Effects of walking speed, strength and range of motion on gait stability in healthy older adults

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ABSTRACT

Falls pose a tremendous risk to those over 65 and most falls occur during locomotion. Older adults commonly walk slower, which many believe helps improve walking stability. While increased gait variability predicts future fall risk, increased variability is also *caused* by walking slower. Thus, we need to better understand how differences in age and walking speed independently affect dynamic stability during walking. We investigated if older adults improved their dynamic stability by walking slower, and how leg strength and flexibility (passive range of motion (ROM)) affected this relationship. Eighteen active healthy older and 17 healthy younger adults walked on a treadmill for 5 min each at each of 5 speeds (80–120% of preferred). Local divergence exponents and maximum Floquet multipliers (FM) were calculated to quantify each subject's inherent local dynamic stability. The older subjects walked with the same preferred walking speeds as the younger subjects ($p = 0.860$). However, these older adults still exhibited greater local divergence exponents ($p < 0.0001$) and higher maximum FM ($p < 0.007$) than the younger adults at all walking speeds. These older adults remained more locally unstable ($p < 0.04$) even after adjusting for declines in both strength and ROM. In both age groups, local divergence exponents *decreased* at slower speeds and increased at faster speeds ($p < 0.0001$). Maximum FM showed similar changes with speed ($p < 0.02$). Both younger and older adults exhibited *decreased* instability by walking slower, in spite of *increased* variability. These increases in dynamic instability might be more sensitive indicators of future fall risk than changes in gait variability.

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1. Introduction

The injuries associated with falls are the primary cause of accidental deaths in those over 65 years (Sterling et al., 2001) and most of these falls occur during locomotion (Berg et al., 1997; Niino et al., 2000). Slower speeds are common among older adults and more pronounced in those with locomotor impairments (Alexander, 1996). Slow gait speeds predicted prospective fall risk in some studies (Bergland et al., 2003). In others, increased gait variability, but *not* slow speed, predicted future risk of falls (Maki, 1997; Hausdorff et al., 2001). However, variability does not quantify *stability* in terms of sensitivity to perturbations (Dingwell and Cusumano, 2000). What is needed is a better understanding of how differences in both age and walking speed affect dynamic stability.

Older people may walk slower because of physical constraints, like decreased strength (DeVita and Hortobagyi, 2000) and/or flexibility (Kerrigan et al., 2001), that prevent them from walking faster. Decreased leg strength in the elderly correlated with poorer functional walking performance related to balance control (Moxley Scarborough et al., 1999) and increased relative muscular demand during obstacle crossing (Hahn et al., 2005). Decreased leg muscle strength also discriminated elderly who fell after a perturbation from those who did not (Pijnappels et al., 2008). Conversely, while strength training increased walking speed in older adults (Chandler et al., 1998; Holviala et al., 2006), this did not reduce their fall risk. In fact, faster walking speeds may instead *increase* fall risk in older adults (Berg et al., 1997; Pavol et al., 1999; van den Bogert et al., 2002). Conversely, walking slower may be a pro-active strategy to improve walking stability (Shkuratova et al., 2004). Patients with diabetic neuropathy who did *not* exhibit significant strength or flexibility declines, walked slower and exhibited slightly *de*creased local dynamic instability

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(Dingwell et al., 2000), despite *in*creased gait variability (Dingwell and Cavanagh, 2001). When walking speed was directly manipulated in younger adults, slower speeds similarly lead to *de*creased local dynamic instability, despite *in*creased variability (Dingwell and Marin, 2006; England and Granata, 2007). Thus, a better understanding of how age-related declines in strength and flexibility interact with changes in walking speed to affect walking stability is also needed.

Surprisingly, while slower speeds and increased gait variability are both commonly reported in older populations, no study to date has directly examined how changes in walking speed affect variability and/or dynamic stability in older adults. We recently demonstrated that very healthy active elderly who walk at the same preferred speeds as younger adults still exhibit greater variability (Kang and Dingwell, 2008). This was only partly explained by their decreased strength and flexibility.

The purpose of this study was therefore, to determine if the older adults tested in Kang and Dingwell (2008) also improved their dynamic stability by walking slower. We hypothesized that these older adults would exhibit greater local dynamic instability than younger adults, even at the same walking speeds. We also hypothesized that these older adults would exhibit decreased instability at slower speeds and greater instability at faster speeds, as do younger adults (Dingwell and Marin, 2006; England and Granata, 2007). Finally, we hypothesized that differences in local dynamic stability between younger and older adults would be explained by changes in leg strength and range of motion (ROM) (Kang and Dingwell, 2008).

2. Methods

Eighteen healthy older adults (age 65–85) and 18 height-, weight-, and gender-matched younger adults (age 18–28), participated after signing an institutionally approved informed consent form (Table 1) (Kang and Dingwell, 2008). Subjects were screened to exclude anyone who reported any history of recent lower extremity injuries or disabilities, any visible gait asymmetries, or were taking any medications that might influence their ability to walk.

Bilateral hip, knee, and ankle passive ranges of motion were measured using a goniometer. Bilateral isometric strength (joint torque) of the hip flexors, extensors, knee flexors, extensors, dorsiflexors and plantarflexors were measured using a hand-held dynamometer (Lafayette Instrument Company, Lafayette IN) (Smidt, 1994). Composite "Strength" and "ROM" scores were defined for each subject using principal components analysis (Dingwell and Cavanagh, 2001; Daffertshofer et al., 2004). Each score was defined as the first principal component of each set of individual measures. This yielded the linear combination of those measures that explained the most variance in the data (Kang and Dingwell, 2008).

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 swing. Subjects wore their own comfortable walking shoes. Each subject's preferred walking speed (PWS) was determined using a previously established protocol (Dingwell and Marin, 2006). This also allowed subjects to acclimate to the treadmill and warm-up. Subjects then completed two

Table 1
Subject characteristics

	Younger adults	Older adults	p-Value
Gender (M/F)	12/5[a]	12/6	0.80[b]
Age (years)	23.3 ± 2.6	72.1 ± 6.0	<0.0001
Height (m)	1.73 ± 0.094	1.70 ± 0.104	0.36
Body mass (kg)	71.1 ± 9.86	73.2 ± 12.3	0.58
Body mass index (BMI)	23.5 ± 1.7	25.4 ± 3.2	0.07
Preferred walking Speed (m/s)	1.30 ± 0.10	1.29 ± 0.15	0.86
PWS range (m/s)	1.16–1.56	0.93–1.52	
Composite strength score	3.07 ± 3.71	−4.29 ± 2.35	<0.0001
Composite ROM score	1.89 ± 1.81	−2.03 ± 2.78	<0.0001

[a] Reflects the number after the data for one younger subject were discarded due to technical problems.
[b] χ^2 test.

5 min walking trials at each of 5 speeds, from $0.8 \times$ to $1.2 \times$ PWS. The order of presentation was pseudo-randomized to avoid consecutive fast trials to minimize fatigue. Subjects rested at least 2 min between trials. Subjects were instructed to look straight ahead and avoid extraneous movements. All data from one younger subject, and 1 trial from one older subject were discarded due to technical problems. A second older subject could not complete one of the $1.2 \times$ PWS trials, so this trial was also discarded.

Trunk kinematics were tracked using a Vicon 612 motion analysis system (Oxford Metrics, London UK) with six markers placed on the left and right acromia, C7 and T10 spinal processes, and the spines of each scapulae. Trunk motions were studied because maintaining dynamic stability of the upper body is a primary objective of human locomotion (Prince et al., 1994). Markers were also placed on both feet to track heel-strike events. The variability of these subjects walking kinematics was reported previously (Kang and Dingwell, 2008).

Marker kinematics were low-pass filtered with a cutoff of 10 Hz using a zero-lag Butterworth filter. Linear trunk motions were defined using a virtual center marker, defined as the average location of the six torso markers (Kang and Dingwell, 2006). This minimized the effects of measurement noise and non-rigid behavior (e.g., subtle bending, raising the shoulders, twisting, etc.) of the trunk. Three-dimensional (3D) trunk rotations were defined using the Cardan $Y{-}x'{-}z''$ (tilt–obliquity–rotation) convention relative to the laboratory reference frame (Goldstein et al., 2001). Velocities and accelerations were calculated using the standard 3-point difference formula to reduce non-stationarities in the displacement data (Dingwell and Marin, 2006). All state variables were then low-pass filtered with a cutoff of 10 Hz using a zero-lag Butterworth filter. These variables formed a 12D state space (e.g., Figs. 1A and B):

$$S(t) = [\dot{x}, \dot{y}, \dot{z}, \ddot{x}, \ddot{y}, \ddot{z}, \dot{\theta}, \dot{\phi}, \dot{\psi}, \ddot{\theta}, \ddot{\phi}, \ddot{\psi}] \in \Re^{12} \tag{1}$$

The dynamic coupling between limb segments (Zajac and Gordon, 1989) ensures that while leg movements were not measured directly, these motions influenced the measured trunk segment motions and were thus captured by our analyses. Likewise, human standing and walking can be well described by as few as five state variables (Dingwell and Cusumano, 2000) and/or muscle synergies (Ivanenko et al., 2007; Torres-Oviedo and Ting, 2007). Thus, this 12D $S(t)$ adequately described walking dynamics (Kang and Dingwell, 2006).

Humans naturally experience many small perturbations during normal walking. These include both *external* (e.g., environmental and/or sensory) and *internal* perturbations created by neuromuscular noise (Faisal et al., 2008). These perturbations manifest themselves as the natural variations exhibited during walking. The trunk's responses to these small inherent perturbations were quantified in two ways (Dingwell et al., 2007). First, *Local* dynamic stability (Fig. 1C) of walking was quantified (Dingwell and Cusumano, 2000). Euclidean distances between neighboring trajectories in state space were computed as a function of time (Fig. 1C) and averaged over many original pairs of initially nearest neighbors (Rosenstein et al., 1993). Local divergence exponents, λ^*, were estimated from the slopes of linear fits to these curves:

$$y(i) = \frac{1}{\Delta t} \langle \ln[d_j(i)] \rangle = [\lambda^*]i + c \tag{2}$$

where $d_j(i)$ is the Euclidean distance between the jth pair of initially nearest neighbors after i discrete time steps (i.e., $i\Delta t$ seconds) and $\langle \cdot \rangle$ denotes the average over all values of j (Rosenstein et al., 1993). Positive exponents indicate local instability, with larger exponents indicating greater instability.

Since the intrinsic time scales (i.e., average stride times) were different for each subject, the time axes of these curves were re-scaled by multiplying by the average stride frequency for each subject. Short-term exponents, λ^*_S, were calculated from the slopes of linear fits to the divergence curves between 0 and 1 stride. Long-term exponents, λ^*_L, were calculated as the slopes between 4 and 10 strides (Fig. 1C) (Dingwell et al., 2000; Dingwell and Marin, 2006).

Second, *Orbital* stability was quantified by calculating maximum Floquet multipliers (FM) using established techniques (Hurmuzlu et al., 1996; Donelan et al., 2004). First, we defined a Poincaré section (Fig. 1D) at each percent of the gait cycle (0–100%), where 0% defined left heel strike. The system state (Eq. (1)), S_k, for each stride, k, at that Poincaré section evolved to the state at the following stride, S_{k+1}, according to the Poincaré map:

$$S_{k+1} = F(S_k) \tag{3}$$

We defined the limit cycle trajectory as the average trajectory across all strides within a trial. This produced a single fixed point in each Poincaré section

$$S^* = F(S^*) \tag{4}$$

We then computed the effects of small perturbations away from these fixed points, using a linearized approximation of Eq. (3)

$$[S_{k+1} - S^*] \approx J(S^*)[S_k - S^*] \tag{5}$$

For $S \in \Re^{12}$, $J(S^*) \in \Re^{12 \times 12}$ had 12 eigenvalues. Conversely, the Poincaré section occupied an $\Re^{11}$ sub-space of S. However, since $[S_k{-}S^*]$ and $[S_{k+1}{-}S^*]$ were both constrained to lie in the Poincaré section (Fig. 1D), the last (i.e., 12th) eigenvalue of $J(S^*)$ had a value of ~0 and was spurious. Therefore, the first 11 eigenvalues of $J(S^*)$ defined the FM that quantified how much small perturbations in the Poincaré
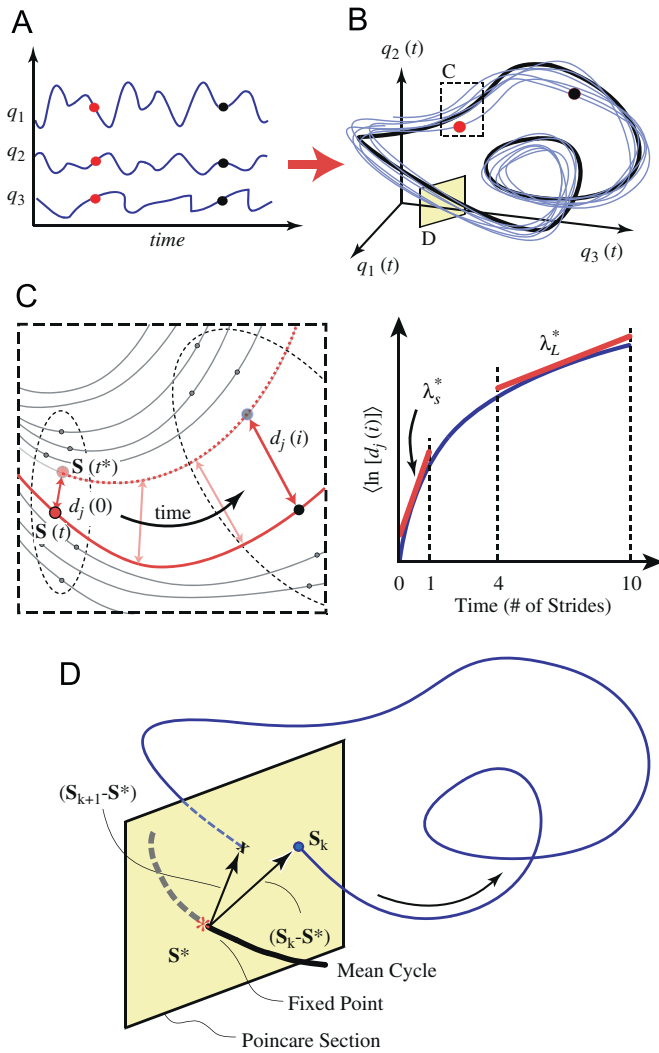
Fig. 1. Schematic representation of state-space construction. (A) The original time series of raw data define the states (q_1, q_2, …) of the system. (B) These states are combined to form the system's trajectory in state space (only a three-dimensional state space is shown here for illustrative purposes). (C) Expanded view of a typical local region. A small perturbation moves the system at $S(t)$ to its closest neighbor $S(t^*)$. 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, were calculated from the slopes of the mean log divergence curve (Eq. 2). (D) Poincaré sections are defined to be orthogonal to the mean (i.e., limit) cycle. The system state, S_k, at stride k evolves to S_{k+1} one stride later. Floquet multipliers quantify, on average, whether the distances between these states and the system fixed point, S^*, grow or decay after one cycle (Eq. 5).

section grew or diminished by the next cycle. If these FM all have magnitude <1, perturbations on average shrink by the next stride, and the system remains stable. We computed the magnitudes of the *maximum* FM for Poincaré sections at 0%, 25%, 50%, and 75% of the gait cycle (Dingwell and Kang, 2007; Dingwell et al., 2007).

The two local divergence exponents, λ^*_S and λ^*_L, and four maximum FM were compared between age groups and speeds using a general linear model analysis of variance (ANOVA) using SPSS 14 (SPSS, Chicago IL). Differences in ability to maintain dynamic stability could be due to differences in strength and/or flexibility. Therefore, these analyses were repeated as analyses of covariance (ANCOVA), including the composite strength and ROM scores as covariates.

3. Results

The older adults did exhibit lower composite strength and ROM scores than the younger adults ($p<0.0001$; Table 1).

However, the PWS of the older adults tested in this study were no different than those of the younger adults ($p = 0.86$; Table 1). These older adults did walk with shorter stride times and slightly shorter step lengths than the younger adults (Fig. 2), even at the same walking speeds.

Both younger and older adults exhibited positive local divergence (Fig. 3A), consistent with local instability, similar to previous results (Dingwell and Marin, 2006; Kang and Dingwell, 2006; England and Granata, 2007). We note that for "chaotic"
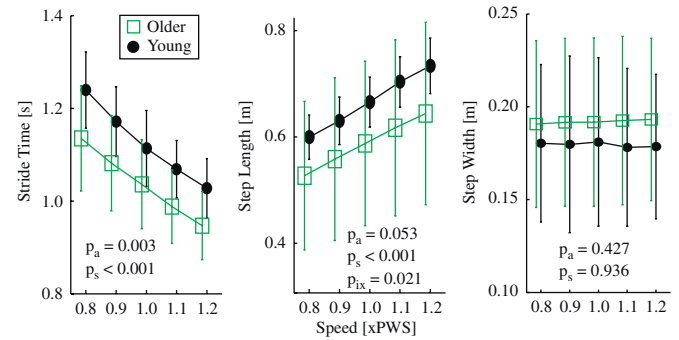

Fig. 2. Spatio-temporal gait metrics vs. walking speed. $1.0 \times$ PWS is preferred walking speed. Error bars denote between-subject standard deviations within each group. Note that symbols for older and younger groups have been offset slightly for clarity. ANOVA results for differences between age groups (p_a), walking speeds (p_s) and group × speed interactions (p_{ix}) are shown. Older adults exhibited shorter stride times at all 5 walking speeds. They also exhibited slightly shorter step lengths, but the differences did not quite reach statistical significance.
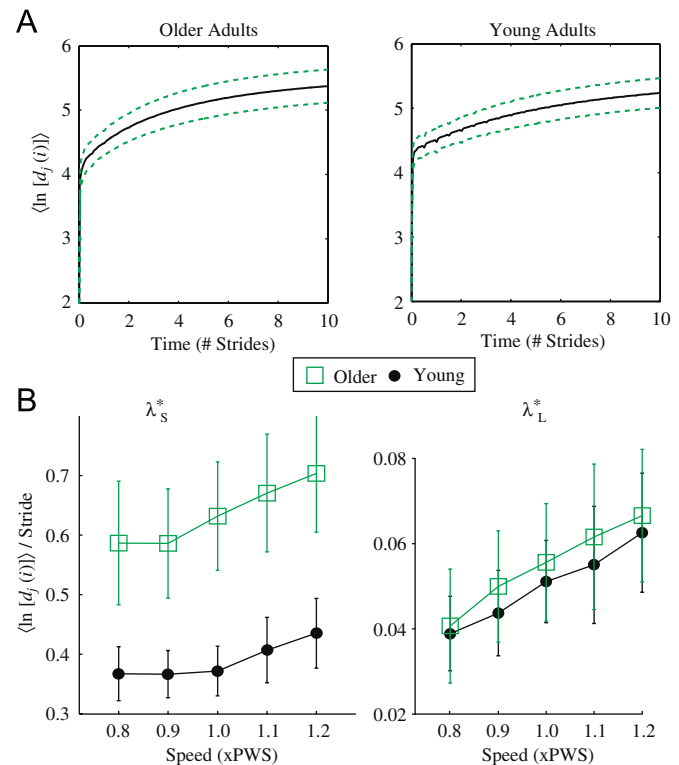

Fig. 3. (A) Sample local divergence curves for the $1.0 \times$ PWS walking speed for each age group. Solid lines are group averages. Dashed lines indicate ± 1 between-subject standard deviation bands. The *slopes* of these curves define local dynamic stability (Fig. 1C). (B) Local divergence exponents (λ^*) vs. speed and age. Error bars denote standard deviations within each group. Note that symbols for older and younger groups have been offset slightly for clarity. Older adults exhibited greater short-term (λ^*_S; $p<10^{-13}$), but not long-term (λ^*_L; $p = 0.192$) local instability. Both groups exhibited decreased local dynamic instability at slower speeds and increased instability at faster speeds ($p<0.001$).

systems, these curves should be straight over a reasonably long time scale, where the slope of that straight line approximates the maximum finite-time Lyapunov exponent for the system (Rosenstein et al., 1993). The curves we obtained were clearly *not* linear. Therefore, human walking is most likely not "chaotic" (Kantz and Schreiber, 2004), even though it is locally unstable (Dingwell and Kang, 2007).

Divergence curves for older adults showed slightly greater *magnitudes*. However, it is the *rates* of divergence, or slopes that determine local stability. Short-term divergence exponents, λ^*_S, were larger in older adults ($p < 10^{-13}$; Fig. 3B). Thus, the older adults were more locally unstable than the younger subjects. For both groups, λ^*_S increased with speed ($p < 0.001$), but this effect was more pronounced in the older adults (interaction $p = 0.007$). Long-term divergence exponents, λ^*_L did not differ between age groups ($p = 0.192$), but did increase with speed ($p < 0.001$; Fig. 3B). Slower walking speeds led to improved local dynamic stability in both the younger and older subjects.

Short-term divergence exponents, λ^*_S, were correlated with both strength and ROM scores (Fig. 4), indicating that lower strength and ROM were both associated with increased instability. However, the differences between age groups for λ^*_S remained significant even after strength ($p < 10^{-8}$), ROM ($p < 10^{-9}$), or both composite scores ($p < 10^{-6}$) were included as covariates (Table 2). Thus, differences in strength and ROM did not explain all the differences in local dynamic stability between younger and older subjects.
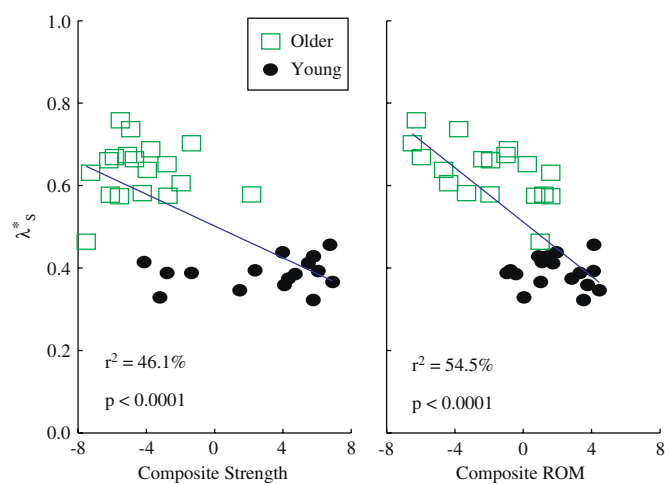
The magnitudes of the maximum FM varied little across the gait cycle (Fig. 5A). Older adults exhibited higher maximum FM values at all four Poincaré sections tested ($p < 0.007$; Fig. 5B). Maximum FM were slightly larger at higher speeds at the 75% Poincaré section ($p < 0.001$), where maximum FM at $0.8 \times$ and $0.9 \times$ PWS speeds were less than at $1.1 \times$ and $1.2 \times$ PWS ($p < 0.003$, pairwise comparisons). However, this effect was small at the 25% and 50% Poincaré sections ($p < 0.02$), and not significant at the 0% Poincaré section ($p < 0.4$).

Similar to λ^*_S, maximum FM were also correlated to strength and ROM scores (Fig. 6), where decreased strength and ROM were associated with greater instability. And likewise, the differences between age groups for these maximum FM remained significant
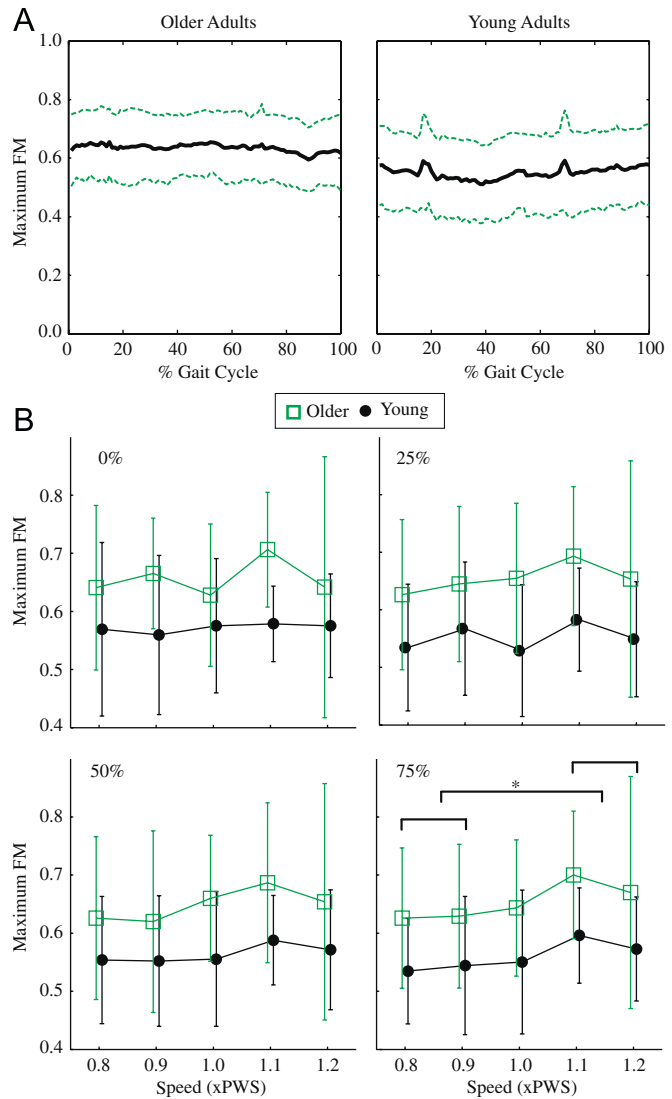

Fig. 5. (A) Maximum Floquet multipliers (FM) across the entire gait cycle for the $1.0 \times$ PWS walking speed for each age group. Solid lines are group averages. Dashed lines indicate ± 1 between-subject standard deviation bands. Maximum FM were consistent across the gait cycle. A few trials displayed a spike at ~20% or ~70% of the gait cycle, but this was not consistent between subjects or groups. Similar results were obtained at the other walking speeds. (B) Maximum FM values vs. speed for both age groups for the Poincaré sections at 0%, 25%, 50% and 75% of the gait cycle. Error bars denote ± 1 between-subject standard deviation within each group. Note that symbols for older and younger groups have been offset slightly for clarity. Older adults (squares) displayed higher maximum FM. The horizontal bracket at the 75% Poincaré section denotes significant Tukey's LSD post-hoc comparisons at $p < 0.005$. Maximum FM at the $0.8 \times$ and $0.9 \times$ PWS speeds were significantly different from those at both the $1.1 \times$ and $1.2 \times$ PWS speeds.
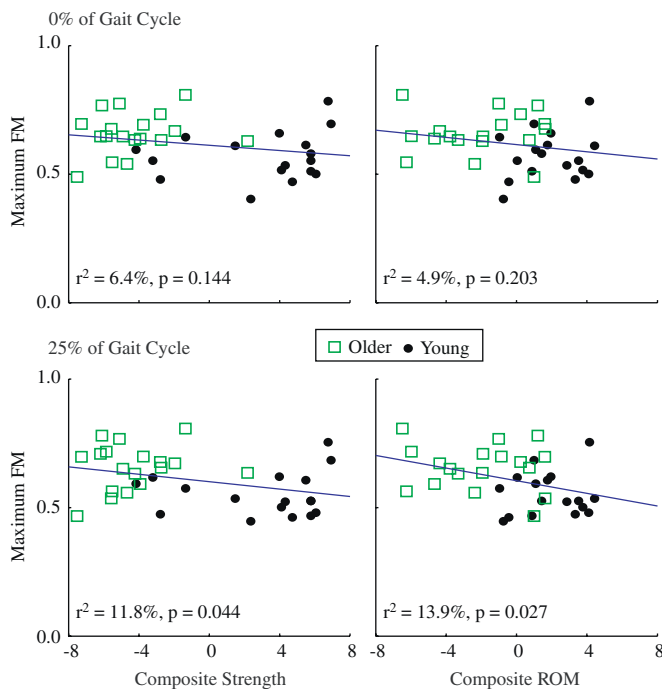

Fig. 4. Short-term local divergence exponents (λ^*_S) vs. composite strength and range of motion (ROM) scores. Each symbol represents the average value for one subject, averaged across speeds. Composite strength and ROM scores were both correlated with λ^*_S. These correlations explain some, but not all, of the age-related differences in λ^*_S (Table 2).

Table 2
Summary of ANOVA and ANCOVA comparison results

Effect	λ^*_S	λ^*_L	0%	25%	50%	75%
Age	<0.0001	0.192	0.0052	0.0013	0.0066	0.0024
Speed	<0.0001	<0.0001	0.344	0.0172	0.0137	<0.0001
Age × speed	0.0068	0.577	0.428	0.217	0.3569	0.525
Age adj. by strength	<0.0001	N/A[a]	0.0100	0.0110	0.0310	0.0035
Age adj. by ROM	<0.0001	N/A[a]	0.0112	0.0196	0.0920	0.0114
Age adj by both	<0.0001	N/A[a]	0.0107	0.0330	0.1101	0.0068

[a] Main effect for age was not significant, therefore, not applicable.

Fig. 6. Maximum Floquet multipliers (FM) vs. composite strength and range of motion (ROM) scores. Each symbol represents that average value for one subject, averaged across speeds. Correlations are shown for maximum FM computed at Poincaré sections at 0% and 25% of the gait cycle. Results from 50% and 75% were similar. Composite strength and ROM scores were only slightly correlated with Maximum FM. These correlations explained little of the age-related differences in maximum FM (Table 2).

($p < 0.02$) even after strength, ROM, or both were included as covariates (Table 2). Thus, differences in strength and ROM also could not explain all of the differences in *orbital* dynamic stability between younger and older subjects.

4. Discussion

Younger adults can improve their dynamic stability by walking slower. However, while most older adults typically walk slower, they still fall more. Therefore, older adults may not improve their stability by walking slower, or other age-related factors may outweigh the benefits of walking slower. We addressed the first of these issues by directly comparing dynamic stability across multiple walking speeds in both healthy younger and older adults. We addressed the second issue in part by determining if age-related declines in lower extremity strength and/or flexibility could explain age-related differences found in walking stability.

The older adults tested in this study had PWS that were not different than those of the younger adults ($p = 0.86$) and thus exhibited no obvious decrements in functional capacity related to walking (Alexander, 1996). Nevertheless, these older adults still exhibited higher λ^* (Fig. 3) and maximum FM (Fig. 5) than the younger adults at all walking speeds. Thus, older adults were *more* sensitive to small intrinsic perturbations (i.e., more locally unstable) than younger adults, *independent* of walking speed. These age-related differences in dynamic stability were not explained by including strength or ROM as covariates (Figs. 4 and 6). This suggests that other age-related physiological changes, like increased neuromotor noise (Dean et al., 2007; Faisal et al., 2008), may be equally important as changes in strength and/or flexibility.

All subjects in this study walked under the same environmental conditions. Thus, any between-subject differences in stochastic inputs arose from differences in (internal) neuromotor noise and not (external) environmental noise. Indeed, these same older adults also exhibited greater movement variability than the younger adults (Kang and Dingwell, 2008). Conversely, however, diabetic neuropathic patients exhibited *less* local instability (Dingwell et al., 2000), in spite of also exhibiting greater variability (Dingwell and Cavanagh, 2001). We also recently determined how dynamic stability varied in a deterministic walking model where we could explicitly *control* the amplitude of the stochastic inputs. Increasing the noise inputs to this model had mixed effects on the local divergence curves and almost no effects on the maximum FM (Su and Dingwell, 2007). This suggests that age-related increases in neuromotor noise may also not entirely explain the increased maximum FM found in our older adults (Fig. 5B).

Both younger and older adults exhibited lower λ^* exponents at slower speeds and larger λ^* exponents at faster speeds (Fig. 3B). Both groups also exhibited slightly smaller maximum FM values at slower speeds and slightly larger values at faster speeds (Fig. 5B). This was in spite of the fact that these same subjects simultaneously exhibited *greater* kinematic variability at both faster *and* slower walking speeds (Kang and Dingwell, 2008). Orbital stability was less affected by changes in walking speed than local stability, similar to previous findings (Dingwell and Kang, 2007). Thus, all subjects demonstrated less sensitivity to small intrinsic perturbations when walking slower and greater sensitivity to these perturbations when they walked faster. This extends previous findings (Dingwell and Marin, 2006; England and Granata, 2007) by demonstrating that healthy older adults can also reduced their local dynamic instability by walking slower, inspite of increasing their variability (Kang and Dingwell, 2008).

Some studies suggested that physical limitations like muscle weakness (DeVita and Hortobagyi, 2000) and/or loss of flexibility (Kerrigan et al., 2001) cause older adults to walk slower. Others suggested older adults *choose* to walk slower to feel safer (Maki, 1997). While the older adults tested here were significantly weaker and less flexible than the younger adults (Table 1), they did not walk any slower as a result. Thus, our results support the second view, as these older adults also exhibited decreased local and orbital instability at slower walking speeds. Our older adults exhibited isometric leg strengths similar to other community-dwelling populations (Melzer et al., 2004), except for stronger knee flexors (75.1 ± 23.2 vs. $48.9 \pm 6.7 \, \text{N m}$). Greater strength declines than those observed here were associated with slower walking speed (Chandler et al., 1998). Therefore, our older subjects may have not yet lost enough strength to force them to walk slower. Reduced stability, strength and ROM seem to occur in healthy older adults *before* slower gait, which is a major clinical predictor of decreased motor function. More work is needed to determine if these changes in stability and ROM also predict future gait disorders.

When walking at the same speeds, some elderly adopt wider step widths to improve their lateral stability, at the expense of slightly increased energetic cost (Dean et al., 2007). Our older adults walked with step widths (Fig. 2) and step width variability (Kang and Dingwell, 2008) very similar to younger adults. Thus, it appears these older adults did not yet need to adopt this compensation either. Nevertheless, these elderly still exhibited significantly increased local (Fig. 3B) and orbital (Fig. 5B) instability. This suggests that these measures might be more sensitive indicators of locomotor impairment and potential future risk of falls than changes in step width and/or step width variability.

The λ^* exponents and maximum FM both quantify "local" stability: i.e., how systems respond to very small perturbations (i.e., "local" in state space). However, these two algorithms quantify this response differently (Dingwell and Kang, 2007; Dingwell et al., 2007). First, the maximum FM are local in space only because differences in stride time are ignored. Conversely, λ^*_S and λ^*_L are local in both space *and* time. Also, Floquet theory assumes an underlying limit cycle (in this case, the mean trajectory). All "perturbations" are taken relative to this limit cycle. Conversely, the λ^* analyses assume no such limit cycle. Initial "perturbations" to each trajectory are taken as the nearest neighbor to *that* trajectory. The positive λ^* demonstrate that, on average, each trajectory diverged away from its *own* nearest neighbor. However, since the maximum FM all had magnitude <1, all trajectories, on average, still converged *towards* the mean trajectory (Su and Dingwell, 2007).

This study only quantified responses to *local* (i.e., very *small*) perturbations. These results may or may not extend to *global* stability, where responses to large perturbations, like tripping (e.g., Pijnappels et al., 2008) or slipping, would be assessed. We do not yet know how much inherent local *in*stability humans can tolerate while remaining globally stable (Dingwell and Kang, 2007; Su and Dingwell, 2007). One important line of future research will be to determine where the boundaries lie between movements that remain globally stable and those that lead to global instability (i.e., falling).

Our subjects walked on a motorized treadmill, which may not properly reflect overground walking in older adults (Wass et al., 2005). Treadmills can artificially reduce the natural variability and local instability of walking by constraining walking speed (Dingwell et al., 2001). However, there were only minimal differences in orbital stability between treadmill and overground walking in younger adults (Dingwell and Kang, 2007). Here we tested all subjects under the same experimental conditions and relative to their own walking speeds, which were not different between the two groups. Therefore, repeating this study overground may yield slightly different values for each dependent measure. However, the observed differences between age groups and speeds are expected to remain.

In summary, even healthy active older adults who walk at the same preferred speeds as younger adults still exhibit significantly increased local and orbital dynamic instability, *independent* of walking speed. Both younger *and* older adults can *decrease* their local and orbital instability by walking slower, in spite of *increased* variability. Finally, decrements in strength and flexibility alone do not account for the increased instability observed in healthy older adults.

Conflict of interest

Acknowledgments

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