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Dynamic Stability of Superior vs. Inferior Segments during Walking in Young and Older Adults

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Abstract

Active control of trunk motion is believed to enable humans to maintain stability during walking, suggesting that stability of the trunk is prioritized over other segments by the nervous system. We investigated if superior segments are more stable than inferior segments during walking and if age-related differences are more prominent in any particular body segments. Eighteen healthy older adults and 17 healthy young adults walked on a treadmill for 2 trials of 5 minutes each at their preferred speed. 3D kinematics of the trunk, pelvis, and left thigh, shank, and foot were recorded. Local divergence exponents and maximum Floquet multipliers (FM) were calculated to quantify each segment's responses to small inherent perturbations during walking. Both older and younger adults walked with similar preferred walking speeds ($p = 0.86$). Local divergence exponents were larger in inferior segments ($p<0.001$), and larger in older adults ($p<0.001$). FM was larger in the superior segments ($p<0.001$), and larger in older adults ($p<0.001$). The age-associated difference in local divergence exponents was larger for trunk motion (interaction $p = 0.02$). Thus, superior segments exhibited less local instability but greater orbital instability. Trunk motion was more sensitive to age-associated differences in dynamic stability during gait. Trunk motion should be considered in studying age-related deterioration of gait.

Keywords

Falls; Aging; Gait; Stability; Upper body

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Conflict of Interest Statement

The authors declare that there is no conflict of interest associated with this work.

Introduction

Falls are a prevalent health problem for the elderly [1]. Most falls occur during whole-body movements like walking [2]. Understanding falls thus requires investigating how the nervous system controls the dynamics of walking, and how this control may change with aging.

The trunk segment constitutes of over half of body mass [3], and greatly influences the dynamics of the rest of the body. "Active," neuromuscular control of trunk motion is believed to enable people to maintain stability during walking [4], suggesting the nervous system may prioritize stability of the trunk over other inferior segments [5]. This implies that the stability of body segments would follow the kinematic chain, where superior segments (trunk) are more stable than inferior segments (feet).

Older adults exhibit less dynamic stability during gait, as assessed through the sensitivity of the whole body [6] or trunk motion [7] to small perturbations during gait. However, it is not clear if these age-related differences in stability are present in other segments, or if differences in the dynamic stability of these other segments may better indicate age-related decrements in gait function.

In this study we determined the dynamic stability of body segments (trunk, pelvis, thigh, shank, and foot) in healthy young and older adults. We hypothesized that superior segments would exhibit less local and orbital dynamic instability, as indicated by lower values of specific metrics that directly quantify the sensitivity of segment movements to small perturbations during walking [7,8]. We also tested whether the age-associated differences in dynamic stability are more prominent in superior segments.

Methods

Eighteen healthy older adults (age 72.1 (6.0); 6 women) and 17 young adults (age 23.3 (2.6); 5 women) participated after providing informed consent as approved by University of Texas at Austin (IRB protocol #2005-03-0013). Subjects were selected such that older adults would have similar gender ratio, height, and body mass (1.70 (.10) m, 73.2 (12.3) kg) as young adults (1.73 (0.09) m, 71.1 (9.86) kg).

Subjects walked on a level treadmill (Desmo S model, Woodway USA, Waukesha WI) while wearing a safety harness (Protecta International, Houston TX) that allowed natural arm and leg swing. Subjects wore their own comfortable walking shoes. Each subject's preferred walking speed (PWS) was determined using an established protocol [9]. This also allowed subjects to acclimate to the treadmill and warm-up. Subjects then completed two 5-minute walking trials at their PWS.

The kinematics of the trunk, pelvis, left thigh, left shank and both feet were tracked using an 8-camera Vicon 612 motion analysis system (Oxford Metrics, London UK). A cluster of markers was used on each segment and feet markers were used to track heel-strike events. Marker kinematics were low-pass filtered using a zero-lag Butterworth filter with a cutoff of 10 Hz. Linear motions of each segment were defined using a virtual center marker, defined as the average location of the markers on each segment [7]. 3D rotations were defined using the Cardan Y-x'-z" (tilt-obliquity-rotation) convention relative to the laboratory reference frame. Velocities were calculated using the standard 3-point difference formula to reduce non-stationarities in the displacement data [9]. These velocities were non-dimensionally scaled to body size [10] to make valid comparisons across subjects.

For each body segment (i.e., trunk, pelvis, left thigh, left shank, and left foot) a 12-dimensional state-space was defined using linear and angular velocities [$\dot{x}\dot{y}\dot{z}\dot{\theta}\dot{\varphi}\dot{\psi}$] and their time-delayed copies, forming a 12-dimensional state space [7,11] (Equation 1).

$$\begin{aligned} S(t) \quad &= [\,\dot{x}(t)\ \ \dot{y}(t)\ \ \dot{z}(t)\ \ \dot{\theta}(t)\ \ \dot{\phi}(t)\ \ \dot{\psi}(t)\ \ \ldots \\ &\quad \dot{x}(t-\tau_1)\ \ \dot{y}(t-\tau_2)\ \ \dot{z}(t-\tau_3)\ \ \dot{\theta}(t-\tau_4)\ \ \dot{\phi}(t-\tau_5)\ \ \dot{\psi}(t-\tau_6)] \end{aligned} \tag{1}$$

Time delays were calculated from the first minimum of the average mutual information function of each time series [12].

From the state-space data, local divergence exponents (λ^*_S, λ^*_L) [9,13], and maximum Floquet multipliers (FM) [6,14] were calculated using standard methods to quantify the local and orbital dynamic stability of each segment, respectively (Figure 1).

The differences in the local divergence exponents (λ^*_S, λ^*_L) and maximum FM at 0, 25, 50, and 75% of the gait cycle (where 0% and 50% corresponded to heel strikes and 25% and 75% corresponded to single limb support) were compared between age groups and segments using a repeated-measures ANOVA using SPSS 14 (SPSS, Chicago IL).

Results

Older subjects walked at nearly the same preferred speeds (1.29 (0.15) m/s) as younger subjects (1.30 (0.10) m/s; $p = 0.86$) [7]. Both λ^*_S and λ^*_L (Fig. 2) were smaller (i.e., less locally unstable) in superior segments ($p < 0.001$). In contrast, maximum FM (Fig. 3) were larger (i.e., more orbitally unstable) in the superior segments at 25, 50, and 75% of the gait cycle ($p < 0.001$). Older adults exhibited both larger λ^*_S and λ^*_L ($p < 0.001$; Fig. 2) and larger maximum FM ($p < 0.001$; Fig. 3) than healthy younger adults. Thus, older adults exhibited greater sensitivity to small perturbations across all segments. The age-associated difference in λ^*_S was larger in the trunk motion (interaction $p = 0.02$).

Discussion

Superior segments (i.e., the trunk) exhibited less local instability (λ^*_S and λ^*_L), but greater orbital instability (FM) compared to inferior segments (e.g., shank, foot). This confirms our previous finding that local dynamic stability and orbital stability quantify different aspects of stability [8]. The results suggest that the superior segments are less sensitive to very small initial perturbations (i.e., smaller λ^*), and thu its motion is initially less affected by these small perturbations, compared to inferior segments. Yet, one gait cycle after the initial perturbation, the motions of the superior segments return toward the average gait pattern more slowly (i.e., larger maximum FM), compared to inferior segments.

These differences are likely due to the greater inertia of the trunk segment. The greater inertia may attenuate the effect of a given perturbation on the trunk motion compared to segments with smaller inertia. Yet, this greater inertia of the trunk may also make it difficult to correct its motion quickly, which may make feed-back control less effective, since it would require longer time and larger forces than for smaller, lighter segments. The nervous system may need to rely more on feed-forward control to regulate trunk movements than for more distal limb segments.

Of note, the calculation of dynamic stability from gait data does not seem sensitive to the formulation of the state space as theoretically predicted, as long as the relative magnitudes of the state variables are not very different [15]. A state-space based on velocities and accelerations was not used as previously [7] because the large differences in the relative magnitudes of the state variables were producing numerical and computational problems, especially in the feet that undergo relatively very large motions. However, repeating the analysis after standardizing these state variable to unit variance produced similar results as shown here, demonstrating the dynamic equivalence of these state spaces [15,16].

Older adults exhibited less dynamic stability across all segments, yet the trunk motion displayed the greatest age-related differences. Thus, trunk motion dynamics appears to provide a more sensitive marker of the decline in gait function in healthy older adults compared to other body segments. Likewise, in older adults at risk for falls, changes in trunk motion dynamics during gait may also be a better marker of impaired gait function than lower extremity motion. Future studies of gait disorders and fall risk in older adults need to consider the dynamic stability of trunk motion during walking.

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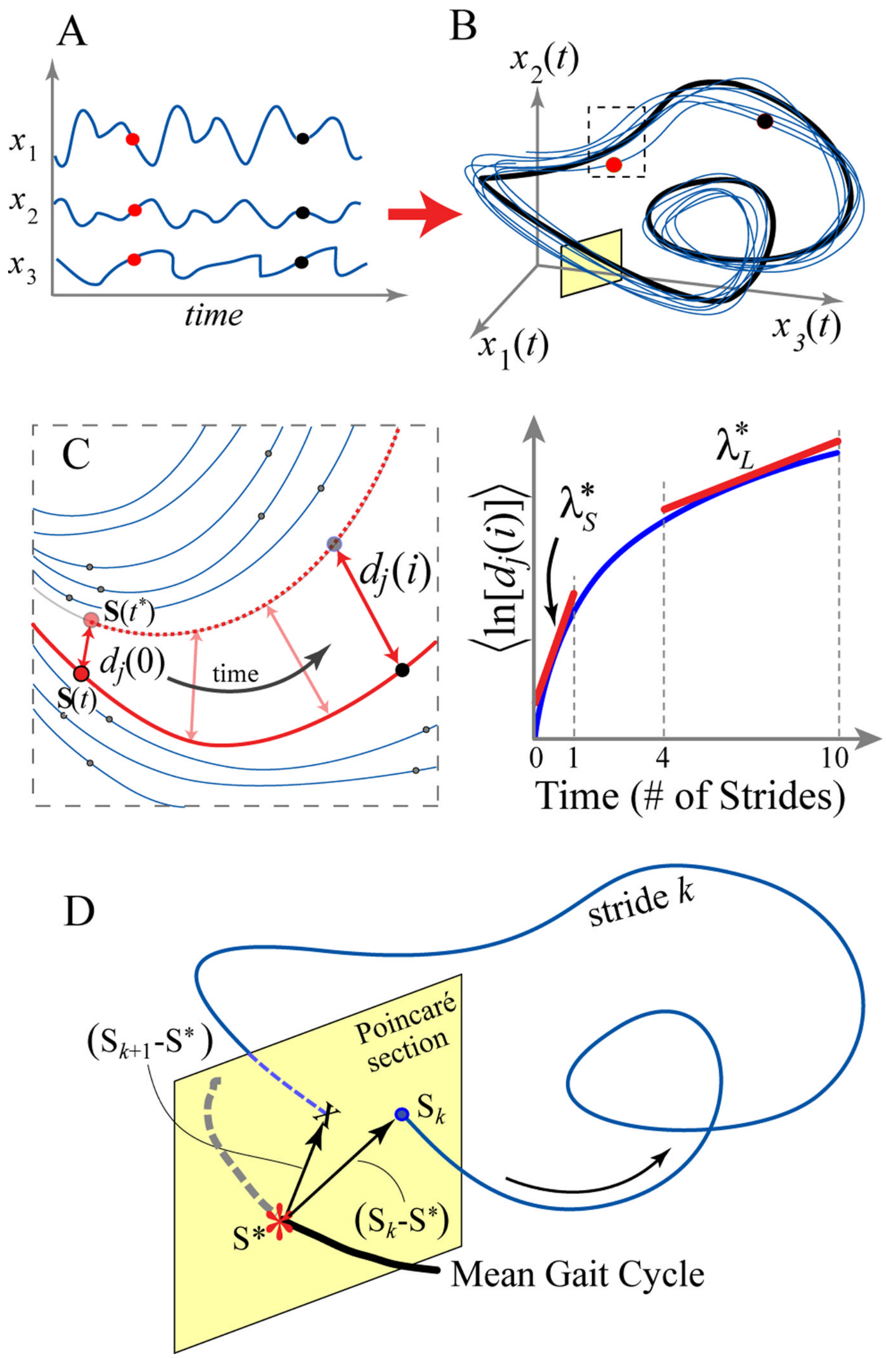

Figure 1.

Schematic Representation of State-space Construction. (***A***) The original time series of linear and angular velocities define the states (x_1, x_2, …) of the system. (***B***) These states are combined to form the system's trajectory in state space (only 3 states are shown here for illustrative purposes). (***C***) Expanded view of a typical local region. A small perturbation moves the system at $\mathbf{S}(t)$ to its closest neighbor $\mathbf{S}(t^*)$. Local divergence is computed by measuring the Euclidean distances between the subsequent points, denoted $d_j(i)$. The local dynamic stability of the system is defined by how quickly, on average, the two trajectories diverge away from each other. Rates of divergence, λ^*_S and λ^*_L(local divergence

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exponents), were calculated from the slopes of the mean log divergence curve. (***D***) Poincaré sections are defined to be orthogonal to the mean (i.e., limit) cycle. The system state, $\mathbf{S}_k$, at stride k evolves to $\mathbf{S}_{k+1}$ one stride later. Floquet multipliers quantify, on average, whether the distances between these states and the system fixed point, $\mathbf{S}^*$, grow or decay after one cycle.

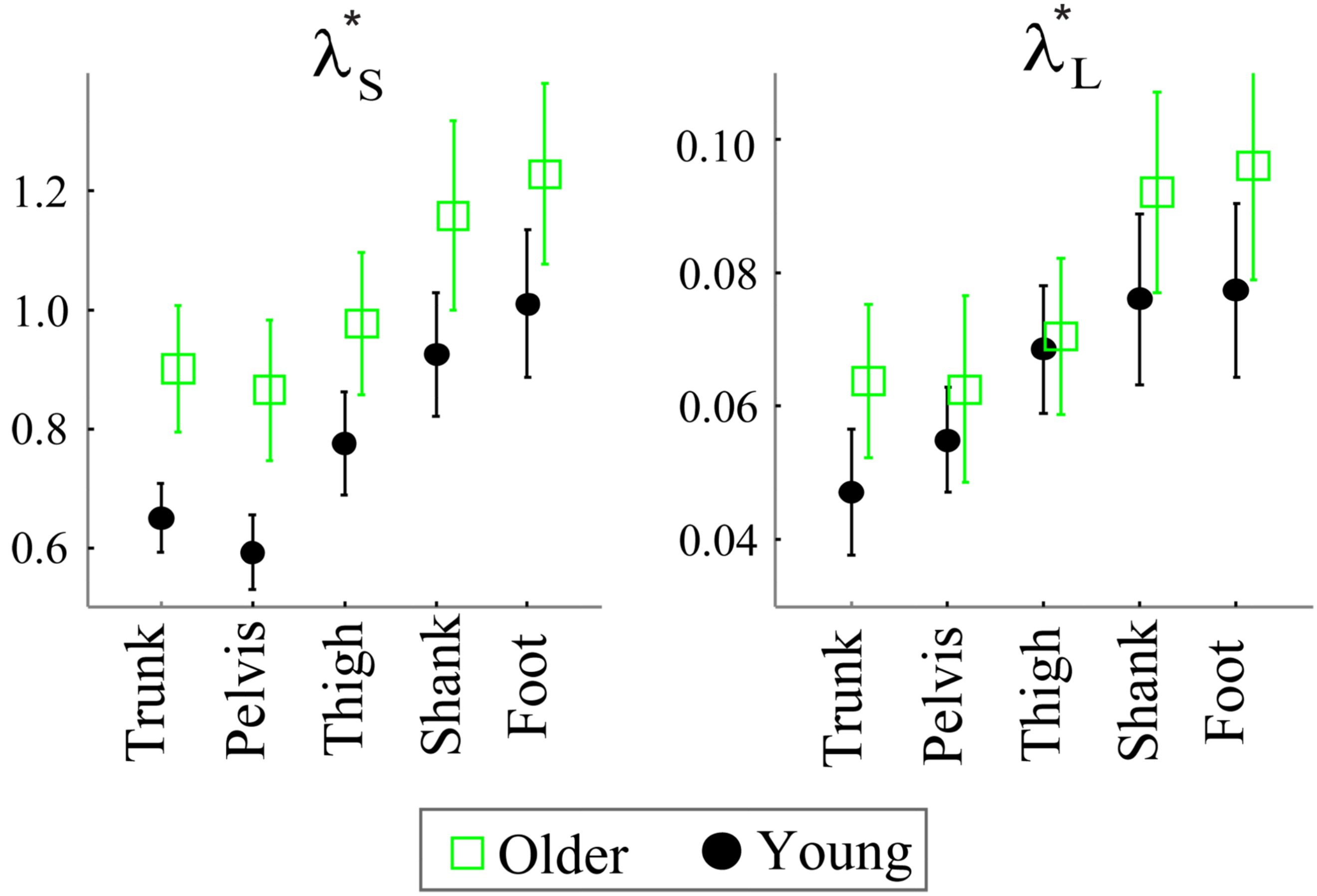

Figure 2.
Local divergence exponents for each body segment and age group
Symbols indicate group means. Error bars denote between-subjects standard deviations within each group.

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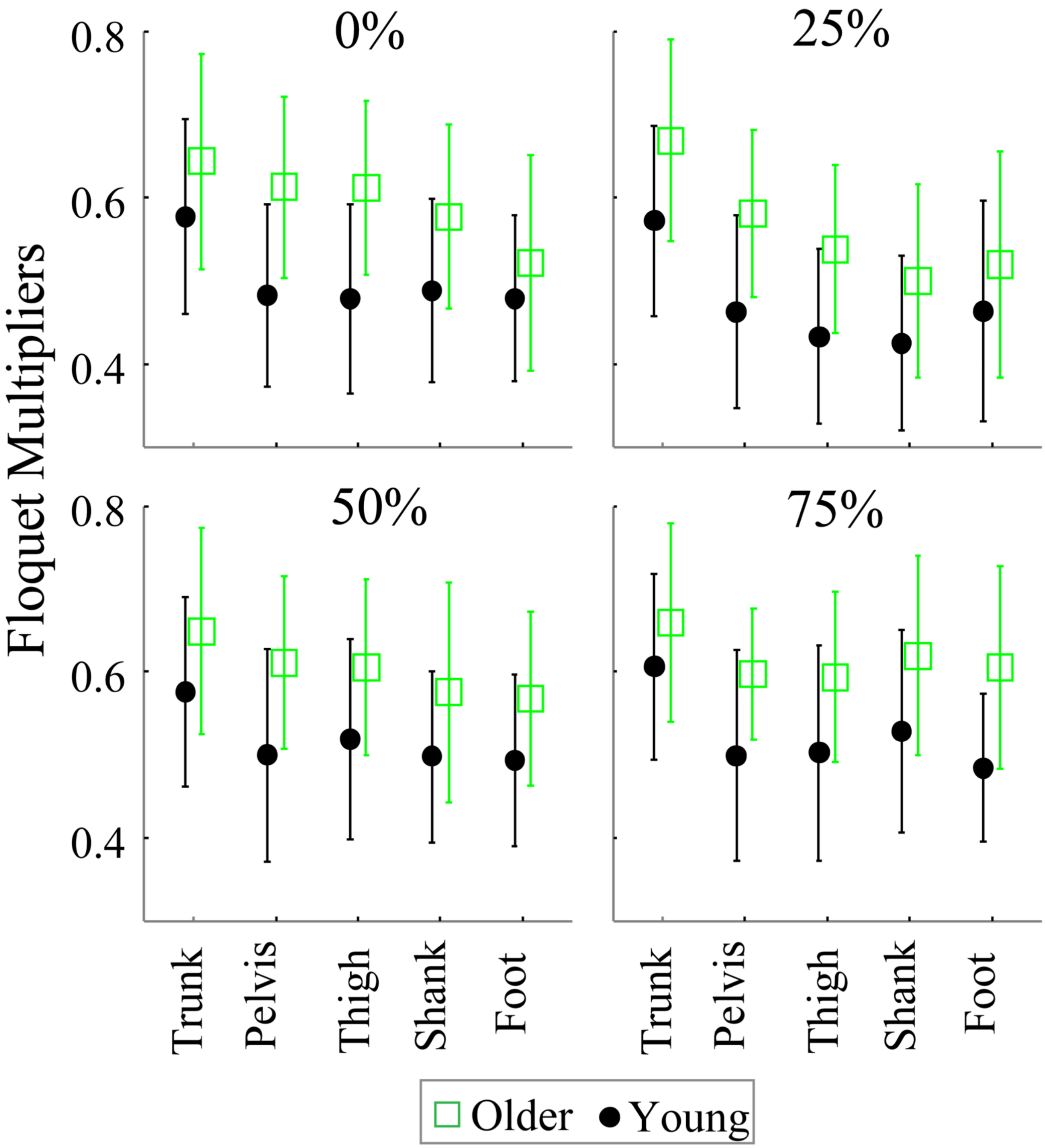

Figure 3.
Maximum Floquet multipliers for each body segment and age group
Error bars denote between-subjects standard deviations within each group.