Validation of simplified centre of mass models during gait in individuals with chronic stroke

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Abstract

Background—The feasibility of using a multiple segment (full-body) kinematic model in clinical gait assessment is difficult when considering obstacles such as time and cost constraints. While simplified gait models have been explored in healthy individuals, no such work to date has been conducted in a stroke population. The aim of this study was to quantify the errors of simplified kinematic models for chronic stroke gait assessment.

Methods—Sixteen individuals with chronic stroke (>6 months), outfitted with full body kinematic markers, performed a series of gait trials. Three centre of mass models were computed: (i) 13-segment whole-body model, (ii) 3 segment head-trunk-pelvis model, and (iii) 1 segment pelvis model. Root mean squared error differences were compared between models, along with correlations to measures of stroke severity.

Findings—Error differences revealed that, while both models were similar in the mediolateral direction, the head-trunk-pelvis model had less error in the anteroposterior direction and the pelvis model had less error in the vertical direction. There was some evidence that the head-trunk-pelvis model error is influenced in the mediolateral direction for individuals with more severe strokes, as a few significant correlations were observed between the head-trunk-pelvis model and measures of stroke severity.

Interpretation—These findings demonstrate the utility and robustness of the pelvis model for clinical gait assessment in individuals with chronic stroke. Low error in the mediolateral and vertical directions is especially important when considering potential stability analyses during gait for this population, as lateral stability has been previously linked to fall risk.

Keywords

Stroke; Gait; Rehabilitation

1. Introduction

Capturing whole-body centre of mass (CoM) motion through a large set of markers placed on anatomical landmarks across the body is considered the gold standard for quantifying whole-body movement (Eng & Winter, 1993), such as walking (Gutierrez-Farewik et al, 2006). However, the setup and capture for a full-body marker set can greatly limit application to clinical practice where clinicians have limited time to spend with their patients. Previous work with healthy adults has demonstrated that a head-trunk-pelvis segment (HTP) combination or simply a pelvic segment (PEL) can estimate whole-body centre of mass displacements during gait, with <2cm mean error compared to the whole body model in the vertical (Saini et al, 1998), anteroposterior, and mediolateral directions (Eames et al, 1999; Vallis & McFadyen, 2005; Worden & Vallis, 2016).

These simplified CoM models for gait have not been validated among individuals who have experienced a stroke. Impaired gait is a common sign following stroke (Arene & Hidler, 2009), involving slower overall speed and reduced cadence (von Schroeder et al, 1995), as well as asymmetry in phases of the gait cycle (Wall & Turnbull, 1986; Patterson et al, 2008). Impaired walking patterns may limit the validity of applying simplified CoM models to gait analysis in people with stroke.

Determining a simplified CoM model is especially important with regards to ascertaining the level of stability an individual has while walking, as lateral instability has been previously linked to an increase in fall risk (Rogers & Mille, 2003). Individuals who have experienced a stroke may be at a greater risk of falls due to lateral instability during gait. Previous work has characterized the increased energetic demands during gait following stroke (Zampano et al, 1995), and with lateral stabilization during gait requiring increased energetic costs (Dean et al, 2007), this may negatively impact the ability to maintain lateral stability post-stroke. Margin of stability is widely used as a measure of stability (Hof et al, 2005), with vertical accuracy of a simplified model directly affecting the pendulum length of the margin of stability calculation, and mediolateral direction accuracy being important to effectively quantify the lateral distance and velocity to the base of support. Therefore, maintaining the accuracy in simplified models of COM is important, as traditional measures of gait (e.g. step width and step width variability) do not fully capture lateral stability compared to the margin of stability (Rosenblatt & Grabiner, 2010).

The purpose of the current study was to quantify the errors between the simplified CoM models and whole body CoM model in individuals with chronic stroke during overground gait. It was hypothesized that both HTP and PEL CoM models would exhibit similarly low CoM position errors in the anteroposterior and mediolateral directions, while only the PEL model would exhibit a low CoM position error in the vertical direction, as the whole-body CoM is often reported to lie within the pelvis segment, anterior to the second sacral vertebrae (Saunders et al, 1953). A secondary purpose was to examine the relationships

between CoM model error (relative to the full body model) and magnitude of gait impairment. We hypothesized that errors for simplified CoM models would be higher among those individuals with more severely-impaired gait.

2.0 Methods

2.1 Participants

We used data from 16 individuals with chronic stroke (mean 63.2 years old (standard deviation 8.9), mean 4.8 years post stroke (standard deviation 4.1)) who were enrolled in a randomized controlled trial (Mansfield et al, 2015). Twenty participants completed the gait assessment; however, marker occlusion during gait trials resulted in the exclusion of 4 individuals from the current analysis. Physical characteristics, as well as Chedoke-McMaster Stroke Assessment (CMSA; Gowland et al, 1993) leg scores, were collected (see Table 1). All procedures were approved by the institution's Research Ethics Board and participants provided written informed consent prior to participation.

2.2 Protocol & data capture

Participants performed 6 gait trials across a 6m walkway while wearing a safety harness attached to an overhead support. Participants were instructed by a research assistant to walk at their usual speed from one end of the walkway to the other. No other instructions were given in regards to arm movements or where to maintain visual focus, and all participants executed the gait trials wearing their usual flat-heeled closed-toe footwear. Of the 6 gait trials conducted, only trials 2–6 were included in analysis. The first gait trial was removed and considered an acclimatization trial for participants to familiarize themselves with the experimental setup and surroundings. Participants were offered rest breaks between gait capture trials, but no participant chose to take a break.

Kinematic data was collected via a Vicon motion capture system (Vicon MX40+, Vicon Motion Capture Systems Ltd., Oxford, UK) at 100Hz. A total of 52 single reflective markers and 8 rigid clusters were placed via adhesive tape on the body, with the following 23 single markers on anatomical landmarks used in this study: four markers on a headband placed on the cranium (positioned so markers were right and left front and back head), acromioclavicular joints, lateral elbows, ulnar styloid processes, xyphoid process, iliac crests, anterior and posterior superior iliac spines, greater trochanters, lateral epicondyle of the knee, and lateral malleoli. Four additional markers, on the heels and 2nd metatarsal joints, were used to track specific gait events (e.g. heel strike, toe off).

2.3 Data analyses

All kinematic data were first labelled in Vicon Nexus software (Vicon Nexus v.1.8.5, Vicon Motion Capture Systems Ltd., Oxford, UK), and then analyzed in Visual 3D v.5 (C-Motion, Germantown, USA). Kinematic data were interpolated and low pass filtered at 6 Hz via a 4th order, dual-pass Butterworth filter. CoM position was calculated using 3 models: (i) the whole-body model (WB); (ii) the head-trunk-pelvis model (HTP), and (iii) the pelvis only model (PEL). The WB model was a modified 13 segment Winter model (Winter et al, 1998) which included the head; upper and lower arms; upper, middle, and lower trunk; pelvis; and

upper and lower legs. The HTP model included just the head, upper trunk, and pelvis segments of the WB model, which has been previously used in gait research for healthy older adults (Worden & Vallis, 2016). The PEL model was made up of four markers (anterior and posterior superior iliac spines; Eames et al, 1999). Gait events were determined using a method developed previously (Ghoussayni et al, 2004). In brief, heel and toe marker displacement data were derived to acquire velocity. Heel strike (HS) was the time when the heel marker velocity in the anteroposterior direction fell below 0.1m/s, and toe off (TO) represents the time when the toe marker (2nd metatarsal) velocity in the anteroposterior direction rose above 0.1m/s. All gait events were visually inspected to ensure no false onsets or offsets were recorded.

Gait velocity was derived from the pelvis CoM displacement (Worden & Vallis, 2014) over one complete gait cycle (heel strike to heel strike on the same foot) in the middle of the trial to maximize marker visibility and data quality. Root mean squared error (RMSE) differences between the WB model and simplified models were calculated (RMSE HTP and RMSE PEL, respectively). RMSE was calculated based on the following equation:

$$RMSE = \sqrt{\frac{1}{n} \sum_{t=1}^n (WB_t - Simplified_t)^2} \quad (1)$$

RMSE values were compared over the two phases of the gait cycle, the paretic step and non-paretic step (Yang & Pai, 2014). Maximum absolute error (MAE) was also calculated over the same two phases of the gait cycle between the WB model and simplified models (MAE HTP and MAE PEL, respectively), using the following equation:

$$MAE = \frac{1}{N} \sum_{t=1}^N |WB_t - Simplified_t| \quad (2)$$

Calculating both RMSE and MAE provides a more complete picture when determining model accuracy and performance (Chai & Draxler, 2014); that is, including both allows the characterization of how well the simplified model performs on average (RMSE) as well as maxima and minima points (MAE).

Swing time and step length asymmetry were calculated as previously reported (Patterson et al, 2010), as the ratio of the paretic and non-paretic swing times and step lengths. The highest value time or length was the numerator to ensure that asymmetry values were ≥ 1 . Upper bound limits of healthy adult asymmetry values were used to determine which participants had asymmetrical swing time (95% confidence upper bound of 1.06) and/or step length (95% confidence upper bound of 1.08; Patterson et al, 2010; Table 1).

2.4 Statistical analyses

Paired sample t-tests were conducted to compare RMSE HTP to RMSE PEL for both the paretic step and non-paretic step. These t-tests were run across all three directions of motion separately: anteroposterior (x-direction), mediolateral (y-direction), and vertical (z-

direction). Following this, Spearman correlations were run for each RMSE value (broken down into direction and stepping limb) versus the CMSA leg score, swing time asymmetry ratio, step length asymmetry ratio, and gait velocity. This was done to determine how the severity of stroke affected the errors within the simplified models. All statistical analyses were completed using SAS 9.2 (SAS Institute, NC, USA) with the alpha set to 0.05, and all data are presented as mean with 95% confidence intervals (CI).

3.0 Results

3.1 RMSE & MAE differences

The paired-samples t-test revealed that RMSE HTP was significantly lower than RMSE PEL in the anteroposterior direction for both the paretic step ($t(15) = -16.85$, $p < 0.001$, 95% CI $[-4.47, -3.47]$ cm) and non-paretic step ($t(15) = -16.95$, $p < 0.001$, 95% CI $[-4.56, -3.54]$ cm; Table 2). Paired-samples t-test also revealed that MAE HTP was significantly lower than MAE PEL in the anteroposterior direction for both the paretic step ($t(15) = -16.60$, $p < 0.001$, 95% CI $[-0.049, -0.038]$ cm) and non-paretic step ($t(15) = -16.65$, $p < 0.001$, 95% CI $[-0.051, -0.039]$ cm; Table 3).

Additionally, RMSE HTP was significantly higher than RMSE PEL in the vertical direction for both the paretic step ($t(15) = 63.42$, $p < 0.001$, 95% CI $[0.26, 0.28]$ cm) and non-paretic step ($t(15) = 61.52$, $p < 0.001$, 95% CI $[0.26, 0.28]$ cm; Table 2). This was also true for MAE HTP, as it was significantly higher than MAE PEL in the vertical direction for both the paretic step ($t(15) = 61.26$, $p < 0.001$, 95% CI $[0.26, 0.28]$ cm) and non-paretic step ($t(15) = 59.32$, $p < 0.001$, 95% CI $[0.26, 0.28]$ cm; Table 3).

There were no significant differences in RMSE between the two models in the mediolateral direction for both the paretic step ($t(15) = 0.47$, $p = 0.64$, 95% CI $[-0.0022, 0.0035]$ cm) and non-paretic step ($t(15) = -0.19$, $p = 0.85$, 95% CI $[-0.0018, 0.0015]$ cm; Table 2). As with the RMSE results, no significant differences were observed in MAE between the two models in the mediolateral direction for either the paretic step ($t(15) = 0.42$, $p = 0.68$, 95% CI $[-0.0026, 0.0039]$ cm) or non-paretic step ($t(15) = -0.82$, $p < 0.001$, 95% CI $[-0.0025, 0.0011]$ cm; Table 3).

3.2 Correlation to measures of stroke severity

There were significant negative correlations observed in the mediolateral direction for the paretic step between RMSE HTP and CMSA leg score ($r = -0.70$, $p = 0.0046$; Table 4), and MAE HTP and CMSA leg score ($r = -0.78$, $p < 0.001$; Table 5). In addition to this, positive correlations were observed in the mediolateral direction for the paretic step between RMSE HTP and step length asymmetry ($r = 0.57$, $p = 0.02$; Table 4), and MAE HTP and step length asymmetry ($r = 0.64$, $p = 0.007$; Table 5).

A significant negative correlation was also observed in the mediolateral direction for the paretic step between RMSE HTP and gait velocity ($r = -0.79$, $p < 0.001$; Table 4), while significant negative correlations were observed between MAE HTP and gait velocity in the mediolateral direction for both the paretic step ($r = -0.86$, $p < 0.001$) and non-paretic step ($r = -0.55$, $p = 0.027$; Table 5). No other significant correlations were observed for RMSE or

MAE HTP, nor were any significant correlations observed across measures of stroke severity and either RMSE or MAE PEL.

4.0 Discussion

The current study investigated whether a simplified model to characterize CoM motion during gait in individuals with chronic stroke could effectively provide CoM information traditionally captured by a multiple segment, full body model. It was hypothesized that both simplified models would have similarly low displacement error compared to the WB model in the anteroposterior and mediolateral directions, while only the PEL simplified model would exhibit low displacement error in the vertical direction. In partial support of this hypothesis, RMSE results demonstrated that both models were similar to the WB model (1cm error) in the mediolateral direction, while PEL was more similar vertically and HTP was more similar in the anteroposterior direction. In the anteroposterior direction, the average error of the PEL relative to the WB model was 5cm. In addition, MAE findings followed the same statistical pattern as RMSE results, with all MAE findings being within 1cm of all RMSE findings. These findings were also similar to previous work that reported RMSE differences of 16.8 cm in the anteroposterior direction, 1.8 cm in the mediolateral direction, and 3.1cm vertically for a sacral model based on a single marker on the sacrum (Yang & Pai, 2014). Anteroposterior differences observed in our findings could be due to forward trunk pitch (forward rotation about the mediolateral axis) during gait, as the addition of the upper trunk and head segments in the HTP model (compared to the PEL model) would result in further anteroposterior displacement of the CoM. Previous work has demonstrated both trunk pitch and trunk pitch velocity increase with age (Goutier et al, 2010), while transverse thoracic range of motion has been shown to increase by 15% in chronic stroke individuals compared to age-matched healthy controls (Hacmon et al, 2012). Additionally, forward trunk pitch could help to compensate for potentially decreased propulsive force from the paretic limb during gait (Balasubramanian et al, 2007). Thus, any potential trunk pitch occurring during the gait cycle of our chronic stroke participants would be captured by both the WB and HTP models but not the PEL model, therefore potentially resulting in the differences observed between the two simplified models in the anteroposterior direction (i.e., higher errors for the PEL model in the anteroposterior direction).

PEL demonstrated a low displacement error in the vertical direction compared to HTP (RMSE of 1.7 cm compared to 28.9 cm on average, respectively). This result was expected, as the HTP model takes into account a large proportion of the upper body CoM but none of the lower body CoM, while the PEL model would be similar in height to where the WB CoM is thought to lie within the body (Saunders et al, 1953). Furthermore, this result is supported by previous simplified models in healthy participants, as both a PEL model (Saini et al, 1998) or even simplified models relying solely on a sacral marker can provide similar vertical displacement measures during gait (Thirunarayan et al, 1996).

The minimal error of the PEL model in both the mediolateral and vertical directions, compared to the HTP model in just the mediolateral direction, is especially important when considering dynamic stability analyses. Stability measured by margin of stability, as

described by Hof et al. (2005), is being increasingly used to measure and characterize stability during gait, both in healthy individuals (Marone et al, 2014; Kao et al, 2015; Vieira et al, 2016) and individuals with stroke (Kao et al, 2014). The minimal error of the PEL model in the vertical direction means that the difference between pendulum lengths calculated from a WB model or a simplified PEL model would be similar, as the pendulum length is the distance between the CoM and the ankle in the vertical direction (Hof et al, 2005). This low vertical error, coupled with the minimal error demonstrated by the PEL model in the mediolateral direction, is important as previous work has demonstrated a link between mediolateral instability and falls (Rogers & Mille, 2003; Hilliard et al, 2008). Furthermore, both the minimal RMSE and MAE in the vertical and mediolateral directions for the PEL model is important, as margin of stability is often computed at specific points of the gait cycle and not as an average signal throughout the gait cycle. Therefore, the simplified PEL model would apply to estimating dynamic stability during gait in individuals with chronic stroke.

A secondary objective of the current study was to investigate how the severity of disability for individuals with chronic stroke influenced the level of error within the simplified models. The CMSA leg score negatively correlated in the mediolateral direction with only RMSE HTP and MAE HTP during the paretic step, while no significant correlation between CMSA leg and RMSE or MAE PEL being observed. Additionally, RMSE HTP and MAE HTP in the mediolateral direction were positively correlated with step length asymmetry for the paretic step, while RMSE HTP was negatively correlated with gait velocity in the mediolateral direction for the paretic step and MAE HTP was negatively correlated with gait velocity in the mediolateral direction for both steps. No significant correlations were found between step length asymmetry or gait velocity for either RMSE or MAE PEL. The relative lack of correlation between simplified model error and gait impairment provides support to how the minimal error found in these models are robust, especially the PEL model, in capturing CoM during gait of individuals with chronic stroke with a wide range of gait impairments. Gait asymmetry has been thought to result in increased metabolic cost in stroke individuals (Brouwer et al, 2009), and is related to a high number of falls post stroke (Lewek et al, 2014). Yet with no correlation of asymmetry to the PEL model (a model which had 1cm RMSE on average when compared to the WB model), our results provide support for the ability to apply a simplified CoM model (e.g. the PEL model) to quantify CoM displacement during gait of individuals with chronic stroke.

It is important to note the limitations in our study. Firstly, only participants with chronic stroke were included in the current study. Potentially, greater negative effects during the acute or sub-acute stages of stroke recovery may result in more dramatic changes to gait velocity (derived from CoM motion), as gait velocity has been shown to increase from sub-acute to chronic stages of stroke (Goldie et al, 1996). Thus, it is still unknown whether a simplified model would be appropriate to apply to individuals in the acute or subacute stages of stroke recovery. Secondly, our whole-body model was based off the Winter model (Winter et al, 1998), and did not take into account specific subject anthropometric variability (Chen et al, 2011). This could theoretically induce errors in the estimate of the CoM, which may have impacted accuracy of the Winter model. Lastly, it is important to note that there is not necessarily an 'acceptable' error of CoM models during overground gait. Acceptable values

of RMSE and MAE may vary with the application (e.g. examining total distance CoM traveled during gait vs. how well the CoM travels on a specified trajectory towards a target during gait), and thus it is ultimately up to the reader to determine if the error in the model is acceptable for their specific application (See Fig 1).

5.0 Conclusion

In summary, the current study sought to determine the level of error and thus potential utility of simplified CoM models to analyze gait in individuals with chronic stroke. Results support the conclusion that the HTP model provides similar displacement to the WB model in the anteroposterior direction, while the PEL model provides similar displacement in the mediolateral and vertical directions. Furthermore, only the HTP model correlated with any measures of stroke severity, while the PEL model did not. The minimal error within the PEL model has especially important implications, as its small (<2 cm) error in the mediolateral and vertical directions would hypothetically make it suitable for lateral dynamic stability analyses (Hof et al, 2005), thus allowing clinics to use complex analyses with a simplified model to increase the efficiency of setup resulting in potential decreases in assessment cost and time.

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References

- Arene N, Hidler J. Understanding motor impairment in the paretic lower limb after a stroke: a review of the literature. *Topics in stroke rehabilitation*. 2009; 16(5):346–356. [PubMed: 19903653]
- Balasubramanian CK, Bowden MG, Neptune RR, Kautz SA. Relationship between step length asymmetry and walking performance in subjects with chronic hemiparesis. *Archives of physical medicine and rehabilitation*. 2007; 88(1):43–49. [PubMed: 17207674]
- Brouwer B, Parvataneni K, Olney SJ. A comparison of gait biomechanics and metabolic requirements of overground and treadmill walking in people with stroke. *Clinical biomechanics*. 2009; 24(9):729–734. [PubMed: 19664866]
- Chai T, Draxler RR. Root mean square error (RMSE) or mean absolute error (MAE)?—Arguments against avoiding RMSE in the literature. *Geoscientific model development*. 2014; 7(3):1247–1250.
- Chen SC, Hsieh HJ, Lu TW, Tseng CH. A method for estimating subject-specific body segment inertial parameters in human movement analysis. *Gait & posture*. 2011; 33(4):695–700. [PubMed: 21458993]
- Dean JC, Alexander NB, Kuo AD. The effect of lateral stabilization on walking in young and old adults. *IEEE transactions on biomedical engineering*. 2007; 54(11):1919–1926. [PubMed: 18018687]

- Eames MHA, Cosgrove A, Baker R. Comparing methods of estimating the total body centre of mass in three-dimensions in normal and pathological gaits. *Human movement science*. 1999; 18(5):637–646.
- Eng JJ, Winter DA. Estimations of the horizontal displacement of the total body centre of mass: considerations during standing activities. *Gait & posture*. 1993; 1(3):141–144.
- Ghoussayni S, Stevens C, Durham S, Ewins D. Assessment and validation of a simple automated method for the detection of gait events and intervals. *Gait & posture*. 2004; 20(3):266–272. [PubMed: 15531173]
- Goldie PA, Matyas TA, Evans OM. Deficit and change in gait velocity during rehabilitation after stroke. *Archives of physical medicine and rehabilitation*. 1996; 77(10):1074–1082. [PubMed: 8857890]
- Gowland C, Stratford P, Ward M, Moreland J, Torresin W, Van Hullenaar S, ... Plews N. Measuring physical impairment and disability with the Chedoke-McMaster Stroke Assessment. *Stroke*. 1993; 24(1):58–63. [PubMed: 8418551]
- Goutier KM, Jansen SL, Horlings CG, Küng UM, Allum JH. The influence of walking speed and gender on trunk sway for the healthy young and older adults. *Age and ageing*. 2010:afq066.
- Gutierrez-Farewik EM, Bartonek Å, Saraste H. Comparison and evaluation of two common methods to measure center of mass displacement in three dimensions during gait. *Human movement science*. 2006; 25(2):238–256. [PubMed: 16458379]
- Hacmon RR, Krasovsky T, Lamontagne A, Levin MF. Deficits in intersegmental trunk coordination during walking are related to clinical balance and gait function in chronic stroke. *Journal of neurologic physical therapy*. 2012; 36(4):173–181. [PubMed: 23095903]
- Hilliard MJ, Martinez KM, Janssen I, Edwards B, Mille ML, Zhang Y, Rogers MW. Lateral balance factors predict future falls in community-living older adults. *Archives of physical medicine and rehabilitation*. 2008; 89(9):1708–1713. [PubMed: 18760155]
- Hof AL, Gazendam MGJ, Sinke WE. The condition for dynamic stability. *Journal of biomechanics*. 2005; 38(1):1–8. [PubMed: 15519333]
- Kao PC, Dingwell JB, Higginson JS, Binder-Macleod S. Dynamic instability during post-stroke hemiparetic walking. *Gait & posture*. 2014; 40(3):457–463. [PubMed: 24931112]
- Kao PC, Higginson CI, Seymour K, Kamerdze M, Higginson JS. Walking stability during cell phone use in healthy adults. *Gait & posture*. 2015; 41(4):947–953. [PubMed: 25890490]
- Lewek MD, Bradley CE, Wutzke CJ, Zinder SM. The relationship between spatiotemporal gait asymmetry and balance in individuals with chronic stroke. *Journal of applied biomechanics*. 2014; 30(1):31–6. [PubMed: 23677889]
- Mansfield A, Aquil A, Centen A, Danells CJ, DePaul VG, Knorr S, ... Mochizuki G. Perturbation training to promote safe independent mobility post-stroke: study protocol for a randomized controlled trial. *BMC neurology*. 2015; 15(1):1. [PubMed: 25595849]
- Marone JR, Patel PB, Hurt CP, Grabiner MD. Frontal plane margin of stability is increased during texting while walking. *Gait & posture*. 2014; 40(1):243–246. [PubMed: 24798610]
- Patterson KK, Parafianowicz I, Danells CJ, Closson V, Verrier MC, Staines WR, ... McIlroy WE. Gait asymmetry in community-ambulating stroke survivors. *Archives of physical medicine and rehabilitation*. 2008; 89(2):304–310. [PubMed: 18226655]
- Patterson KK, Gage WH, Brooks D, Black SE, McIlroy WE. Evaluation of gait symmetry after stroke: a comparison of current methods and recommendations for standardization. *Gait & posture*. 2010; 31(2):241–246. [PubMed: 19932621]
- Rogers MW, Mille ML. Lateral stability and falls in older people. *Exercise and sport sciences reviews*. 2003; 31(4):182–187. [PubMed: 14571957]
- Rosenblatt NJ, Grabiner MD. Measures of frontal plane stability during treadmill and overground walking. *Gait & posture*. 2010; 31(3):380–384. [PubMed: 20129786]
- Saini M, Kerrigan DC, Thirunarayan MA, Duff-Raffaele M. The vertical displacement of the center of mass during walking: a comparison of four measurement methods. *Journal of biomechanical engineering*. 1998; 120(1):133–139. [PubMed: 9675692]
- Saunders JB, Inman VT, Eberhart HD. The major determinants in normal and pathological gait. *Journal of bone & joint surgery - American*. 1953; 35(3):543–558.

- Thirunarayan MA, Kerrigan DC, Rabuffetti M, Della Croce U, Saini M. Comparison of three methods for estimating vertical displacement of center of mass during level walking in patients. *Gait & posture*. 1996; 4(4):306–314.
- Vallis LA, McFadyen BJ. Children use different anticipatory control strategies than adults to circumvent an obstacle in the travel path. *Experimental brain research*. 2005; 167(1):119–127. [PubMed: 16177831]
- Winter DA, Patla AE, Prince F, Ishac M, Gielo-Perczak K. Stiffness control of balance in quiet standing. *Journal of neurophysiology*. 1998; 80(3):1211–1221. [PubMed: 9744933]
- Worden TA, Vallis LA. Stability control during the performance of a simultaneous obstacle avoidance and auditory Stroop task. *Experimental brain research*. 2016; 234(2):387–396. [PubMed: 26487180]
- Vieira MF, e Souza GSDS, Lehnen GC, Rodrigues FB, Andrade AO. Effects of general fatigue induced by incremental maximal exercise test on gait stability and variability of healthy young subjects. *Journal of electromyography and kinesiology*. 2016; 30:161–167. [PubMed: 27451361]
- von Schroeder HP, Coutts RD, Lyden PD, Nickel VL. Gait parameters following stroke: a practical assessment. *Journal of rehabilitation research and development*. 1995; 32(1):25. [PubMed: 7760264]
- Wall JC, Turnbull GI. Gait asymmetries in residual hemiplegia. *Archives of physical medicine and rehabilitation*. 1986; 67(8):550–553. [PubMed: 3741082]
- Worden TA, Vallis LA. Concurrent performance of a cognitive and dynamic obstacle avoidance task: influence of dual-task training. *Journal of motor behavior*. 2014; 46(5):357–368. [PubMed: 24914575]
- Yang F, Pai YC. Can sacral marker approximate center of mass during gait and slip-fall recovery among community-dwelling older adults? *Journal of biomechanics*. 2014; 47(16):3807–3812. [PubMed: 25468302]
- Zamparo P, Francescato MP, Luca G, Lovati L, Prampera PE. The energy cost of level walking in patients with hemiplegia. *Scandinavian journal of medicine & science in sports*. 1995; 5(6):348–352. [PubMed: 8775719]

Figure 1.

A representative trial of center of mass (CoM) displacement over the course of one gait cycle, defined as right heel strike (RHS), to left heel strike (LHS), to the next RHS. The first 50% of the gait cycle (RHS to LHS) represents the paretic step, while the second 50% of the gait cycle (LHS to RHS) represents the non-paretic step. The **A**) anteroposterior (X, + forward), **B**) mediolateral (Y, + left), and **C**) vertical (Z, + up) COM displacements are shown for the whole-body model (WB), head-trunk-pelvis model (HTP), and the pelvis only model (PEL).

Table 1

Physical characteristics, CMSA leg score, gait velocity, and asymmetry ratio for all participants. CMSA leg score is out of 7, with a higher number relating to healthier function. Data at the bottom of the table is presented in mean (standard deviation) where applicable.

	Age (yrs)	Sex (F/M)	Height (cm)	Weight (kg)	Time Post Stroke (yrs)	Affected Side	CMSA Leg Score	Gait Velocity (m/s)	Swing Time Asymmetry	Step Length Asymmetry
<i>Participant</i>										
1	59	M	161	69.2	11.8	Left	5	0.58	1.33 ^a	1.98 ^a
2	64	M	181	93.9	2.2	Right	3	0.31	1.11 ^a	1.51 ^a
3	61	F	166	71.6	4.3	Left	6	1.09	1.13 ^a	1.11 ^a
4	65	M	179	77.5	0.7	Right	7	0.90	1.04	1.13 ^a
5	68	M	166	76.5	1.3	Left	5	0.26	1.11 ^a	1.27 ^a
6	72	F	167	72.5	1.2	Right	5	0.86	1.26 ^a	1.05
7	51	F	163	87.9	3.4	Left	7	0.88	1.05	1.04
8	65	M	178	69.7	1.6	Right	7	1.10	1.01	1.03
9	43	F	149	64.6	6.2	Left	5	0.68	1.26 ^a	1.14 ^a
10	73	M	178	74.1	9.0	Left	6	0.93	1.14 ^a	1.18 ^a
11	61	M	171	64.1	4.2	Left	5	1.01	1.34 ^a	1.15 ^a
12	72	M	164	74.0	6.8	Left	7	1.26	1.10 ^v	1.06
13	60	M	172	82.0	14.4	Left	5	0.49	1.03	1.28 ^a
14	65	M	171	84.4	2.1	Left	5	0.69	1.12 ^a	1.34 ^a
15	78	M	168	75.6	0.5	Left	5	0.69	1.23 ^a	1.22 ^a
16	54	M	168	76.0	5.1	Right	5	0.59	1.20 ^a	1.32 ^a
Means	63.2 (8.9)	4F, 12M	167 (11)	75.9 (8.0)	4.8 (4.1)	11L, 5R	5.5 (1.1)	0.77 (0.28)	1.16 (0.10)	1.24 (0.24)

^a represents an asymmetry ratio above the upper bound (95% confidence interval) of healthy adult ratios outlined by Patterson et al. (2010). The upper bound for swing time and step length asymmetry is 1.06 and 1.08, respectively.

Table 2

T-test results comparing the root of the mean squared error (RMSE) difference of displacements between the whole-body model and the head-trunk-pelvis model (RMSE HTP), versus the RMSE difference of displacements between whole-body model and the pelvis only model (RMSE PEL). Data is separated into direction of motion (x: anteroposterior, y: mediolateral, z: vertical) and paretic step (PS) or non-paretic step (NS), in mean [95% CI]. Bold results indicate a significant difference between RMSE HTP and RMSE PEL ($p < 0.05$).

	RMSE HTP (cm)	RMSE PEL (cm)	Significance
<i>Anteroposterior</i>			
PS	1.03 [0.90, 1.15]	4.97 [4.74, 5.20]	$p < 0.001$
NS	1.01 [0.89, 1.13]	5.03 [4.77, 5.29]	$p < 0.001$
<i>Mediolateral</i>			
PS	1.02 [0.88, 1.17]	0.99 [0.87, 1.10]	$p = 0.64$
NS	0.80 [0.70, 0.90]	0.80 [0.71, 0.89]	$p = 0.85$
<i>Vertical</i>			
PS	28.89 [28.49, 29.29]	1.71 [1.43, 2.00]	$p < 0.001$
NS	28.80 [28.40, 29.19]	1.63 [1.35, 1.91]	$p < 0.001$

Table 3

T-test results comparing the maximum absolute error (MAE) difference of displacements between the whole-body model and the head-trunk-pelvis model (MAE HTP), versus the RMSE difference of displacements between whole-body model and the pelvis only model (MAE PEL). Data is separated into direction of motion (x: anteroposterior, y: mediolateral, z: vertical) and paretic step (PS) or non-paretic step (NS), in mean [95% CI]. Bold results indicate a significant difference between MAE HTP and MAE PEL ($p < 0.05$).

	MAE HTP (cm)	MAE PEL (cm)	Significance
<i>Anteroposterior</i>			
PS	1.61 [1.44, 1.79]	5.95 [5.69, 6.21]	$p < 0.001$
NS	1.47 [1.30, 1.63]	5.94 [5.66, 6.22]	$p < 0.001$
<i>Mediolateral</i>			
PS	1.48 [1.29, 1.67]	1.44 [1.30, 1.59]	$p = 0.68$
NS	1.12 [1.05, 1.30]	1.24 [1.12, 1.36]	$p = 0.42$
<i>Vertical</i>			
PS	29.14 [28.74, 29.54]	2.04 [1.74, 2.33]	$p < 0.001$
NS	29.02 [28.63, 29.41]	1.97 [1.68, 2.27]	$p < 0.001$

Table 4

Correlation results of Chedoke-McMaster Stroke Assessment (CMSA) leg score, asymmetry (swing time and step length), and gait velocity compared to the root of the mean squared error (RMSE) difference of displacements between the whole-body model and the head-trunk-pelvis model (RMSE HTP) and the RMSE difference of displacements between the whole-body model and the pelvis only model (RMSE PEL). Data is separated into direction of motion (x: anteroposterior, y: mediolateral, z: vertical) and step limb (paretic (PS) or non-paretic (NS)), with the Spearman coefficient followed by statistical significance in brackets below. Data is presented as mean (p value) with bold results indicating a significant finding ($p < 0.05$).

	CMSA Leg Score	Swing Time Asymmetry	Step Length Asymmetry	Gait Velocity
RMSE HTP (m)				
PSx	-0.10 (0.70)	-0.047 (0.86)	0.25 (0.37)	-0.29 (0.27)
NSx	0.21 (0.42)	-0.23 (0.39)	-0.044 (0.87)	-0.056 (0.84)
PSy	-0.70 (0.0046)	0.21 (0.43)	0.57 (0.022)	-0.79 (<0.001)
NSy	-0.35 (0.19)	-0.050 (0.85)	0.35 (0.18)	-0.41 (0.11)
PSz	-0.041 (0.88)	-0.38 (0.14)	0.19 (0.48)	-0.032 (0.91)
NSz	-0.041 (0.88)	-0.41 (0.11)	0.20 (0.46)	-0.068 (0.80)
RMSE PEL (m)				
PSx	0.092 (0.74)	-0.36 (0.17)	-0.085 (0.75)	-0.11 (0.70)
NSx	0.19 (0.48)	-0.36 (0.17)	-0.16 (0.56)	0.074 (0.79)
PSy	-0.13 (0.64)	-0.026 (0.92)	0.33 (0.21)	-0.39 (0.13)
NSy	-0.29 (0.27)	-0.047 (0.86)	0.17 (0.52)	-0.39 (0.13)
PSz	-0.095 (0.73)	0.032 (0.91)	0.094 (0.73)	0.002 (0.99)
NSz	0.14 (0.62)	0.13 (0.64)	0.088 (0.76)	-0.029 (0.91)

Table 5

Correlation results of Chedoke-McMaster Stroke Assessment (CMSA) leg score, asymmetry (swing time and step length), and gait velocity compared to the maximum absolute error (MAE) difference of displacements between the whole-body model and the head-trunk-pelvis model (MAE HTP) and the MAE difference of displacements between the whole-body model and the pelvis only model (MAE PEL). Data is separated into direction of motion (x: anteroposterior, y: mediolateral, z: vertical) and step limb (paretic (PS) or non-paretic (NS)), with the Spearman coefficient followed by statistical significance in brackets below. Data is presented as mean (*p* value) with bold results indicating a significant finding ($p < 0.05$).

	CMSA Leg Score	Swing Time Asymmetry	Step Length Asymmetry	Gait Velocity
MAE HTP (m)				
PSx	-0.19 (0.47)	0.015 (0.96)	0.30 (0.25)	-0.35 (0.19)
NSx	0.27 (0.32)	-0.26 (0.34)	0.059 (0.83)	-0.038 (0.89)
PSy	-0.78 (<0.001)	0.28 (0.29)	0.64 (0.007)	-0.86 (<0.001)
NSy	-0.45 (0.084)	-0.018 (0.95)	0.44 (0.085)	-0.55 (0.027)
PSz	-0.015 (0.96)	-0.37 (0.16)	0.16 (0.56)	0.0029 (0.99)
NSz	-0.041 (0.88)	-0.41 (0.11)	0.20 (0.46)	-0.068 (0.80)
MAE PEL (m)				
PSx	0.052 (0.85)	-0.28 (0.29)	-0.074 (0.79)	-0.15 (0.57)
NSx	0.083 (0.76)	-0.42 (0.10)	-0.12 (0.66)	-0.076 (0.79)
PSy	-0.20 (0.47)	0.14 (0.60)	0.28 (0.29)	-0.43 (0.097)
NSy	-0.29 (0.27)	-0.015 (0.96)	0.21 (0.54)	-0.44 (0.092)
PSz	-0.051 (0.85)	0.041 (0.88)	0.062 (0.82)	0.018 (0.95)
NSz	-0.15 (0.57)	0.085 (0.75)	0.11 (0.70)	-0.053 (0.85)