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Does local dynamic stability during unperturbed walking predict the response to balance perturbations? An examination across age and falls history

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ABSTRACT

Background: Older adults are at an exceptionally high risk of falls, and most falls occur during locomotor activities such as walking. Reduced local dynamic stability in old age is often interpreted to suggest a lessened capacity to respond to more significant balance challenges encountered during walking and future falls risk. However, it remains unclear whether local dynamic stability during normal, unperturbed walking predicts the response to larger external balance disturbances.

Research question: We tested the hypothesis that larger values of local dynamic instability during unperturbed walking would positively correlate with larger changes thereof due to optical flow balance perturbations.

Methods: We used trunk kinematics collected in subjects across a spectrum of walking balance integrity – young adults, older non-fallers, and older fallers – during walking with and without mediolateral optical flow perturbations of four different amplitudes.

Results: We first found evidence that optical flow perturbations of sufficient amplitude appear capable of revealing independent effects of aging and falls history that are not otherwise apparent during normal, unperturbed walking. We also reject our primary hypothesis; a significant negative correlation only in young adults indicated that individuals with more local dynamic instability during normal, unperturbed walking exhibited *smaller* responses to optical flow perturbations. In contrast, most prominently in older fallers, the response to optical flow perturbations appeared independent of their baseline level of dynamic instability.

Significance: We propose that predicting the response to balance perturbations in older fallers, at least that measured using local dynamic stability, likely requires measuring that response directly.

1. Introduction

Balance control, particularly that during locomotor activities like walking, is negatively affected by aging and disease. Given the tremendous functional consequences of falls, numerous techniques have emerged to quantify balance integrity for diagnostics and prevention. Of these techniques, local dynamic stability (*i.e.*, Lyapunov exponents) has garnered widespread use and significant scientific and clinical interest. For example, older adults (*versus* young) and people with neurodegenerative disease (*versus* controls) exhibit reduced local dynamic stability during walking [1,2]. These reductions are often interpreted to suggest a lessened capacity to respond to more significant balance challenges encountered during walking and future falls risk. However, local dynamic stability quantifies resilience to small, naturally occurring kinematic deviations arising normally during walking [3,4], and may not reflect resilience to larger external perturbations that could elicit a fall.

Determining the extent to which local dynamic stability during normal, unperturbed walking can predict one's resilience to external balance perturbations may be an especially timely goal for those in our aging population; recent evidence suggests that the rate of injurious falls among older adults is increasing [5]. Detecting a balance disturbance and executing the corrective motor responses that follow depends on integrating reliable sensory feedback [6,7]. Visual feedback, in particular, involves ~70% of the sensory neurons in the human brain during walking [8], can override other sensory modalities [9,10], and becomes the sensorimotor locus for balance control in older adults [11,12]. Perhaps consequently, perturbations of optical flow during

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walking yield more intense responses in older than in young adults [13,14]. These perturbations can also elicit age-related differences in local dynamic stability that are not otherwise apparent during unperturbed walking [15]. Hence, optical flow perturbations represent a promising paradigm to test for associations between local dynamic stability normally exhibited during walking and the response to larger, external balance challenges.

Older adults with a history of falls exhibit disproportionate changes in the integrity of balance control compared to non-fallers [16]. Indeed, compared to non-fallers, older fallers walk normally with increased movement variability (though, see [17]) and decreased local dynamic stability – the latter also retrospectively associated with the number of falls in the preceding year [18]. Accordingly, young adults, older non-fallers, and older fallers coalesce to provide a spectrum of walking balance integrity as quantified using local dynamic stability. However, prior work reporting response to optical flow perturbations has thus far: (*i*) excluded that in older adult fallers and (*ii*) focused on group-average comparisons that may confound relations between unperturbed walking and response to balance perturbations.

The purpose of this study was to investigate the extent to which local dynamic stability during normal, unperturbed walking predicts the response to optical flow perturbations. We hypothesized that larger values of local instability during unperturbed walking would positively correlate with larger changes thereof due to optical flow perturbations. We designed our recruitment plan to target a spectrum of walking balance integrity. Specifically, we tested our hypothesis across groups of young adults, older non-fallers, and older adults with a history of falls walking in a virtual environment with and without mediolateral optical flow perturbations. Accordingly, we tested the secondary hypothesis that aging and falls history increase local dynamic instability, particularly in the presence of optical flow perturbations.

2. Materials and methods

2.1. Subjects

Based on previously published results [15], we estimated our sample size to have 80% power to detect ($p < 0.05$): (*i*) a 21.5% reduction in local dynamic stability in older non-fallers due to small amplitude (20 cm) optical flow perturbations (n = 8 subjects), and (*ii*) between-group differences in local dynamic stability between young and older non-fallers walking in the presence of same small amplitude perturbations (*i.e.*, 1.38 ± 0.20 *vs.* 1.19 ± 0.10 bits·stride^{-1}) (n = 11 subjects/group). Thus, we recruited 11 young (5 males, 6 females), 11 older adult non-fallers (5 males, 6 females), and 11 older adults self-reporting at least one fall in the last year (4 males, 7 females). Table 1 summarizes their characteristics. Falls counting toward the self-reported total were defined according to the Kellogg International Work Group – unintentionally coming to the ground or some lower level and other than as a consequence of sustaining a violent blow, loss of consciousness, or sudden onset of paralysis [19]. We excluded subjects with BMI ≥ 30, sedentary lifestyle, orthopedic or neurological condition, or taking medication that causes dizziness. Subjects completed a falls efficacy scale to assess fear of falling (Table 1), and gave written informed consent according to the UNC Chapel Hill IRB.

Table 1
Subject characteristics.

	Young (A)	Non-fallers (B)	Fallers (C)	Main Effect
Age (yrs)	24.8 ± 4.8	75.3 ± 5.4^{A}	78.3 ± 7.6^{A}	$P < 0.001$
Mass (kg)	67.2 ± 8.8	73.4 ± 16.1	69.3 ± 14.0	$P = 0.553$
Height (m)	1.72 ± 0.09	1.75 ± 0.10	1.66 ± 0.12	$P = 0.159$
Falls Efficacy	–	10.3 ± 0.6	12.5 ± 3.0^{B}	$P = 0.028$

Mean ± standard deviation.

2.2. Protocol and data collection

We reanalyzed young adult data from a previously published study [20]. In that study, young adults walked on an instrumented, dual-belt treadmill (Bertec Corp., Columbus, OH) at 1.25 m s^{-1} – a speed not significantly different from their preferred overground walking speed (1.29 ± 0.18 m s^{-1}, $p = 0.527$). Here, older fallers (1.03 ± 0.22 m s^{-1}) and non-fallers (1.19 ± 0.20 m s^{-1}) walked on the same treadmill at their preferred overground speed, calculated from the average of two times taken to traverse the middle 3 m of a 10 m walkway. We note that these speeds differed significantly only between older fallers and young adults ($p = 0.007$). All subjects began by walking at their preferred walking speed on the treadmill for 5 min to become acclimated to the laboratory environment.

Subjects watched a speed-matched, immersive virtual hallway rear-projected onto a semi-circular screen surrounding the treadmill (1.45 m radius × 2.54 m height, Fig. 1). To the motion of the virtual hallway, we added continuous mediolateral oscillations of optical flow as a sum of three sine waves (0.125, 0.250, and 0.442 Hz), applied such that the foreground moved at full amplitude while the end of the hallway remained nearly stationary. These frequencies replicate those in prior studies and fall below those that would elicit discomfort, but enough to provoke instability [21]. In fully-randomized order, subjects walked for 2 min without perturbations and for 2 min each exposed to one of three amplitude continuous optical flow perturbations per condition (*i.e.*, 20, 35, and 50 cm). To enhance perturbation complexity that would be difficult for subjects to anticipate, the full amplitude was applied at 0.250 Hz, and half that amplitude was applied at 0.125 Hz and 0.442 Hz. All subjects wore a harness secured *via* an overhead support system. A 14-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA) operating at 100 Hz recorded the 3D trajectories of markers placed on subjects' heels, posterior sacrum, and C7 vertebrae (*i.e.*, a surrogate for upper torso motion used in prior studies that would be insensitive to head turns [22]).

2.3. Data analysis

Consistent with prior studies [1], we filtered marker trajectories using a 4th-order zero-lag low-pass digital Butterworth filter with cutoff frequency of 8 Hz, which preserved more than 99% of the power in the raw data (see Supplementary material for effects of cutoff frequency). We then quantified local dynamic stability by estimating the maximum exponential rates of divergence (*i.e.*, λ, bits·stride^{-1}) derived from 3D C7 marker velocity time series [4,23] (Eq. (1)). We used velocity, rather than position, time series to reduce signal non-stationarities that can compromise local dynamic stability estimates [1]. We used the following to construct the state space, $\boldsymbol{S}(t)$,

$$\boldsymbol{q}(t) = (\dot{x}, \dot{y}, \dot{z}) \tag{1}$$

$$\boldsymbol{S}(t) = [\boldsymbol{q}(t), \boldsymbol{q}(t+\tau), \boldsymbol{q}(t+2\tau), \cdots, \boldsymbol{q}(t+(d_E-1)\tau)] \tag{2}$$

where, $\dot{x}$, $\dot{y}$, and $\dot{z}$ correspond to C7 velocity components in the anterior-poster, mediolateral, and vertical directions, respectively. We then computed maximum rates of divergence of initially neighboring trajectories from $\mathbf{S}(t)$ using prescribed values for the embedding dimension (d_E) and time delay (τ) (Eq. (2)) [23]. We determined the embedding dimension ($d_E = 4$) using a false nearest neighbor algorithm and a 10% criterion [24,25]. By convention, we prescribed the time delay (τ) as one quarter of subjects' average stride time for each condition [26], calculated as the average duration between successive heel strikes [27]. We time normalized the divergence curves to account for differences in stride period and then calculated each subject's maximum short-term λ_S (0–1 stride) and long-term λ_L (4–10 strides) divergence exponents for each condition. Here, larger positive values of λ signify larger local dynamic instability. Finally, we determine the sensitivity of these divergence exponents to the selection of embedding dimension