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Gait adaptations of older adults on an uneven brick surface can be predicted by age-related physiological changes in strength

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ABSTRACT

Background: Outdoor falls in community-dwelling older adults are often triggered by uneven pedestrian walkways. It remains unclear how older adults adapt to uneven surfaces typically encountered in the outdoor built-environment and whether these adaptations are associated to age-related physiological changes.
Research question: The aims of this study were to (1) compare gait parameters over uneven and flat brick walkways, (2) evaluate the differences between older and young adults for these two surfaces, and (3) assess if physiological characteristics could predict adaptations in older adults.
Methods: Balance, strength, reaction-time, full-body marker positions, and acceleration signals from a trunk-mounted inertial measurement unit were collected in seventeen older (71.5 ± 4.2 years) and eighteen young (27.0 ± 4.7 years) healthy adults to compute lower-limb joint kinematics, spatio-temporal parameters, dynamic stability, and accelerometry-derived metrics (symmetry, consistency, and smoothness).
Results: Both groups increased hip flexion at foot-strike, while decreasing ankle dorsiflexion, margin of stability, symmetry, and consistency on the uneven, compared to flat, surface. Older, compared to young, adults showed a larger increase in knee flexion at foot-strike and a larger decrease in smoothness on the uneven surface. Only young adults decreased hip abduction on the uneven surface. Strength, not balance nor reaction-time, was the main predictor of hip abduction in older adults on both surfaces.
Significance: While older adults may be especially vulnerable, uneven surfaces negatively impact gait, irrespective of age, and could represent a risk to all pedestrians.

1. Introduction

Falls are a major public health concern given our world's aging population [1]. In the United-States, falls occur in more than 30% of community-dwelling older adults [2,3] and are the leading cause of traumatic injury-related death [4]. More than 50% of falls in community-dwelling older adults occur outside the home and are mainly precipitated by uneven surfaces (47.6%) [5].

Studying gait biomechanics in physiologically challenging situations may allow for increased sensitivity to identify age-related changes in walking [6]. One such physiologically challenging situation may be walking on uneven surfaces. Menz et al. [7] reported increased step-time variability, decreased gait speed, and decreased step-length for healthy older, compared to young, adults walking on an uneven surface, while others [8] observed that older adults decreased their stride-width on an uneven, compared to flat, surface. In older women with diabetic peripheral neuropathy, stride-width was found to increase with age on an uneven surface [9]. These studies constructed uneven surfaces from wooden blocks randomly placed beneath a malleable surface [6–9] which may not adequately represent real-world walking surfaces, and focused on spatio-temporal parameters. It is unclear if similar adaptations occur on surfaces more analogous to those found on pedestrian walkways and if modifications in other measures, such as joint kinematics, dynamic stability [10], and accelerometry-derived metrics [11] also present. Finally, associating the effects of uneven surfaces on gait parameters to physiological function may help determine the factors limiting safe locomotion on these challenging surfaces [7] and could direct future interventions.

The aims of this study were to (1) compare gait parameters over flat and uneven brick walkways, (2) evaluate the differences between older

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and young adults over these surfaces, and (3) determine if physiological characteristics could predict adaptations observed in the older adult group. We hypothesized that (1) gait parameters would differ across uneven and flat surfaces, (2) unique age-specific adaptations, consistent with a conservative and careful gait pattern, would be observed in older adults, and (3) these adaptations would correlate with balance, reaction-time, lower-limb strength, and/or age itself.

2. Methods

2.1. Participants

Forty-six community-dwelling young (18–35 years) and older (65+ years) adults were recruited and provided written informed consent before participating in this study. Ten participants were excluded after a medical screen revealed neurological disorders, vertigo, dizziness, lower-limb musculoskeletal abnormalities (joint replacement, pain, fractures within the last two years), diabetes, or obesity (body mass index > 30). Participants reporting additional conditions (cataracts, glaucoma, osteoarthritis, osteopenia, osteoporosis, or rheumatoid arthritis) were retained if they scored above the age-matched lower range of values [12] in balance ability [13]. One older adult was excluded based on this criterion. Thus, data from seventeen older (12 females, 71.5 ± 4.2 years, 165.7 ± 9.3 cm, 67.6 ± 12.6 kg) and eighteen young (8 females, 27.0 ± 4.7 years, 171.6 ± 8.8 cm, 69.5 ± 14.7 kg) healthy adults were analyzed. The Harvard Institutional Review Board approved this study.

2.2. Data collection

Participants completed the study within a single session. The miniBESTest was used to assess balance [13]. The miniBESTest is a clinical test of dynamic balance comprised of 14 items with a maximum score of 28 [13]. Reaction time was evaluated as the time delay between 10 emitted audio-cues and the depression of the space key recorded by a computer-based program. Strength was assessed using the average peak torque normalized to body mass via three isokinetic maximum effort knee extensions at 60°/s (Biodex Medical Systems Inc., Shirley, USA). A standardized warm up was provided. Additional sets were recorded if the coefficient of variation exceeded 15% [14]. A single leg was analyzed per subject using a counter-balanced approach. For the gait assessment, an eighteen-camera motion-capture system (Motion Analysis Corp., Santa Rosa, USA) tracked retro-reflective markers placed according to a modified Plug-in Gait model (Vicon Motion Systems Ltd., Oxford, UK) while participants performed a series of walking trials at self-selected speed over flat and uneven brick walkways (Fig. 1). Static trials, in which the participants stood motionless in the anatomical position, were also recorded. Motion capture was sampled at 100 Hz. An Opal inertial measurement unit (APDM, Inc. Portland, USA) sampling at 128 Hz with a ± 6 g sensing range, affixed to the trunk (L5 vertebra), was synced with the motion capture system. Participants wore standardized athletic shoes (Nike Inc., Beaverton, USA) and were tethered to a safety harness (Fig. 1c). Presentation of surface was counter-balanced across participants.

2.3. Data processing and analysis

Marker data were labeled and gap filled in Cortex (Motion Analysis Corp., Santa Rosa, USA) for four trials per condition and participant. All files were subsequently imported into Matlab (v2016b, The Mathworks Inc., Natick, USA) for processing using biomechZoo [15] and custom code. Marker data were filtered using a 4th order low-pass Butterworth filter (8 Hz cut-off frequency). The location of the hip [16], knee, and ankle [17] joint centers were computed and orientation of thigh wand markers were optimized [18] in anticipation of kinematic computations [19]. Walking trial knee angle offsets were corrected using the static trial knee angles. Timing of foot-strike and foot-off events were estimated from marker data according to the coordinate-based algorithm of Zeni et al. [20]. Events were used to partition trials and to compute spatio-temporal parameters: stride-velocity, double-support time, stride-length, and stride-width. The perpendicular distance between the extrapolated center of mass and the line joining the ankle and additional distal fifth metatarsal marker was used to compute the minimum medio-lateral margin of stability for each foot (Eq. A.2 in [10]). Discrete events at foot strike during a single gait cycle for hip flexion, hip adduction, knee flexion, and ankle dorsiflexion were extracted. For the trunk accelerometric measures, the resultant acceleration was computed from the vertical, medio-lateral, and antero-posterior accelerations over seven steps. Step-to-step symmetry and stride-to-stride consistency were quantified by step-regularity and stride-regularity, respectively, computed from the unbiased autocorrelation procedures described by Moe-Nillsen et al. [21]. Movement smoothness was calculated from the acceleration spectral arc length (SPARC), in which increasingly large negative values represent less smooth movement [22]. Raw time-series kinematic and acceleration data are presented in Supplemental Figs. 1 and 2, respectively.

2.4. Statistical analysis

The difference in male-female representation across groups was evaluated via Fischer's exact test. Anthropometric, balance, reaction-time, and strength differences across groups were assessed via unpaired *t*-tests (parametric) or Wilcoxon rank-sum tests (non-parametric). The mean value of each dependent variable for each participant and surface was computed.

To test hypotheses 1 and 2, examining the effects of surface and age, 2 × 2 mixed analyses of variance (ANOVAs) with within-subject factor surface (flat, uneven) and between-subject factor age (dichotomous: young, old) were performed. A surface x age interaction term was included to test the second hypothesis that the effects of the surface differed for older adults. Partial Eta-Squared (η^2) measured effect sizes for all ANOVA tests. In the case of a significant interaction, follow up paired comparisons (young/flat vs young/uneven and old/flat vs old/uneven) were evaluated using paired *t*-tests (parametric assumptions satisfied). Effect size was estimated via Cohen's d (d) [23]. Otherwise, main effects of surface and of age were reported. Parametric assumptions were verified using QQ-plots and Levene's test.

To test the third hypothesis that gait adaptations in older adults are related to physiology, stepwise linear regressions for dependent variables in which significant surface × age interactions were observed for the ANOVAs were constructed using balance, reaction-time, strength, and age (continuous) as predictor variables. Stepping method entered or removed predictors based on a p-value threshold of 0.05 and 0.10, respectively. The variance inflation factor was used to test multicollinearity between predictors before running the regression models. All predictors were retained based on a low variance inflation factor (≤2.64) [24].

Significance was set at $\alpha = 0.05$ for all tests. ANOVAs and linear regressions were run in SPSS (v23, IBM Corp., Armonk, USA). Remaining statistical tests were evaluated in Matlab (v2016a, The Mathworks, Inc. Natick, USA).

3. Results

There were no differences in male-female representation across groups ($p = .176$), height ($p = .062$), nor weight ($p = .675$); however, the older adults were significantly older and exhibited decreased strength (1.00 ± 0.23 vs 1.44 ± 0.20 Nm/kg) and lower balance (24.1 ± 2.3 vs. 27.5 ± 0.5) ($p < .001$, rank-sum test), compared to young adults. Reaction times were not significantly increased in older (0.39 ± 0.16 s) compared to young (0.32 ± 0.11 s) adult participants ($p = .127$). Self-selected mean gait speed was invariant across surfaces