11-2022

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Publication Information


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Musculoskeletal Adaptation of Young and Older Adults in Response to Challenging Surface Conditions

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Keywords: Gait; challenging surfaces; aging, falls; electromyography

Word count: Abstract - 225
Introduction to Discussion – 3,529
Abstract:

Over 36 million adults over 65 years of age experience accidental falls each year. The underlying neuromechanics (whole-body function) and driving forces behind accidental falls, as well as the effects of aging on the ability of the musculoskeletal system to adapt, are poorly understood. We evaluated differences in kinematics (lower extremity joint angles and range of motion), kinetics (ground reaction force), and electromyography (muscle co-contraction), due to changes in surface conditions during gait in 14 older adults with a history of falling and 14 young adults. We investigated the impact of challenging surfaces on musculoskeletal adaptation and compared the mechanisms of adaptation between age-groups. Older adults displayed greater hip and knee flexion and range of motion during gait, reduced initial vertical loading, and 13% greater knee muscle co-contraction during early stance compared to young adults. Across age groups, the presence of an uneven challenging surface increased lower-limb flexion compared to an even surface. On a slick surface, older adults displayed 30% greater ankle muscle co-contraction during early stance as compared to young adults. Older adults respond to challenging surfaces differently than their younger counterparts, employing greater flexion during early stance. This study underscores the need for determining lower-limb musculoskeletal adaptation strategies during gait and assessing how these strategies change with age, risk of accidental falls, and environmental surfaces to reduce the risk of accidental falls.
1. Introduction

Accidental falls present a large financial and functional burden among older adults (over 65 years of age). Every year, one in three older Americans experience an accidental fall (Ambrose et al., 2013; Homer et al., 2017). Non-fatal accidental fall medical costs totaled $121.5 billion (6.8 million falls) in 2020 (CDC, 2021). Fall-related injuries and fear of falling decrease quality of life, function, and independence for older adults (Berry and Miller, 2008; Nevitt et al., 2016). Older adults must increase lower-limb muscle co-contraction to maintain stability and decrease fall risk (DaSilva et al., 2021; Hortobágyi and Devita, 2000). While elevated co-contraction assists with stabilization, it may also increase the risk of joint damage and metabolic cost of walking, leading to muscular fatigue and elevated fall risk (Hahn et al., 2005; Lee et al., 2017). Muscular fatigue and decreased muscle strength in older adults (Saywell et al., 2012) may also lead to decreased lower-limb sagittal range of motion (ROM) (Soucie et al., 2011).

Almost half of older adult accidental falls occur walking outdoors (DaSilva et al., 2021; Li et al., 2006). Considering almost three-quarters of these falls happen when walking on a “challenging” surface (Allin et al., 2016; Li et al., 2006), understanding how the musculoskeletal system adapts to environmental conditions is crucial to reduce the risk of accidental falls. Locomotion on uneven or slick surfaces presents complex neuromuscular demands, which can include increased lower-limb sagittal-plane ROM, flexion angles at initial contact, and loading rate of the stance limb during early stance (Ippersiel et al., 2022; Voloshina et al., 2013; AminiAghdam and Blickhan, 2018; Dixon et al., 2018). Challenging surfaces are also associated with increased knee and ankle muscle co-activation, functioning as a stiffening strategy during single leg support (Schmitz et al., 2009; Voloshina et al., 2013).

Few studies to date have integrated neuromechanical data – joint kinematics, kinetics, and muscle activation (Lencioni et al., 2019; Moreira et al., 2021) – for both young (Ippersiel et al., 2022) and older adults (Eckardt and Rosenblatt, 2018; Menant et al., 2009) to assess adaptation and stability during stance due to challenging surfaces. Stability is particularly important during loading response, where the stance limb absorbs the shock from weight acceptance, and midstance, to preserve forward momentum (Perry, 1996). Change in vertical ground reaction force (vGRF) from initial contact through peak vGRF, and the resultant vGRF impulse (force acting over time) during this loading response, must be absorbed via muscle co-contraction and
joint loading (Saywell et al., 2012); however, the age-related differences in shock absorption during gait are not well understood when navigating challenging surfaces.

Identifying specific neuromuscular differences between older and younger adults may assist clinicians in developing targeted strengthening strategies for the older population. The aims of this study were to quantify and compare lower-limb neuromechanics for young and older adults walking across challenging surfaces (uneven and slick). We hypothesized that older adults would demonstrate reduced lower-limb sagittal-plane ROM, greater muscle activation, and larger vGRF resulting in increased impulse during loading response and greater muscle activation during midstance, compared to young adults. Additionally, we hypothesized that participants would demonstrate increased lower-limb sagittal-plane kinematics (flexion at initial contact and stance ROM) and muscle co-activation, and decreased vGRF (initial force and impulse) when encountering challenging surfaces.

2. Methods

2.1. Participants

Fourteen older (> 65 years old) and 14 young (18-25 years old) adults recruited from the local community participated in this study. Older participants, recruited from the Osher Lifelong Learning Institute, were screened consecutively via email correspondence for self-reported ability to perform study tasks and presence of one or more accidental falls in the past 12 months (Rispens et al., 2015), and absence of any neurological or musculoskeletal disease or surgery that would alter their kinematics. Older adults at risk of falling were selected to maximize the potential to observe the largest differences in neuromuscular adaptation (Sadeghi et al., 2021) to inform targeted interventions. Power analysis of preliminary knee flexion indicated 14 participants per cohort were needed to detect 5-degree differences in flexion angle (90% statistical power, α=0.05). Exclusion criteria for all participants consisted of: history of surgery, current (< 6 months) pain/injury for lower extremity or back, pregnancy, known neurological disorders, or inability to balance single-legged for 20 seconds. All participants provided informed consent to a protocol approved by the local Institutional Review Board. Participants were matched across cohorts according to sex, height, and weight (Table 1).
2.2. Data Collection

During neuromechanical testing, participants performed a walk task on three surfaces – normal, uneven, and slick. For this task, participants walked at a self-selected speed through the motion capture volume and contacted a force-plate mounted modular surface with their dominant limb (Figure 1). Prior to testing, the self-selected speed for each participant was determined as the average time (from 5-10 trials) taken to walk at a comfortable pace between two pairs of timing gates (1.83 m apart). The uneven surface was composed of nine squares of differing heights attached to the modular surface. The slick surface consisted of nine squares of differing heights attached to the modular surface. The slick surface consisted of slide board (American Lifetime, Miami Beach, FL) mounted on the modular surface. Participants wore Lycra booties over their shoes to produce a coefficient of friction, $\mu$, of 0.19, like ice ($\mu=0.1$, (College, 2012)). Participants performed three successful walk trials on each surface. A successful trial required the participant to walk ±5% of their preferred speed and only contact the surface (i.e., force platform) with their dominant limb. Dominant limb was identified as the leg used to kick a soccer ball. During testing, all participants wore a safety harness to prevent a fall-related injury. Prior to testing, the order of surfaces was randomized for each participant using a Latin Square Design.

2.3. Biomechanics Data and Analysis

For each walk trial, synchronized motion capture and ground reaction force (GRF) data were collected for three-dimensional lower-limb biomechanics analysis. Ten high-speed optical cameras (Vantage, Vicon Motion Systems Ltd., Oxford, UK) tracked trajectories (240 Hz) from 50 retroreflective markers (see Appendix) while one force platform (OR6, Advanced Mechanical Technology Inc., Watertown, MA) recorded GRF data (2400 Hz). Marker trajectories and GRF data from each trial were low-pass filtered using 4th-order Butterworth filter (12 Hz) in Visual3D (v6.0, C-Motion, Rockville, MD) and exported for biomechanical analysis.

The filtered vGRF data were exported into Python (3.8.5) for processing. Before analysis, the vGRF for each trial and participant were normalized to body weight (BW) and divided into functional gait phases (initial contact (IC), loading response (LR), and midstance (MS)) during
stance, according to Perry and Burnfield (2010). Stance phase was identified as the period between vGRF first exceeding and then falling below 20 N (heel-strike and toe-off, respectively). Specifically, IC occurred at 0% stance (vGRF > 20N, Kawaji and Kojima, 2019), LR was from IC to the first vGRF peak, and MS was from LR to the local vGRF minima around 50% stance (Perry and Burnfield, 2010). For analysis, vGRF (% BW) at the bounds of each gait phase of interest were calculated as common “snapshots” across stance to compare between cohorts, versus general percentages of stance which might not correspond to the same gait events and musculoskeletal demands across trials. Vertical loading rate (slope of vGRF from IC to 80% LR (Larsen et al., 2008)) and impulse during LR (area under the vGRF curve from IC to the end of LR) were also calculated.

Filtered marker and GRF data were exported into OpenSim 4.1 to create subject-specific musculoskeletal models for kinematic analysis. A generic lower-limb and torso model with 23 degrees of freedom and 92 musculotendon actuators (gait2392_simbody.osim) was scaled for each participant to match their experimental data (Delp et al., 2007). Inverse kinematic analysis was applied to the scaled model to quantify sagittal-plane hip, knee, and ankle joint angles during stance phase for each trial. Flexion angle, as well as ROM (peak minus minimum) during LR and stance phases, were calculated at the bounds of each gait phase of interest to compare common “snapshots” across stance.

2.3. Electromyography Data and Analysis

During each walk trial, electromyography (EMG) data recorded each participant’s dominant limb muscle activity. EMG data (2400 Hz) were collected with a common-mode rejection ratio of 80 dB, amplified using a 16-channel EMG system (Trigno, Delsys, Inc., Natick, MA, gain 1000), and synchronized with GRF data via the motion capture system. Surface electrodes (DE 2.1 single-differential, parallel-bar configuration, 99.9% Ag) with 10 mm inter-electrode distance were positioned over the muscle bellies and in line with the muscle fibers of the following thigh and shank muscles – tibialis anterior (TA), gastrocnemius lateralis (GL), vastus lateralis (VL), rectus femoris (RF), vastus medialis (VM), semimembranosus (SM) and biceps femoris (BF) according to Hermens et al. (2000). Before electrode placement, the participant’s skin was abraded and cleaned with alcohol to reduce impedance.
Collected EMG data were filtered using a 2nd-order Butterworth bandpass filter (20-500 Hz) to remove motion artifacts, and then full-wave rectified and low pass filtered (10 Hz) with a 2nd-order bidirectional Butterworth filter to create linear envelopes for analysis (Robertson et al., 2013). The linear envelope for each muscle was normalized to muscle-specific peak activity during all successful walk trials (DaSilva et al., 2021; Saywell et al., 2012). Simultaneous agonist/antagonist muscles co-contraction at the knee and ankle were found for each gait phase (LR and MS) as well as across stance according to Winter (2009):

\[
\% COCON = 2 \times \frac{Common(A,B)}{Area(A) + Area(B)} \times 100\%
\]

where Area(A) was the area of the EMG activation profile for either quadriceps (sum of VL, VM, and RF) or TA, Area(B) was the area for hamstring (sum of BF and SM) or GL, and Common(A,B) was the area under the EMG curve common to both muscle groups.

2.6. Statistical Analysis

R 4.0.3 was used for all statistical analyses. Participant demographics, including age, height, weight, and walking speed, were submitted to independent t-tests to identify cohort differences. The dependent vGRF, joint kinematic, and muscle co-contraction variables were assessed for normality (Shapiro-Wilk Test) and then submitted to two-way mixed model analysis of variance (rstatix package, Kassambara (2020); 25 total) to test the main effects of, and interactions between, age (young and older) and surface (normal, uneven, and slick). Significant main effects and interactions were submitted to Dunnett test multiple comparisons (Andri et al., (2020)) to determine changes from the normal surface, and Tukey tests to identify differences in adaptation strategies between challenging surfaces. An alpha level of p < 0.05 was set a priori to denote statistical significance.

3. Results

3.1. Participant Demographics

There was a significant difference in age (p < 0.001), but not height, weight, or walking speed (p>0.05) between cohorts (Table 1).
3.2. Ground Reaction Force

There was a significant age by surface interaction for loading impulse (p = 0.013, Table 2). The older adults reduced loading impulse on the uneven (p < 0.001), but not normal or slick, surface (p>0.05) compared to young adults. The older adults reduced loading impulse on the uneven, compared to normal and slick, surface (both: p < 0.001), but young adults exhibited no impact of surface on loading impulse.

Older adults exhibited significantly smaller vGRF at the end of LR compared to young adults (main effect (ME): p =0.024, Table 2). Surface impacted vGRF at the end of LR (ME: p =0.047), but no significant differences between surfaces were observed with post-hoc comparisons.

3.3. Kinematics

Ankle Plantarflexion

The ANOVA revealed a significant age by surface interaction for IC ankle flexion (p = 0.015, Table 3). Older adults exhibited greater IC ankle plantarflexion on the uneven, compared to normal, surface (p=0.048), but differences were not observed between any other surface (p>0.05).

Age did not impact ankle flexion during stance (all ME: p>0.05, Table 3). Surface impacted ankle flexion LR ROM as well as at the end of LR (both ME: p<0.001) and MS (ME: p =0.024, Figure 2). On the uneven surface, participants decreased LR ROM and increased ankle dorsiflexion at the end of LR compared to both normal (p < 0.001; p =0.002) and slick (p < 0.001; p =0.039) surfaces, respectively. After correcting for type I error, no significant differences between surfaces were observed for ankle flexion at the end of MS.

Knee Flexion/Extension

The ANOVA revealed a significant age by surface interaction for MS knee flexion (p=0.022, Table 3). Older adults exhibited significantly greater knee flexion on the uneven, compared to
normal, surface (p=0.042) at the end of MS. Further, on both the normal (p =0.018) and uneven (p < 0.001), but not slick, surfaces older adults increased knee flexion at the end of MS, compared to young adults (Figure 3).

Older adults increased knee flexion at IC (ME: p=0.03), at the end of LR (ME: p=0.001), and ROM during LR (ME: p=0.005) compared to young adults. Knee flexion at IC (ME: p<0.001) increased on the uneven compared to normal and slick surfaces (both: p < 0.001) and increased at the end of LR (ME: p<0.001) on the uneven compared to the normal surface (p=0.007). Knee flexion ROM (ME: p<0.001) decreased on uneven compared to both normal (p<0.001) and slick (p=0.020) surfaces (Figure 2).

**Hip Flexion/Extension**

The ANOVA revealed a significant age by surface interaction for MS hip flexion (p=0.008, Table 4). Older adults exhibited significantly greater hip flexion on the uneven, compared to normal, surface (p=0.023). Further, older adults exhibited greater hip flexion than young adults at the end of MS on both the normal (p=0.040) and uneven (p=0.003) surfaces.

Older adults demonstrated greater hip flexion than young adults at the end of LR (ME: p=0.036, Table 3). Hip flexion at IC (ME: p<0.001) increased on the uneven, compared to normal (p<0.001) and slick (p=0.008), surface and increased at the end of LR (ME: p<0.001) on the uneven, compared to normal, surface (p=0.028). Hip flexion RIM (ME: p<0.001) decreased on the uneven, compared to normal and slick, surface (both: p<0.001, Figure 2). A surface main effect was found for hip LR ROM (ME: p<0.001), but no post-hoc comparisons between surfaces were significant.

**3.4. Muscle Activation**

The ANOVA revealed a significant age by surface interaction for ankle co-contraction during MS (p=0.028, Table 4). We found that only the slick, not normal or uneven, surface increased ankle co-contraction during MS for older, versus young, adults (p=0.004, Figure 4, right).

Older adults exhibited more knee co-contraction than young adults during LR (ME: p=0.030, Table 4 and Figure 4). While surface main effects were found for knee (ME: p=0.004) and ankle
(ME: p=0.008) co-contraction during MS and LR, respectively, no significant differences between surface were observed with post-hoc comparisons for either joint or phase of stance.

4. Discussion

Older adults adopt compensatory mechanisms during gait. Additionally, the presence of a challenging surface (slick or uneven) impacts musculoskeletal adaption. Finally, older adults respond differently to a challenging surface than younger adults. The clinical implications of these findings are that targeted physical therapy regimens could be developed to strengthen specific muscles and retrain gait such that older adults at risk of falling can perform daily activities safely.

Older adults demonstrated neuromuscular adaptations to maintain stability. Contrary to our hypothesis, older adults exhibited 13% less loading impulse on uneven surfaces, and 4% less vGRF across all surfaces. Additionally, older adults demonstrated 5° greater hip flexion and 5-8° greater knee flexion than young adults. Reduced knee flexion and ROM have been shown to result in reduced ability to absorb the impulse during loading of the stance limb (Saywell et al., 2012). Therefore, increased flexion results in a greater ability to absorb impulse, seen by decreased loading impulse on the uneven surface. To assess potential differences in loading impulse arising from a difference in net average force, the magnitude (%BW) of the first loading peak (at the end of LR) for the challenging surfaces were compared (% difference) to this peak on the normal surface. The differences in loading impulse may have resulted from differences in vertical force (2.7% difference, normal vs uneven; see Appendix). While the reason for increased lower-limb flexion for older adults it is not immediately clear, it may stem from weaker musculature. Muscle strength for the study cohorts was concurrently assessed via isometric hip, knee, and ankle dynamometer testing where preliminary results indicated that older adults had weaker knee and ankle muscles than young adults (see Appendix). Weakness of the muscles surrounding the knee may necessitate older adults to increase hip and knee flexion, as well as knee muscle co-contraction, to prevent knee buckling and maintain joint stability during walking (Peterson and Martin, 2010). Specifically, the 13% greater co-contraction exhibited by older adults may be a compensatory strategy to stiffen the knee and increase lower-limb stability to prevent falling. Additionally, the 30% greater ankle co-contraction employed by older versus
young adults during MS, on the slick surface, may also be a compensatory strategy to maintain stability with weaker muscles. While walking speed is expected to decrease with age, physical activity has been shown to improve gait speeds (McMullan et al., 2020). The older adults were from a more active population, assessed by questionnaire, consequently narrowing the gap between walking speeds, and potentially explaining some of the conflicting findings.

The presence of environmental challenges (uneven and slick surfaces) presented unique demands on the neuromusculoskeletal system to maintain stability. No significant differences between surfaces were observed for vertical loading (initial force and loading impulse). However, the presence of an uneven surface increased lower-limb sagittal-plane kinematics with 4° greater ankle flexion at the end of LR and increased knee flexion by 6-9° and hip flexion by 4-6° through LR and ROM across stance. Additionally, older adults, specifically, demonstrated 5° greater hip flexion and 6° greater knee flexion at MS on an uneven, compared to normal, surface. Increased flexion at these joints lowers the center of mass, potentially improving stability on a challenging surface (Dixon et al., 2018). Additionally, a flatter foot, resulting from decreased ankle dorsiflexion, may reduce the risk of slipping (Gates et al., 2012; Menant et al., 2009). Older adults contacted the uneven floor with 5° less flexion, in agreement with previous work (Dixon et al., 2018). Further, adaptation to a challenging surface differed between the uneven and slick surfaces; only the uneven, not slick, surface increased hip and knee flexion at IC. The slick surface did not result in similar kinematic results as the uneven surface, perhaps in part due to the design of the laboratory walkway with smaller differences in surface slickness (coefficient of friction) between the walkway and the modular slick surface. Contrary to our hypothesis, no significant differences between surfaces were observed for muscle co-contraction. The uneven surface did tend to increase muscle co-activation at the ankle during LR (15% increase) in response to the process of weight acceptance and at the knee during MS (14% increase) to maintain stability, but these observations lacked statistical significance in the current work.

The design of modular surfaces may limit this study. Lacking an analogue encountered in daily life, the adaptations to the uneven surface may not be completely transferable to surfaces with lower variations in height such as cobblestone, natural surface trails, or previously used experimental walkways (DaSilva et al., 2021; Dixon et al., 2018). However, Dixon et al. (2018) suggest that larger changes in surfaces result in larger changes in gait; therefore, these results are likely representative of the direction of adaptations for less abrupt irregular surfaces. Another
potential limitation stems from cohort selection lacking healthy non-faller older adults, potentially obscuring which adaptation strategies arise from age differences versus prior falls. These cohorts were selected to identify the biggest spatiotemporal and kinematic differences between young adults and older adults vulnerable to falling (Sadhegi et al., 2021). Additionally, no significant differences have been found for gluteus medius activity between ages during gait (Lim et al., 2022). Therefore, the omission of this muscle group still provides useful information regarding neuromuscular stabilization during loading response. Further, the use of medications (e.g., benzodiazepines) may impact balance. Our sample size was too small to assess the potential variability in medication history, instead we screened for balance with a single-legged balance test and competency with study tasks. While exclusive attention to sagittal-plane kinematics may be a potential limitation, motions in this plane are the most significant during gait. Other planes may be of more interest for out-of-plane activities such as pivoting or cutting.

Supplementary research regarding how these trends observed during normal gait translate to other, higher-demand activities of daily life (stair navigation and turning) is important as a next step. Furthermore, research is warranted to determine whether these observations during laboratory-controlled gait translate to locomotion over real world surfaces.

In conclusion, lower-limb neuromechanical adaptations exhibited by older adults may be adopted to maintain stability, particularly when walking over uneven surfaces. The older adults exhibited lower-limb neuromechanical adaptations, including greater hip and knee flexion, and co-contraction of the knee musculature than the younger adults, reported to increase limb stability. When walking over a challenging surface, particularly uneven, all participants increased lower-limb flexion and co-contraction of the knee and ankle musculature to maintain stability. Yet, the older adults, who tend to have compromised balance, exhibited greater neuromechanical changes than the younger adults on the uneven surface. Specifically, to avoid an accidental fall, the older adults adopted greater hip and knee flexion, from initial contact through midstance, and reduced the impulse of the vGRF during loading response.

**Declaration of Competing Interest**

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.
Acknowledgements

We greatly thank all the participants who were involved in our study for volunteering their time. This study was supported by grants from the NIH National Institute on Aging (R15AG059655) and NIH Institutional Development Awards (IDeA) from the National Institute of General Medical Sciences (P20GM109095).
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List of Tables

**Table 1.** Mean (standard deviation) as well as 95% confidence intervals (CI) for participant demographics

**Table 2.** Mean (standard deviation) for vertical ground reaction force parameters from analysis of variance: Loading Response (N/BW, %BW), Loading Rate (%BW \cdot %stance−1), Loading Impulse (%BW \cdot %stance, N.stance.BW), and Midstance (%BW).

**Table 3.** Mean (standard deviation) for sagittal-plane joint angle kinematic parameters (°) from analysis of variance.

**Table 4.** Mean (standard deviation) for knee (quad-ham) and ankle (TA-GL) muscle co-contraction (%) from analysis of variance.
List of Figures

Figure 1. (Left) A representative gait trial on the uneven surface. Safety harness can be seen as the yellow straps across the torso as well as the tether behind the participant’s head, connecting the participant to the gantry above (not shown). (Right) The uneven surface (45mm x 45mm) was composed of nine smaller squares of differing heights within the modular surface.

Figure 2. Ankle (left), knee (center), and hip (right) flexion/extension range of motion (ROM) during loading response and across all of stance phase. Loading response is from initial contact to the first vertical ground reaction peak at the end of loading response. Significant surface main effects were found for ankle loading response and knee and hip stance ROM. Knee ROM had a significant age-related difference. Negative values for loading response ROM indicates an increase in extension during this period of stance. Significant differences indicated with * (p < 0.05), ** (p < 0.01), *** (p < 0.001).

Figure 3. Knee flexion/extension across early to midstance for young and older adults (left) and on normal, uneven, and slick surfaces (right). Similar trends were observed for hip kinematics. For age by surface interaction at the end of midstance, box color indicates surface. Significant differences indicated with * (p < 0.05), ** (p < 0.01), *** (p ≤ 0.001).

Figure 4. Percent co-contraction for knee (quadriiceps-hamstring, left) during loading response and ankle (tibialis anterior-gastrocnemius lateralis, right) during midstance. Quadriiceps muscles include the sum of muscle activity recorded by vastus lateralis, rectus femoris, and vastus medialis EMG sensors. Hamstring muscles represent the sum of biceps femoris and semimembranosus EMG sensors. Percent co-contraction represents the portion where both muscle groups were simultaneously activated. For age by surface interaction for ankle co-contraction, box color indicates surface. Significant differences indicated with ** (p < 0.01), *** (p < 0.001).
Table 1
Mean (standard deviation) as well as [95% confidence intervals, CI] for participant demographics.

<table>
<thead>
<tr>
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<th>Young (n = 14)</th>
<th>Older (n = 14)</th>
<th>p-value</th>
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<tr>
<td>Age (year)</td>
<td>21.8 (2.1)</td>
<td>70.1 (3.0)</td>
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<td>[20.7, 22.9]</td>
<td>[68.6, 71.7]</td>
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<td>Sex (No. Male)</td>
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<tr>
<td>Height (cm)</td>
<td>174.6 (9.7)</td>
<td>171.5 (12.0)</td>
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<td>[169.5, 179.7]</td>
<td>[165.2, 177.8]</td>
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<tr>
<td>Weight (kg)</td>
<td>73.0 (15.8)</td>
<td>74.2 (17.9)</td>
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<td>[64.8, 81.3]</td>
<td>[64.9, 83.6]</td>
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<tr>
<td>Speed (m/s⁻¹)</td>
<td>1.06 (0.07)</td>
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<td>[1.03, 1.10]</td>
<td>[0.96, 1.13]</td>
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*p < 0.05. Significant differences between young and older adults.

Table 2
Mean (standard deviation) for vertical ground reaction force parameters from analysis of variance: Loading Response (N/BW, %BW), Loading Rate (%BW · %stance⁻¹), Loading Impulse (%BW · %stance · N.stance.BW), and Midstance (%BW).

<table>
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<td>Older</td>
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<td>4.0 (0.7)</td>
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<td>Loading Impulse</td>
<td>16.7 (1.5)</td>
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<td>Midstance</td>
<td>83.4 (2.8)</td>
<td>82.4 (4.8)</td>
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Statistically significant age and surface main effects p-values bolded; age by surface interaction effect parameter name and p-value bolded; * (α = 0.05) or ** (α = 0.01).
Table 3
Mean (standard deviation) for sagittal plane joint angle kinematic parameters (°) from analysis of variance.

<table>
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<td>5.6 (5.5)</td>
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<td>L/R ROM</td>
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<td>-5.1 (2.6)</td>
<td>-5.4 (3.4)</td>
<td>0.607</td>
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<td>6.4 (3.3)</td>
<td>8.7 (6.0)</td>
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<td></td>
<td>19.5 (4.8)</td>
<td>21.7 (3.6)</td>
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<tr>
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<td>8.2 (6.2)</td>
<td>12.8 (7.7)</td>
<td>0.030*</td>
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<tr>
<td>Loading</td>
<td>14.4 (5.6)</td>
<td>22.2 (7.2)</td>
<td>0.001*</td>
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<tr>
<td>Response</td>
<td>L/R ROM</td>
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<td></td>
</tr>
<tr>
<td></td>
<td>7.9 (3.7)</td>
<td>11.6 (3.8)</td>
<td>0.005**</td>
</tr>
<tr>
<td></td>
<td>8.5 (5.1)</td>
<td>15.0 (7.1)</td>
<td>0.004**</td>
</tr>
<tr>
<td></td>
<td>36.7 (7.1)</td>
<td>33.1 (8.0)</td>
<td>0.132</td>
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<td>Hip</td>
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<td>29.4 (5.4)</td>
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<td>L/R ROM</td>
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<td>-9.6 (2.7)</td>
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<td>6.1 (5.0)</td>
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<td>37.8 (4.6)</td>
<td>39.3 (4.8)</td>
<td>0.233</td>
</tr>
</tbody>
</table>

Statistically significant age and surface main effects p-values barded; age by surface interaction effect parameter name and p-value barded; * (α = 0.05), ** (α = 0.01), *** (α = 0.001).

Table 4
Mean (standard deviation) for knee (quad-ham) and ankle (TA-GL) muscle co-contraction (%) from analysis of variance.

<table>
<thead>
<tr>
<th>Quad-Ham</th>
<th>Age</th>
<th>Surface</th>
<th>Interaction</th>
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<tbody>
<tr>
<td></td>
<td>Young Adult</td>
<td>Older Adult</td>
<td>p-value</td>
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<tr>
<td>Loading</td>
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<td>73.3 (10.6)</td>
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<tr>
<td>Response</td>
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<td>66.3 (15.4)</td>
<td>0.065</td>
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<td>68.5 (10.8)</td>
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<td>Full Stance</td>
<td>37.8 (4.6)</td>
<td>39.3 (4.8)</td>
<td>0.233</td>
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<tr>
<td>TA-GL</td>
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</tr>
<tr>
<td>Loading</td>
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<td>54.0 (16.3)</td>
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<tr>
<td>Response</td>
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<td>64.1 (11.4)</td>
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<tr>
<td>Midstance</td>
<td>54.3 (11.1)</td>
<td>56.6 (8.3)</td>
<td>0.484</td>
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</tbody>
</table>

Statistically significant age and surface main effects p-values barded; age by surface interaction effect parameter name and p-value barded; * (α = 0.05) or ** (α = 0.01).
Figure 1: (Left) A representative gait trial on the uneven surface. Safety harness can be seen as the yellow straps across the torso as well as the tether behind the participant’s head, connecting the participant to the gantry above (not shown). (Right) The uneven surface (45mm x 45mm) was composed of nine smaller squares of differing heights within the modular surface.

Figure 2: Ankle (left), knee (center), and hip (right) flexion/extension range of motion (ROM) during loading response and across all of stance phase. Loading response is from initial contact to the first vertical ground reaction peak at the end of loading response. Significant surface main effects were found for ankle loading response and knee and hip stance ROM. Knee ROM had a significant age-related difference. Negative values for loading response ROM indicates an increase in extension during this period of stance. Significant differences indicated with * (p < 0.05), ** (p < 0.01), *** (p < 0.001).
Figure 3: Knee flexion/extension across early to midstance for young and older adults (left) and on normal, uneven, and slick surfaces (right). Similar trends were observed for hip kinematics. For age by surface interaction at the end of midstance, box color indicates surface. Significant differences indicated with * \( (p < 0.05) \), ** \( (p < 0.01) \), *** \( (p \leq 0.001) \).

Figure 4: Percent co-contraction for knee (quadricep-hamstring, left) during loading response and ankle (tibialis anterior-gastrocnemius lateralis, right) during midstance. Quadricep muscles include the sum of muscle activity recorded by vastus lateralis, rectus femoris, and vastus medialis EMG sensors. Hamstring muscles represent the sum of biceps femoris and semimembranosus EMG sensors. Percent co-contraction represents the portion where both muscle groups were simultaneously activated. For age by surface interaction for ankle co-contraction, box color indicates surface. Significant differences indicated with ** \( (p < 0.01) \), *** \( (p < 0.001) \).