Gait & Posture 35 (2012) 472–477

Contents lists available at SciVerse ScienceDirect

Gait & Posture

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

Voluntarily changing step length or step width affects dynamic stability of human walking

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ARTICLE INFO

Article history:
Received 1 March 2011
Received in revised form 17 October 2011
Accepted 7 November 2011

Keywords:
Dynamic stability
Walking
Step width
Step length

ABSTRACT

Changes in step width (SW), step length (SL), and/or the variability of these parameters have been prospectively related to risk of falling. However, it is unknown how voluntary changes in SW and SL directly alter variability and/or dynamic stability of walking. Here, we quantified how variability and dynamic stability of human walking changed when individuals voluntarily manipulated SW and SL. 14 unimpaired, young adults walked on a treadmill at their preferred walking speed with normal gait, with a metronome and with narrower, wider, shorter and longer steps than normal. Taking narrower steps caused increased SL variability while mediolateral (ML) movements of the C7 vertebra (i.e., trunk) became locally more stable ($p < 0.05$) and anterior–posterior (AP) C7 movements became locally less stable ($p < 0.05$). Taking wider steps caused increased SW and SL variability, while ML C7 movements became both locally and orbitally less stable ($p < 0.05$). Any change in SL caused increased SW, SL, and stride time variability. When taking shorter steps, ML C7 movements exhibited greater short-term local and orbital instability, while AP C7 movements exhibited decreased short-term and long-term local instability ($p < 0.05$). When taking longer steps, AP, ML, and vertical C7 movements all exhibited increased long-term local instability and increased orbital instability ($p < 0.05$). Correlations between mean SW, SL and dynamic stability of C7 marker motions were weak. However, short-term voluntary changes in SW and SL did significantly alter local and orbital stability of trunk motions.

1. Introduction

In clinical studies, gait parameters are frequently used to indicate patients' dynamic stability. However, the correct interpretation of these results is not clear. Walking with wider and shorter steps than normal is often termed "cautious". However, retrospective studies indicate that individuals who exhibit increased fall risk sometimes walk with shorter [1,2], longer [3], narrower [1], wider and/or faster [3] steps than normal. Slower walking speeds alone lead to decreased local instability [4–6] yet are also associated with history of falling [7]. Slower walking speeds also increase motion variability [5,6], which may or may not also indicate increased risk of falling. In one study, too much *or* too little step width variability was associated with fall history in older adults who walked at normal speeds (>1 m/s) [8]. Results of these and other retrospective studies do not reveal a clear understanding of the relationship between gait characteristics and fall risk.

Prospective studies on gait characteristics and falling are similarly mixed. Maki [9] found that while shorter and slower steps predicted increased *fear* of falling, increased variability of stride length and speed doubled an individual's actual likelihood of falling. Additionally, individuals who fell while walking took wider steps with less step width (SW) variability [9]. Contrary to Maki [9], Hausdorff and colleagues [10] found that increased stride time (ST) variability over a 6 min walk predicted falls in older adults. Verghese and colleagues [11] found that slow gait speeds and increases in swing time and stride length variability all predicted increased fall risk in older adults. However, in both studies [10,11], the greater variability observed could have been due simply to slower walking speeds [5,6]. DeMott et al. [12] determined that falls in older individuals with peripheral neuropathy were predicted by greater step time variability when walking on irregular surfaces but not on smooth surfaces. Recently, others found that no primary gait variables (means nor variability) predicted falls, but subtle left–right asymmetries in statistical persistence of stride time did [13]. Thus, all five of these prospective studies reached different findings with similar methods and measures.

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doi:10.1016/j.gaitpost.2011.11.010

The fact that shorter, wider and/or more variable steps in patients are associated with increased fall risk might suggest that people who exhibit these patterns are more unstable. However, when external lateral stabilization was applied, young and older subjects both took narrower steps, without changing their mean step length (SL), SW variability, or SL variability [14]. Likewise, when we destabilized healthy young subjects by applying continuous perturbations, they took shorter, wider and faster steps and exhibited greater variability both of stepping parameters and trunk kinematics [15]. These gait changes were accompanied by specific increases in measures of local dynamic instability [16], a measure of within-step dynamic stability. If individuals adopted these gait characteristics to increase their stability, then voluntarily taking shorter or wider steps during unperturbed walking should lead specifically to decreased variability and local instability of stepping parameters and/or trunk movements.

Collectively, these earlier findings suggest two opposing ideas: people increase their risk of falling *because* they take shorter and/or wider steps, or they take wider and/or shorter steps *because* they are at greater risk of falling. We wanted to test the latter idea: i.e., that adopting wider and/or shorter steps would decrease individuals' instability thereby decreasing their risk of falling. We hypothesized that individuals could alter their orbital (i.e., step-to-step) and local dynamic (i.e., within-step) trunk stability by voluntarily changing their SW and SL. We further hypothesized that individuals would exhibit decreased orbital and local instability when walking with wider steps or shorter steps than when walking normally. Our findings would indicate whether and how voluntarily changing gait characteristics contributes to local and orbital stability during walking, the latter of which, in particular, has been linked to fall-risk status [17].

2. Methods

14 young healthy adults (seven male, seven female; age 18–35) participated. Participants were excluded for any history of lower extremity injuries, surgery or neurological conditions, which could affect their gait. The University of Texas Institutional Review Board approved this study, and all participants provided written, informed consent prior to participation.

Participants walked on a motorized treadmill (Desmo ProXL model, Woodway USA, Waukesha, WI). Each subject completed a ~10 min warm-up. The first 5 min were used to determine participants' preferred walking speeds (PWS) using an established protocol [5]. During the second 5 min, participants walked at their PWS. Participants then completed three 3-min walking trials for each of six experimental conditions. During the normal (NO) condition, participants walked normally. During the normal with metronome (NM) condition, participants walked normally with a metronome matched to their preferred cadence. During the SW manipulations, participants were instructed to walk with wider (WI) and narrower (NA) steps than normal. During the SL manipulations, participants walked with shorter (SH) and longer (LO) steps. The latter were achieved by walking in time with a metronome, set to a cadence that was 10 steps/min faster or slower, respectively, than their preferred step cadence. Participants walked at their PWS for all conditions. The NO condition was always presented first and the remaining five conditions were presented in a random order to minimize learning effects. Rest breaks, during which the treadmill was stopped, were provided between conditions.

Participants wore reflective markers on their trunk and feet. Ten Vicon (Oxford Metrics, Oxford, UK) MX cameras captured participants' motion at 60 Hz. Vicon Nexus software was used to reconstruct, label and export the data for further processing in Matlab (The Mathworks, Inc.).

SL was defined as the anterior posterior distance between the heel markers at heel strike. SW was defined as the lateral distance between heel markers at heel strike. Stride time (ST) was the amount of time elapsed between two consecutive heel strikes of the same foot. Means and standard deviations of SL, SW and ST were calculated for each trial.

For stability analyses, we focused on trunk motion stability as indicated by the C7 vertebral marker motion. Trunk motions were studied because maintaining dynamic stability of the upper body is a primary objective of human locomotion [18]. Delay embedded state spaces describing trunk motion were constructed for the anterior–posterior (AP), mediolateral (ML) and vertical directions from the C7 marker velocity and time-delayed copies of the C7 marker velocity [5,16]:

$$\mathbf{S}(t) = [\nu(t), \nu(t+\tau), \ldots, \nu(t+(d_E-1)\tau)] \tag{1}$$

where $\mathbf{S}(t)$ is the d_E-dimensional state vector, $\nu(t)$ is the original 1-dimensional data (i.e., C7 velocity in the AP, ML or vertical direction), τ is the time delay and d_E is the embedding dimension. Time delays were set equal to τ = 0.333, 0.250 and 0.167 (i.e., 20, 15 and 10 data samples) for the ML, AP and vertical directions, respectively. Results were not expected to be sensitive to the exact values of τ [19]. For the local stability analysis, 120 consecutive strides of data were normalized to 12,000 total data points, or approximately 100 data points per stride [4,20], prior to defining the state space.

Floquet multipliers (FMs) estimated orbital instability, which is defined as the ability of a periodic system (i.e., one's motion) to return to a "preferred" state (i.e., a limit cycle) within one stride after being perturbed away from that preferred limit cycle state. FMs are specifically defined for use with periodic systems or motions. The calculations are based on well-established techniques [16,21–23] and are explained in detail in the Supplementary material. When the magnitude of the largest FM (MaxFM) is <1, this indicates orbital stability (i.e., state space trajectories converge towards the limit cycle after successive strides). Relative increases in MaxFM indicate increases in orbital instability.

Local divergence exponents (λ^*) estimated local dynamic instability, which is defined as the rate at which a system responds in real time to infinitesimally small perturbations away from some nominal state. Unlike Floquet multipliers, these λ^* exponents are defined to quantify stability of aperiodic systems. It is thus assumed that there is no specific "preferred" state or limit cycle [22]. Here, short-term (λ^*_S) and long-term (λ^*_L) exponents were calculated between 0 and 1 strides (λ^*_S) and between 4 and 10 strides (λ^*_L), respectively [24]. Positive λ^* indicate local instability (i.e., state space trajectories diverge away from each other in real time). Smaller, positive λ^* indicate less instability than larger, positive λ^*.

Both sets of nonlinear stability analyses were conducted here because walking is neither purely periodic nor purely aperiodic, but lies somewhere in between. Thus, it is appropriate to calculate both orbital and local dynamic stability, as each of these measures quantifies unique aspects of how humans respond to small perturbations [22]. All stability calculations were performed separately for C7 marker movements in the ML, AP and vertical directions.

Two-factor (Condition × Subject) analyses of variance (ANOVA) were used to assess statistical differences between SW, SL, ST and SW, SL and ST variability, MaxFM, λ^*_S and λ^*_L for the SW and SL conditions separately (i.e., NO vs. WI and NA and NM vs. LO and SH). p-values <0.05 were considered significant. All statistical analyses were conducted using PASW Statistics 18 (SPSS, Inc., Chicago, IL). Correlations were calculated using Matlab.

3. Results

Instructing participants to walk with narrower or wider steps than normal resulted in SWs that were narrower and wider than normal ($p < 0.0005$; Fig. 1A), as expected. Walking with wide steps significantly increased SW variability ($p < 0.0005$; Fig. 1B). Narrow steps were associated with increased mean SL ($p = 0.002$) and SL variability ($p < 0.0005$) whereas wide steps decreased mean SL ($p < 0.005$) and increased SL variability ($p < 0.0005$). Both narrower and wider steps caused increases in mean ST ($p < 0.0005$ and $p = 0.004$, respectively) and decreases in ST variability ($p = 0.001$ and $p < 0.0005$, respectively). There were significant subject interactions for mean SL ($p < 0.0005$), SL variability ($p = 0.027$) and mean ST ($p < 0.0005$) but not mean SW ($p = 0.276$), SW variability ($p = 0.08$) or ST variability ($p = 0.105$).

When walking with wide steps, participants' C7 marker movements exhibited increased short-term local instability (λ^*_S) in all directions of motion ($p < 0.0005$; Fig. 2A). Walking with wide steps was also associated with greater long-term local instability (λ^*_L; Fig. 2B) and less orbital stability (i.e., larger MaxFM; Fig. 2C) of the ML C7 marker movements ($p < 0.0005$). AP C7 marker movements became more long-term locally stable, however, when walking with wide steps ($p < 0.0005$; Fig. 2B). When walking with narrow steps, ML C7 marker motions became locally more stable (i.e., decreased λ^*) in both the short- ($p < 0.0005$) and long-term ($p = 0.014$) (Fig. 2A and B). Conversely, AP C7 marker motions became locally more unstable in both the short- and long-term ($p < 0.0005$; Fig. 2A and B). Vertical C7 marker motions exhibited greater short-term local instability ($p < 0.0005$; Fig. 2A and B) when walking with narrow steps. There were significant subject interactions for λ^*_S calculations in all directions ($p < 0.05$) and for λ^*_L and MaxFM in the AP direction ($p < 0.02$).

Walking with longer steps significantly increased mean SW and SW variability ($p < 0.0005$; Fig. 3A). However, walking with short