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Aging effects on leg joint variability during walking with balance perturbations

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ABSTRACT

Background: Older adults are more susceptible to balance perturbations during walking than young adults. However, we lack an individual joint-level understanding of how aging affects the neuromechanical strategies used to accommodate balance perturbations.
Research question: We investigated gait phase-dependence in and aging effects on leg joint kinematic variability during walking with balance perturbations. We hypothesized that leg joint variability would: 1) vary across the gait cycle and 2) increase with balance perturbations. We also hypothesized that perturbation effects on leg joint kinematic variability would be larger and more pervasive in older versus young adults.
Methods: We collected leg joint kinematics in young and older adults walking with and without mediolateral optical flow perturbations of different amplitudes.
Results: We first found that leg joint variability during walking is gait phase-dependent, with step-to-step adjustments occurring predominantly during push-off and early swing. Second, young adults accommodated perturbations almost exclusively by increasing coronal plane hip joint variability, likely to adjust step width. Third, perturbations elicited larger and more pervasive increases in all joint kinematic outcome measures in older adults. Finally, we also provide insight into which joints contribute more to foot placement variability in walking, adding that variability in sagittal plane knee and coronal plane hip joint angles contributed most to that in step length and step width, respectively.
Significance: Taken together, our findings may be highly relevant to identifying specific joint-level therapeutic targets to mitigate balance impairment in our aging population.

1. Introduction

Balance control during walking depends on integrating reliable sensory feedback to plan and execute corrective motor responses [1]. These motor responses presumably manifest first at the individual muscle and joint levels and contribute to coordinated balance corrections at the whole-body level *via* adjustments in foot placement from one step to the next. Motivated by these ideas, we recently revealed that variability of foot placement (*i.e.*, step width and step length) in older adults is more susceptible than in young adults to optical flow perturbations during walking, presumably governed by an age-related dependence on visual feedback for motor planning and execution [2–6]. However, foot placement variability is determined by step-to-step adjustments in the angles of the leg joints. Our understanding of which joints contribute more to foot placement variability and when during the gait cycle those joint-level adjustments occur, particularly in the presence of balance perturbations, is fundamentally incomplete.

Although average profiles of leg joint kinematics are ubiquitous in gait biomechanics research, there are surprisingly few studies describing gait phase-dependent changes in step-to-step variability at the joint level. However, it is highly intuitive that leg joint kinematic variability would exhibit gait phase-dependence, increasing during the gait cycle when step-to-step adjustments in hip, knee, and ankle joint kinematics are most prevalent. For example, regulating step length and step width during step-to-step transitions is critical for walking balance control, alluding to the potential for particularly large variability in leg joint kinematics [7]. Indeed, local dynamic stability, at least of sagittal plane knee joint kinematics, exhibits local maximums during heel strike and toe-off events in walking [8]. However, we currently lack empirical evidence for whether leg joint variability similarly exhibits phase-dependence across the gait cycle, even in healthy young adults.

Compared to young adults, older adults often walk with greater variability of foot placement (*e.g.*, step width and step length) than young adults [9], and this variability can be retrospectively associated

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with falls history [10]. However, accompanying highly prevalent somatosensory changes with aging [11–16], there is a well-documented dependence on vision for movement and balance control that are even more pronounced in older adults with a history of falls [3–6,17]. Accordingly, we have shown that optical flow perturbations applied during walking (*i.e.*, the visual self-perception of walking imbalance) can elicit larger age-related decrements in foot placement variability than during unperturbed walking [5,18]. These findings imply that optical flow perturbations could provide a unique opportunity to study how step-to-step adjustments in step width and step length are regulated *via* those in hip, knee, and ankle joint angles in the context of age-related changes in walking balance. Ultimately, understanding these joint-level adjustments and their relation to whole-body balance corrections *via* step width and step length could point to specific joint-level therapeutic targets to mitigate falls risk.

Therefore, this study aimed to investigate gait phase-dependence in and the effects of age on leg joint kinematic variability during walking with and without optical flow perturbations of different amplitudes. We hypothesized that leg joint kinematic variability would: 1) vary across the gait cycle (*i.e.*, exhibit gait phase-dependence), and 2) increase in the presence of optical flow perturbations. We also hypothesized that 3) compared to young adults, the effects of optical flow perturbations on leg joint kinematic variability would be larger and more pervasive in older adults. Finally, the second goal of this study was to determine how step-to-step adjustments in hip, knee, and ankle joint angles contribute to those in step length and step width in old and young subjects − a question that informed a series of linear regressions between foot placement variability and leg joint variability.

2. Materials and methods

2.1. Subjects

We recruited 11 healthy young (mean ± standard deviation, age: 24.8 ± 4.8 yrs, mass: 67.2 ± 8.83 kg; height: 172 ± 9 cm, 5 males, 6 females) and 11 healthy older adults (age: 75.3 ± 5.4 yrs, mass: 73.4 ± 16.1 kg, height: 175 ± 10 cm, 5 males, 6 females) to participate in this study. All subjects had normal or corrected to normal vision. Based on a health questionnaire, we excluded subjects based on the following: BMI ≥ 30, sedentary lifestyle, orthopedic or neurological condition, taking medication that causes dizziness, or any falls in the prior year. All subjects signed written informed consent according to the UNC-Chapel Hill Institutional Review Board, and none reported any adverse events.

2.2. Experimental protocol and data collection

Previously published young adult data were reanalyzed for comparison to older adults [19]. In that study, young subjects completed testing on an instrumented dual-belt treadmill (Bertec Corp., Columbus, OH) at 1.25 m s^{-1}, a speed comparable to their preferred overground walking speed (*i.e.*, 1.29 ± 0.18 m s^{-1}). Here, older subjects walked on the same treadmill at their preferred overground walking speed (1.19 ± 0.20 m s^{-1}). Despite this subtle difference in design, tested treadmill speeds were not significantly different between older and young adult subjects ($p = 0.15$).

All subjects watched a speed-matched, immersive virtual hallway rear-projected onto a semi-circular screen (1.45 m radius × 2.54 m height) surrounding the treadmill. To the motion of the hallway, we added continuous mediolateral oscillations of optical flow as a sum of three sine waves (0.125, 0.250, and 0.442 Hz) [19], applied such that the foreground moved at full amplitude while the end of the hallway remained stationary. These frequencies replicate those used in prior studies [19,20] and facilitate complex mediolateral oscillations that would be difficult for subjects to anticipate. In fully-randomized order, subjects walked for 2 min each, both with and without optical flow perturbations prescribed by three amplitudes (*i.e.*, 20, 35, and 50 cm). To further increase perturbation complexity, these amplitudes were applied such that the full amplitude was applied at 0.250 Hz, and half that amplitude was applied at 0.125 Hz and 0.442 Hz. Finally, all subjects wore a harness secured *via* an overhead support system. A 14-camera, three-dimensional motion capture system (100 Hz, Motion Analysis Corp., Santa Rosa, CA) recorded the trajectories of 33 retroreflective markers and marker clusters placed on the right and left foot, shank, and thigh and on the pelvis for all conditions.

2.3. Data analysis

We filtered marker trajectories using a 4th-order zero-lag low-pass Butterworth filter with a cutoff frequency of 6 Hz. Using a standing calibration trial, and right and left hip joint centers derived from a leg circumduction task [21], we scaled a 7-segment, 18 degree-of-freedom (DoF) model of the pelvis and legs [22], including three DoFs at each hip, one DoF at each knee, and one DoF at each ankle. We then calculated hip, knee, and ankle joint kinematics using a global optimization inverse kinematics routine that minimized the weighted sum of squared differences between measured and modeled marker positions [23]. Specifically, we calculated time series of sagittal plane hip, knee, and ankle joint angles, and coronal plane hip joint angles. The instants of right and left heel strikes were determined by detecting the peak anterior positions of the heel marker relative to the sacrum marker [24]. The time series of leg joint angles were time-normalized and interpolated across successive heel strikes. We calculated stride-averaged profiles and the variability of leg joint kinematics – the latter using the standard deviation computed at 10% intervals across the gait cycle (*e.g.*, that at 5%, 15%, 25%, *etc.*).

Finally, we calculated time series of step length and step width. Specifically, we calculated step lengths as the relative anterior-posterior position of consecutive heel markers at heel strike plus the treadmill belt translation during that step. We calculated step widths as the mediolateral distance between consecutive heel positions at heel strike. From their time series, we calculated mean step length and step width and their respective variabilities – the latter reported as the standard deviation.

2.4. Statistical analysis

We used a mixed design 3-way factorial ANOVA to determine significant effects of main factors and their effect sizes (Gait phase: 10% bins; Perturbations: 0 *vs*. 20, 35, and 50 cm; Age: young *vs*. older) [25]. We used a Greenhouse-Geisser correction for outcome measures that violated the sphericity assumption [26]. We sought to be conservative in our post-hoc analysis and minimize the number of pair-wise comparisons. Thus, for those outcome measures with a significant perturbation main effect, we focused our post-hoc pairwise comparisons on elucidating the group and gait cycle phase in which those effects occurred. Finally, we used linear regressions to calculate correlations between step length variability and sagittal plane hip, knee, ankle joint angle variability and between step width variability and coronal plane hip joint angle variability, all at the instant of heel strike. Statistical analyses were coded in MATLAB (R2017a, MathWorks Inc., Natick, MA) using an alpha level of 0.05.

3. Results

3.1. Gait cycle phase and age group effects

During unperturbed walking, hip (Figs. 1 and 2), knee (Fig. 3), and ankle (Fig. 4) joint angle variabilities differed significantly across different phases of the gait cycle (main effect, p's < 0.001), and all reached their maximum following the instant of ipsilateral toe-off (*i.e.*, ~60–70% cycle). Compared with young adults, older adults walked