration. *Exp Brain Res*. 2001; 140:326–334. [PubMed: 11681308]
- Suetterlin KJ, Sayer AA. Proprioception: where are we now? A commentary on clinical assessment, changes across the life course, functional implications and future interventions. *Age Aging*. 2014; 43:313–318.
- Turnbull GI, Charteris MS, Wall JC. A comparison of walking speeds between normal and hemiplegic subjects. *Scand J Rehabil Med*. 1995; 27:175–182. [PubMed: 8602480]
- Vaughan, CL., Davis, BL., O'Connor, JC. *Dynamics of Human Gait*. 2. Kihobo Publishers; Cape Town, South Africa: 1999.
- Vidoni ED, Boyd LA. Preserved motor learning after stroke is related to degree of proprioceptive deficit. *Behav Brain Func*. 2009; 5:36.
- Walker ER, Hyngstrom AS, Schmit BD. Sensory electrical stimulation improves foot placement during targeted stepping post-stroke. *Exp Brain Res*. 2014; 232:1137–1143. [PubMed: 24449007]
- Wall JC, Hogan DB, Turnbull GI, Fox RA. The kinematics of idiopathic gait disorder. *Scand J Rehabil Med*. 1991; 23:159–164. [PubMed: 1962159]
- Wang Y, Srinivasan M. Stepping in the direction of the fall: the next foot placement can be predicted from current upper body state in steady-state walking. *Biol Lett*. 2014; 10:20140405. [PubMed: 25252834]
- Weerdesteyn V, de Niet M, van Duijnhoven HJR, Geurts ACH. Falls in individuals with stroke. *J Rehab Res Dev*. 2008; 45:1195–1214.
- Zissimopoulos A, Stine R, Fatone S, Gard S. Mediolateral foot placement ability during ambulation in individuals with chronic post-stroke hemiplegia. *Gait Posture*. 2014; 39:1097–1102. [PubMed: 24582515]

Highlights

- Functional mobility of individuals with stroke is often limited by gait instability
- Accurate mediolateral foot placement typically helps stabilize gait
- Individuals with stroke were less accurate at abducting and adducting their paretic hip
- Hip abduction accuracy was predictive of post-stroke gait performance

Figure 1.

Hip oscillation task description. A) Participants actively moved their legs between hip abduction and adduction visual targets in time with a metronome. Near real-time (updated at 60-Hz) visual feedback of the leg angle was provided in the form of a vertical line, illustrated here for a right leg trial. B) Peak abduction and adduction angles were quantified for each movement cycle, and movement period was calculated as the time between peak abduction angles.

Figure 2.

Average hip oscillation task performance. The effects of group (control vs. non-paretic vs. paretic legs) and prescribed amplitudes (10° vs. 15° vs. 20°) are illustrated for peak abduction angle (A), peak adduction angle (B), and movement period (C). For all panels, data points indicate means and error bars indicate 95% confidence intervals. Asterisks (*) illustrate significant post-hoc differences between the indicated prescribed amplitudes, collapsed across leg groups.

Figure 3.

Hip oscillation task errors. The effects of group (control vs. non-paretic vs. paretic legs) are illustrated for peak abduction error (A), peak adduction error (B), and movement period error (C). Data points indicate means and error bars indicate 95% confidence intervals. Asterisks (*) illustrate significant post-hoc differences between the indicated leg groups.

Table 1

Demographic characteristics of participants. All numerical values are reported as means (standard deviations), with none of these metrics differing significantly between controls and stroke survivors. All uninjured controls self-reported as right leg dominant.

	Controls (n=12)	Stroke survivors (n=28)
Age (years)	56 (6)	55 (15)
Mass (kg)	87 (16)	80 (14)
Height (cm)	175 (8)	171 (11)
Leg length (cm)	94 (6)	93 (8)
Gender	4F / 8M	14 F / 14 M
Paretic side	N/A	12 L / 16 R
Time since stroke (mo)	N/A	34 (27)

Table 2

Correlations between hip oscillation metrics (columns) and gait metrics (rows). Results are presented as the Pearson correlation coefficient, followed by the corresponding p-value in parentheses. Significant correlations are bolded and in italics

	Abduction error	Adduction error	Period error
Paretic step width	<i>0.58 (0.001)</i>	<i>0.41 (0.032)</i>	<i>0.48 (0.010)</i>
Paretic step period	<i>0.59 (0.001)</i>	<i>0.39 (0.040)</i>	<i>0.42 (0.027)</i>
Paretic step width variability	0.17 (0.38)	0.11 (0.58)	−0.06 (0.76)
Gait speed	<i>−0.55 (0.002)</i>	<i>−0.45 (0.016)</i>	<i>−0.49 (0.008)</i>

Table 3

Regression results between hip oscillation metrics and gait metrics. Each row presents the results of a multiple regression predicting the indicated gait metric. For each hip oscillation metric, the regression coefficient is presented, followed by the 95% confidence interval in brackets, and the partial η^2 value below. Significant contributing factors are indicated in bolded italics. The final two columns present the R^2 values and p-values for the total regression models, with significance again indicated by bolded italics.

	Total Model				
	Abduction error	Adduction error	Period error	Constant term	R^2 p-value
Paretic step width	8.7 [0.8 16.7] $\eta^2=0.18$	-2.7 [-11.0 5.6] $\eta^2=0.02$	7.0 [-2.3 16.3] $\eta^2=0.09$	9.2 [4.4 14.0]	0.40 0.006
Paretic step period	0.33 [0.06 0.60] $\eta^2=0.21$	-0.11 [-0.39 0.18] $\eta^2=0.03$	0.17 [-0.15 0.49] $\eta^2=0.05$	0.46 [0.29 0.62]	0.38 0.008
Paretic step width variability	1.0 [-1.3 3.3] $\eta^2=0.03$	0.0 [-2.4 2.4] $\eta^2=0.00$	-1.1 [-3.8 1.6] $\eta^2=0.03$	2.6 [1.2 4.0]	0.06 0.70
Gait speed	-43 [-98 12] $\eta^2=0.10$	1 [-56 58] $\eta^2=0.00$	-49 [-113 15] $\eta^2=0.09$	134 [101 167]	0.37 0.01