Gait & Posture 39 (2014) 129–134

Contents lists available at SciVerse ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Relationships between frontal-plane angular momentum and clinical balance measures during post-stroke hemiparetic walking

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ARTICLE INFO

Article history:
Received 17 August 2012
Received in revised form 23 May 2013
Accepted 11 June 2013

Keywords:
Stability
Gait
Dynamic balance
Hemiparesis
Clinical measures

ABSTRACT

Stroke has significant impact on dynamic balance during locomotion, with a 73% incidence rate for falls post-stroke. Current clinical assessments often rely on tasks and/or questionnaires that relate to the statistical probability of falls and provide little insight into the mechanisms that impair dynamic balance. Current quantitative measures that assess medial–lateral balance performance do not consider the angular motion of the body, which can be particularly impaired after stroke. Current control methods in bipedal robotics rely on the regulation of angular momentum (H) to maintain dynamic balance during locomotion. This study tests whether frontal-plane H is significantly correlated to clinical balance tests that could be used to provide a detailed assessment of medial–lateral balance impairments in hemiparetic gait. H was measured in post-stroke ($n = 48$) and control ($n = 20$) subjects. Post-stroke there were significant negative relationships between the change in frontal-plane H during paretic single-leg stance and two clinical tests: the Dynamic Gait Index (DGI) ($r = -0.57$, $p < 0.001$) and the Berg Balance Scale (BBS) ($r = -0.54$, $p < 0.001$). Control subjects showed timely regulation of frontal-plane H during the first half of single-leg stance, with the level of regulation depending on the initial magnitude. In contrast, the post-stroke subjects who made poorer adjustments to frontal-plane H during initial paretic leg single stance exhibited lower DGI and BBS scores ($r = 0.45$, $p = 0.003$). We conclude that H is a promising balance indicator during steady-state hemiparetic walking and that paretic single-leg stance is a period with higher instability for stroke patients.

Published by Elsevier B.V.

1. Introduction

There is a 73% incidence of falls among individuals post-stroke, with 37% that fall sustaining injuries requiring medical treatment [1] that ultimately leads to a significant decrease in activity due to the fear of falling [2,3]. Additionally, balance needs are particular to the task being performed [4] and static balance measures may not reflect the complexities of dynamic balance. Studies of dynamic walking have suggested that active control is needed to regulate medial–lateral balance, but not sagittal-plane balance [5]. Quantitative measures of medial–lateral balance during walking have been developed based on the concept of the "extrapolated center-of-mass" [6,7], but these have not been applied to the post-stroke population. Further, they assume that the angular acceleration of the trunk can be neglected, which may not be justified in the hemiparetic population. Thus, an important step toward understanding the increased falls risks in persons post-stroke may be developing a quantitative measure of medial–lateral dynamic balance performance during hemiparetic walking that incorporates the angular motion of the trunk.

A number of clinical balance measures have been proposed that attempt to assess balance [8–10]. However, Mancini et al. summarized the most used balance assessment tools and noted that while most successfully identify a balance problem, they typically fail to direct clinical rehabilitation toward solving the underlying balance disorders [11]. This highlights the need for a quantitative measure that can be linked to underlying mechanisms.

Two of the most common clinical measures to assess balance ability are the Berg Balance Scale (BBS) [12] and the Dynamic Gait Index (DGI) [13]. The BBS, which tests mostly static balance, has been shown to have a sensitivity of 91% and specificity of 82% to falls when coupling the test with a self-reported history of imbalance [14]. The DGI, which tests dynamic balance during walking tasks but allows assistive devices, has been described as a useful tool for evaluating balance with reported sensitivity of 77% and specificity of 90% to falls in persons with vestibular deficits [13]. Although these clinical measures provide good reliability,

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0966-6362/$ – see front matter. Published by Elsevier B.V.
http://dx.doi.org/10.1016/j.gaitpost.2013.06.008

they are ordinal rating scores that are observational and not quantitative. Thus, they cannot provide a quantitative step-by-step measure by which to assess balance performance during walking and reveal little about the underlying mechanisms.

In contrast to ordinal assessment scales, continuous, quantitative measures can potentially provide insight into the mechanisms of dynamic balance regulation. Developments in bipedal robotics have used whole-body angular momentum (H) in trajectory planning to maintain dynamic balance in bipedal gait using the concept of Zero Moment Point (ZMP) [15–17]. The ZMP principle seeks to reduce the external moments about the center-of-mass (COM) to zero so that the whole-body angular orientation does not change from its initial condition. Studies in human gait suggest that H is highly regulated by the central nervous system and kept at a low value [18,19]. However, since H is known to fluctuate in human walking, efforts associated with balance might be better represented by the change in the angular momentum, or more specifically the time derivative of H ($\dot{H}$), which is equal to the sum of the external moments acting about the COM. Thus, frontal-plane $\dot{H}$ is directly related to medial–lateral and vertical ground reaction forces, and hence medial–lateral balance control.

This study will test walking on an instrumented treadmill as opposed to overground, despite some previous concerns in the literature that treadmill walking has reduced asymmetry compared to overground walking in post-stroke individuals [20,21]. However, Kautz et al. [22] recently showed in a large study of 56 persons with hemiparesis that immediate improvements in symmetry (either temporal or spatial) do not occur when subjects walk on a treadmill without support (e.g., holding a hand rail or with body weight support from a harness). Instead, treadmill walking increased step length asymmetry. Treadmill walking appeared to provide a challenge and exacerbated hemiparetic participants' existing motor control deficits because the differences observed between treadmill and overground walking did not influence key kinematic and EMG measures of motor control deficits [22]. Thus, we believe that treadmill walking is a valid method for studying control of angular momentum during walking.

This study aims to determine whether frontal-plane $\dot{H}$ differs between subjects with hemiparesis and speed-matched controls and whether frontal-plane $\dot{H}$ can serve as a quantitative measure of dynamic balance performance during walking. Specifically, we seek to determine the regions of the gait cycle where significant $\dot{H}$ differences occur between control and hemiparetic subjects. The relation between clinical balance assessment and $\dot{H}$ in these regions will also be tested. It is expected that, based on the robotics literature, larger values of $\dot{H}$ (i.e., large changes in H) will relate to poorer dynamic balance (and thus lower scores in clinical balance assessments). Since both BBS and DGI are reasonable predictors of falls [10,12–14,23–27], if this relationship proves true in hemiparetic walking, $\dot{H}$ may provide a valuable quantitative measure to assess balance disorders during walking. To help interpret any observed differences, we also quantified the variability in $\dot{H}$ and lateral foot placement.

2. Methods

2.1. Experimental procedure and demographics

Forty-eight subjects with post-stroke hemiparesis (29 males; age = 58.3 ± 12.0 years; 5.1 ± 3.1 years post-stroke) walked at their comfortable self-selected treadmill walking speed on a split belt instrumented treadmill (Techmachine, Andrezieux Boutheon, France) for multiple trials of 30 s. The average number of steps per subject was 18.4 steps, with a range from 6 to 28 steps. They were also asked to walk over an instrumented mat (GaitRite, Havertown, PA) to determine their self-selected over-ground walking speed. BBS and DGI data were collected. The subjects were divided into three groups based on self-selected walking speeds on the treadmill to establish the effect of gait speed on the H. The slow group were subjects who walked slower than 0.4 m/s on the treadmill (n = 26). The medium speed group walked between 0.4 m/s and 0.8 m/s on the treadmill (n = 15) while the fast group walked between 0.8 m/s and 1.2 m/s (n = 7). Twenty control subjects (4 males; age = 65.1±10.4 years) also walked on the treadmill for 30 s at each of three different speeds (0.3 m/s, 0.6 m/s, and 0.9 m/s). The averages of the control group at 0.3 m/s, 0.6 m/s, and 0.9 m/s were used as speed matched controls for the slow, medium and fast post-stroke groups, respectively.

2.2. Data collection

Kinematic data were collected at 100 samples/sec by a 12 camera motion capture system (VICON, Los Angeles, USA). Reflective markers were placed in rigid clusters on 13 segments (pelvis, head, trunk and each foot, shank, thigh, upper arm, and lower arm). A custom model template was created in Visual3D (C-Motion, Germantown, MD, USA) and applied to the markers to determine body segment kinematics.

2.3. Data analysis

The gait cycle was divided into six regions. Region 1 begins with paretic foot strike and ends with non-paretic foot off (first double support). Regions 2 and 3 are defined as the first and second halves of paretic single-leg stance, respectively. Region 4 is the second double support and Regions 5 and 6 are the first and second halves of paretic swing, respectively.

Whole body angular momentum (H) was calculated as the summation of the H of each segment about the COM. The time derivative of H ($\dot{H}$) was then calculated and normalized by each subject's weight and instantaneous COM vertical height to provide a non-dimensional measure. The mean $\dot{H}$ for each region (normalized by the time duration of the corresponding region) was considered the change in H for that region.

We also assessed two measures of variability for each subject: the $\dot{H}$ step-to-step variability (standard deviation of $\dot{H}$ over the paretic single stance phase for all steps of a walking trial) and the foot placement variability (standard deviation of the lateral distance between the paretic foot and the COM at the onset of region 2 for all steps of a walking trial).

2.4. Statistical analysis

Spearman's correlation was used to correlate the H measures with the clinical measures. Pearson correlations were used to correlate H measures at two different points in the gait cycle on a step-to-step basis. Additionally, Spearman's correlation was used to correlate the resulting correlation coefficients to DGI and BBS scores. Significant differences between groups were calculated using a two-tailed t-test. In all cases, significance was set at $p < 0.05$.

3. Results

3.1. Berg balance scale and dynamic gait index

Of the 48 subjects tested, 9 were considered at risk of falling according to the BBS (score < 45) [28] and 30 by the DGI (score < 19) [29]. All subjects classified as potential fallers by the BBS test were also classified as potential fallers by DGI. For simplicity, we will henceforth refer to potential fallers as "fallers" and those above the threshold score as "non-fallers". Significant