ARTICLE IN PRESS

Human Movement Science xxx (xxxx) xxx–xxx

Contents lists available at ScienceDirect

Human Movement Science

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

Full Length Article

Relationship between margin of stability and deviations in spatiotemporal gait features in healthy young adults

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ARTICLE INFO

Keywords:
Gait
Accidental Falls
Kinematics
Motor Control

ABSTRACT

Increased variability of spatio-temporal features while walking is related to increased risk of falls. It is thought that variability in foot placement and timing reflects responses to mechanical instability while walking. The purpose of this study was to determine whether 'extreme' values of step length, width and time follow transient periods of low mechanical stability during the single support phase of gait in healthy young adults. We conducted secondary analysis of a portion of an existing dataset. Eleven healthy adults walked on an instrumented treadmill. Participants were outfitted with reflective markers and completed two 1-min periods of walking at each of 3 speeds (0.8 m/s, 1.2 m/s, and 1.6 m/s). Margins of stability were calculated relative to the anterior, posterior, lateral, and medial boundaries of the base of support, and the value at heel strike and the minimum value during the first half of each single-support phase were extracted. Step length, swing time, and step width were calculated from motion capture and ground reaction force data. Extreme values for consecutive steps were identified using Poincaré plots, and margins of stability in each direction were compared between 'normal' and 'extreme' steps. Margins of stability in both the anterior and medial direction were lower prior to long and wide steps, respectively. Margins of stability in the anterior and medial directions were lower prior to quick steps, and margins of stability in the posterior and lateral directions were lower prior to slow steps. There were either no significant differences in margin of stability between 'normal' and 'extreme' steps at heel strike, or the direction of the relationship was reversed to that observed during single support. These data suggest that spatio-temporal variability may reflect adjustments in step placement and timing to compensate for transient periods of low mechanical stability when walking.

1. Introduction

One in three seniors fall annually, with significant consequences such as injury, fear of falling, and increased risk of admission to long-term care (Overstall, 2004). Most falls occur when walking (Berg, Alessio, Mills, & Tong, 1997). Changes in gait patterns are associated with aging and fall risk. Increased stride-to-stride variability in spatio-temporal features of walking is related to increased fall risk in daily life (Callisaya et al., 2011; Hausdorff, Rios, & Edelberg, 2001; Maki, 1997). Gait variability in older adults may result

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http://dx.doi.org/10.1016/j.humov.2017.09.014
Received 19 June 2017; Received in revised form 22 August 2017; Accepted 21 September 2017
0167-9457/

from age-related physiologic changes, such as neurodegeneration, decreased range-of-motion, pain, and skeletal muscle deconditioning (Callisaya, Blizzard, Schmidt, McGinley, & Srikanth, 2010; Hausdorff, 2005; Kang & Dingwell, 2008; Tian et al., 2017). Variability may also reflect repeated adjustments in step placement and timing to correct for periods of low mechanical stability when walking (Callisaya et al., 2010; Hak, Houdijk, Steenbrink, Mert, van der Wurff, Beek, & van Dieën, 2012).

Mechanical stability is defined by the relationship between the state of centre of mass (COM) and base of support (BOS; Pai & Patton, 1997). 'Margin of stability' (MOS) has been proposed (Hof, Gazendam, & Sinke, 2005) to quantify mechanical stability during dynamic tasks such as walking (Hof, 2008). MOS is defined as the distance between the BOS and the extrapolated COM, with the extrapolated COM accounting for both COM position and velocity. MOS quantifies how close an inverted pendulum is to falling, and indicates when a control strategy (e.g., taking a step or executing a reach-to-grasp reaction) may be needed to prevent a fall (Bruijn, Meijer, Beek, & van Dieën, 2013). Generally, if the extrapolated COM is within the BOS boundary, the individual is mechanically stable, whereas an extrapolated COM outside the BOS suggests instability. During walking, corrective action to low mechanical stability may include adjusting step length, width, or time. For example, swing duration is shortened following an external pelvic perturbation in either the anterior or lateral directions during walking (Vlutters, van Asseldonk, & van der Kooij, 2016).

Although spatio-temporal gait variability and MOS have been studied independently as measures of balance control during gait, few studies have investigated the relationship between these measures. Previous studies have examined changes in step placement and timing following external perturbations to gait (Hak et al., 2012; Vlutters et al., 2016); however, step placement and timing is variable even in the absence of an apparent external perturbation. The extent to which 'internal' perturbations, or transient bouts of low mechanical stability, during normal walking contribute to altered step placement and timing is unknown. It is possible that low, yet positive, values of MOS are sufficient to evoke changes in step placement and timing to preserve mechanical stability in walking. As an initial step towards examining the relationship between mechanical stability, gait variability, and fall risk in older adults, the purpose of the current study was to determine whether 'extreme' values of step length, width, or time follow periods of low mechanical stability during the single support phase among healthy young adults. From previous work of external perturbations to gait (Vlutters et al., 2016), we hypothesized that: a) anterior and posterior MOS would be decreased prior to long and short steps, respectively; b) medial and lateral MOS would be decreased prior to wide and narrow steps, respectively; c) anterior and medial MOS would be decreased prior to quick steps; and d) posterior and lateral MOS would be decreased prior to slow steps. Adjustments in step placement and timing should preserve mechanical stability (McAndrew Young & Dingwell, 2012). Therefore, we also aimed to determine the effect of altered step placement and timing on mechanical stability at heel strike. We hypothesized that there would be no differences in MOS at heel strike between short/long and normal steps, between narrow/wide and normal steps, or between quick/slow and normal steps.

2. Methods

This study involved secondary analysis of a portion of an existing publically-available dataset (J. Moore, Hnat, & van de Bogert, 2014; J. K. Moore, Hnat, & van den Bogert, 2015). Data collection methods are summarized below, with full details in the original paper (J. K. Moore et al., 2015).

2.1. Participants

The dataset from (J. K. Moore et al., 2015) consisted of data from 15 young adults with no neurological or musculoskeletal conditions that limited mobility. Only 11 individuals were included in the present analysis; the first three individuals were tested using a slightly different protocol, and there were problems detecting gait events for another participant (see Section 2.3). Included participants were three women and eight men, with mean (standard deviation) age, height, and weight of 24 (4.4) years, 1.73 (0.1) m, and 73.1 (10.9) kg, respectively.

2.2. Procedures

Participants were instrumented with 47 reflective markers positioned over anatomical landmarks on the head, arms, trunk, pelvis, legs, and feet. Participants' knee- and ankle widths were measured and recorded. A 10-camera Osprey motion capture system with Cortex software was used to measure marker movement (100 Hz sampling rate; Motion Analysis, Santa Rosa, California, USA). Participants walked on an R-Mill treadmill (Forcelink, Culemborg, Netherlands) with one built-in 6-degree-of-freedom force plate under each foot; force plate data were sampled at 100 Hz. Participants wore a safety harness during all trials, which likely had no influence on characteristics of walking (Stout, Wittstein, LoJacono, & Rhea, 2016). A calibration pose was initially completed. Participants then completed 10-min walks at each of three predetermined speeds (0.8 m/s, 1.2 m/s, 1.6 m/s). Each 10-min walk consisted of 1 min of unperturbed walking, 8 min of perturbed walking (pseudo-random fluctuations in the speed of the treadmill belt, such that perturbations were in the antero-posterior direction), and another minute of unperturbed walking (Fig. 1). A rest break was provided before the next speed.

2.3. Data processing

Motion capture and force plate data were filtered in Matlab v.R2014a (The Mathworks, Natick, Massachusetts, USA) using dual-