Lower-limb coordination and variability during gait: The effects of age and walking surface.

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Abstract

Background: Falls among community-dwelling older adults are often triggered by uneven walkways. Joint coordination and its variability change with age and may place older adults at risk of falling. It is unclear how irregular surfaces impact lower-limb joint coordination and if such changes are exacerbated by aging.
Research question: To what extent do lower-limb inter-joint coordination and its variability, over flat and uneven brick walkways, differ between older and young healthy adults?
Methods: A motion-capture system collected kinematic data from walking trials on flat and uneven walkways in seventeen older (72.0 ± 4.2 years) and eighteen younger (27.0 ± 4.7 years) healthy adults. Continuous relative phase analyses were performed for the Knee-Hip and Ankle-Knee joint pairs. Mean Absolute Relative Phase (MARP) quantified coordination amplitude. Deviation Phase (DP) quantified coordinative variability. Two-way mixed ANOVA's tested for effects of age, surface, and age × surface interactions.
Results: Uneven surfaces prompted more in-phase MARP inter-joint coordination in adults during most gait phases ($p \leq 0.024$). Age × surface interactions were observed during initial contact (Ankle-Knee: $p = 0.021$, Knee-Hip: $p = 0.001$) and loading response (Knee-Hip: $p = 0.017$), with post-hoc analyses showing coordination accentuated in older adults. Uneven surfaces induced higher DP in Knee-Hip ($p = 0.017$) and Ankle-Knee joint coupling ($p < 0.001$) during gait, largely independent of age. An age × surface interaction was observed during mid-swing ($p = 0.050$), with post-hoc analysis revealing increased variability in older adults.
Significance: More in-phase and variable lower-limb gait behavior was observed on uneven walkways. These differences were accentuated in older adults during early stance phase (more coordinated) and mid-swing (more variable). This may reflect a cautious gait strategy on challenging walkways to maintain stability and help prevent falls.

1. Introduction

Falls among older adults are a major public health concern across many countries. In the United States, between 30 and 60% of persons aged 65 and older experience a fall each year [1]. These events are linked to negative health outcomes [2] and are leading causes of traumatic brain injury and hip fractures in this population [3, 4]. Among community-dwelling older adults, falls typically occur while walking and are often triggered by environmental factors [1]. Of these, uneven walkways are especially problematic in the outdoor built environment [5].

Biomechanical gait analysis can identify impairments during walking and help elucidate factors that contribute to falls. Specifically, patterns of inter-joint coordination (i.e. relationships between joints during motion) and its variability provide insight into neuromuscular control during functional movement [6, 7] . For instance, a lack of coordinative variability during gait reflects an overly rigid motor system, while excessive variability is suggestive of unstable neuromuscular control [8]. Both behaviors point to a less adaptable motor system and may place an individual at risk of falls due to a lack of robust "movement options" when navigating a complex environment [9]. Similar to other gait metrics, patterns of lower limb inter-joint coordination and coordinative variability in the sagittal plane change with healthy aging [10, 11], and these changes may have implications for falls in the elderly. For instance, Chiu et al. reported greater Ankle-Knee coordinative variability during stance phase in older, fall-prone adults, and correlated their findings to clinical measures indicating poor balance [9]. Hafer and Boyer observed more in-phase foot-shank coordination during mid-stance in older, compared to young, adults; findings which may reflect an age-specific attempt to maintain stability during single-leg loading [10]. These studies, however, were conducted on flat surfaces and extrapolating these findings to adults walking on uneven walkways is difficult.

The impact of walking surfaces on joint coordination are not well researched. This has been studied from the standpoint of comparing treadmill versus ground walking [12], or examining the impact of obstacle crossing on gait [13]; however, these situations may not reflect the real-life walking environment of community-dwelling adults. Marigold et al. studied gait variability on multi-surface terrain from the perspective of step parameters, reporting greater step-length and step-width variability, independent of age [14]. In contrast, others have shown age-specific adaptations on irregular surfaces, reporting greater step-width and step timing variability in older, compared to younger, adults [15, 16]. While measures of step parameter variability are useful metrics, they may not reflect patterns of joint/segment coordinative variability in the lower extremity. The study of coordination and its variability will extend this literature base by providing insight into motor control during gait.

Irregular walking surfaces represent a challenge to the neuromuscular system and can increase sensitivity in identifying age-related changes during gait [17, 18]. It is unclear if irregular surfaces similarly impact lower-limb coordination and variability, and if such changes are greater in older adults. Better understanding of changes in lower-limb motor control strategies during gait could have implication for falls and future fall prevention strategies. Therefore, the aims of this study were to (1) compare patterns of lower-limb inter-joint coordination and coordinative variability over flat and uneven brick walkways, and (2) assess differences between older and young healthy adults for these two surfaces. We hypothesized that (1) inter-joint coordination and variability will differ between walking surfaces, and (2) these differences will be exacerbated in older adults.

2. Methods

2.1. *Participants*

Forty-six older (65+ years) and younger (18-35 years) community-dwelling adults were recruited from the Hopkinton, MA area and provided informed consent prior to participating in this study. Twelve participants were excluded based on screening for neurological impairments, musculoskeletal abnormalities (e.g. joint replacement), diabetes, elevated body mass indexes (>30), and poor balance ability. Full screening procedures are described elsewhere [18]. Data from seventeen older (12 females, 71.5 ± 4.2 years, 67.6 ± 12.6 kg, 165.7 ± 9.3 cm) and eighteen younger adults (8 females, 27.0 ± 4.7 years, 69.5 ± 14.7 kg, 171.6 ± 8.8 cm) were analyzed. The Harvard Institutional Review Board approved this study.

2.2. *Data Collection*

Data were collected in a single testing session. An eighteen-camera motion-capture system (Motion Analysis Corp., Santa Rosa, USA) collected retro-reflective marker data at 100 Hz based on a modified Plug-in Gait (PiG) model (Vicon Motion Systems Ltd., Oxford, UK), while healthy adults performed walking trials. Participants performed four straight-walking trials at self-selected walking speeds on both flat and irregular brick surfaces. The irregular surface consisted of bricks randomly tilted in the anterior-posterior or medial-lateral direction via dowels of different diameter placed beneath the bricks (c.f. [18] for more details). Participants wore standardized athletic shoes (Nike Inc., Beaverton, USA) and were tethered to a safety harness.

2.3. Data processing and analysis

Marker data were labeled and gap filled in Cortex (Motion Analysis Corp., Santa Rosa, USA). All data were imported into Matlab (v2020a, The Mathworks Inc., Natick, USA) for further processing with biomechZoo [19] and custom code. Procedures are described in detail elsewhere [18]. In brief, marker data were filtered using a 4^{th} order low-pass Butterworth filter (8Hz cut-off frequency). Joint centers of the hip [20], knee, and ankle [21] were determined, and orientation of thigh wand markers were optimized [22] in anticipation of lower-limb joint angle computations according to the PiG modeller [23]. Discrete events for foot-strike and foot-off were extracted using marker data according to a coordinate-based algorithm [24], and one full gait stride per trial (total trials = 4), across all conditions, for each subject, was extracted for further analyses (Fig. 1). All retained joint angles were in the sagittal plane.

Continuous Relative Phase (CRP) analysis, which consists of determining a signal based on the difference in phase angles of two adjacent joints/segments during functional movement, was used to describe inter-joint coordination and variability [6, 25] (Fig. 1). First, gait data were padded with extraneous kinematic data to the nearest gait event, on either end, to control for potential data distortion associated with CRP analyses [26]. Phase angles for the hip, knee, and ankle joints were determined using the Hilbert transform approach as described by Lamb and Stöckl [27]. This involves amplitude-centering the kinematic signal around zero, and calculating phase angles using joint angles at time t, $x(t)$, and their Hilbert transformation $H(t)=H(x(t))$. The Hilbert transform results in a complex signal, $\zeta(t)$, where the $H(t)$ of $x(t)$ serves as the imaginary portion of the analytical signal (Eq 1).

Eq 1. $$\zeta(t) = x(t) + iH(t)$$

Next, at any point in time t_j, the phase angle of a joint can be determined as the inverse tangent of the transformed signal $H(t_i)$ divided by the measured signal $x(t)$ (Eq. 2).

Eq 2. $\phi(t_i) = \arctan((H(t_i)/x(t_i)))$

These procedures were repeated for each walking trial, across both walking surfaces, for all participants. Data pads were removed, and the absolute difference in phase angles for two adjacent joints quantified coordination for the Knee-Hip and Ankle-Knee joint pairs, across the individual gait cycles. To adjust for discontinuities, CRP values greater than 180 were subtracted from 360 [27]. The resulting CRP values are between 0 and 180, where 0 represents fully in-phase coupling (i.e. more tightly coordinated), while 180 represent fully out-of-phase coupling (i.e. less tightly coordinated) [7]. Knee-Hip and Ankle-Knee CRP curves were time normalized to 100% of the gait cycle, and partitioned into stance (initial contact, loading response, mid-stance, terminal stance, and pre-swing) and swing (initial swing, mid-swing, terminal swing) phases using normative gait cycle event timings [28]. To summarize patterns of coordination amplitude and coordinative variability, the Mean Absolute Relative Phase (MARP) and Deviation Phase (DP) were used, respectively [29]. MARP was determined by taking the mean of the ensemble averaged CRP curves of the four trials, while DP was quantified by averaging the standard deviations of ensemble CRP curves over the gait cycle (Fig. 1). MARP and DP were calculated for each phase of gait, across both walking surfaces, for each joint pair, in all participants.

2.4. Statistical analysis

To test our hypotheses examining the effects of age and surface, 2 × 2 mixed design Analysis of Variance (ANOVAs) were run using age (old/young) as a between-subject factor and surface (flat/uneven) as a within-subject factor. An age × surface interaction was included to test our second hypothesis that the effects of surface differed in older adults. Partial Eta-Squared (η^2) was used to quantify effect sizes. Separate analyses were performed for our dependent variables, MARP (coordination) and DP (coordinative variability), for each event (initial contact, loading response, mid-stance, terminal stance, pre-swing, initial swing, mid-swing, terminal swing), for the Knee-Hip and Ankle-Knee joint pairs. Deviations from normality were not tested during the ANOVAs as the design is robust against departures from normality [30]. In the event of a significant interaction, paired post-hoc tests with Bonferroni corrections were used to compare differences in group by condition (young/flat vs. young/uneven and old/flat vs. old/uneven). These data were checked for normality using QQ plots, and Levene's test examined homogeneity of variance. If parametric assumptions were not met, Wilcoxon Signed-Rank tests were run instead of paired t-tests (see results for test implemented). Cohen's d and Glass's delta (**Δ**) quantified effect sizes for the results of the paired t-tests and Wilcoxon Signed-Rank tests, respectively [31]. Level of significance for all tests was set at $\alpha \leq 0.05$. All statistical analyses were performed using SPSS (Version 23, IBM, Chicago, USA).

3. Results

A post-hoc mixed ANOVA showed no main effect of age ($p = 0.090$) nor walking surface ($p = 0.293$) on mean self-selected gait speed.

3.1. Inter-joint Coordination (MARP)

For the Ankle-Knee joint pair, significant surface effects were observed during initial contact, loading response, mid-swing and terminal swing, while age effects were observed during initial contact, initial swing and terminal swing (Table 1). Significant age × surface interactions occurred during initial contact ($p = 0.021$). Follow-up Wilcoxon Signed-Rank tests showed that older adults adopted more in-phase Ankle-Knee coordination (i.e. lower MARP) to a greater extent on uneven, compared to flat, surfaces (median difference -67.1°; $p = 0.002$, $\boldsymbol{\Delta} = 0.848$) than young adults (median difference -23.8°, $p = 0.020$, $\boldsymbol{\Delta} = 0.247$) (Fig. 2). The uneven walkway prompted lower MARP during loading response ($p = 0.024$), mid-swing ($p < 0.001$) and terminal swing ($p < 0.001$) phases, independent of age. Age effects occurred during initial ($p = 0.012$) and terminal swing ($p = 0.020$) phases, suggesting less and more in-phase behavior respectively, in older adults, regardless of walking surface.

For Knee-Hip coupling, the uneven surface resulted in lower MARP values, indicating more in-phase coupling across all gait phases ($p = 0.023$) (Table 1). Significant age × surface interactions were observed during initial contact ($p = 0.001$) and loading response ($p = 0.017$). Follow up paired t-tests revealed that an uneven surface prompted more in-phase Knee-Hip coupling to a greater extent in older, versus younger adults during initial contact (Old: mean change -12.8°, $p < 0.001$, $d = 2.031$; Young: mean change -5.4°, $p = 0.001$, $d = 0.947$) and loading response (Old: mean change -10.5°, $p = 0.001$, $d = 1.125$; Young: mean change -3.9°, $p = 0.014$, $d = 0.647$) (Fig. 2). Age effects were observed during initial ($p = 0.040$), mid ($p = 0.008$), and terminal swing ($p = 0.048$), suggesting that older adults were more in-phase during swing phase, independent of surface.

3.2. Inter-joint Coordinative Variability (DP)

Regarding Ankle-Knee coupling, the uneven walkway induced more variable patterns of coordination (higher DP values), independent of age, during most gait phases ($p < 0.001$) (Table 2). The exception was an age × surface interaction during mid-swing ($p = 0.050$). Follow-up Wilcoxon Signed-Rank tests revealed more variable Ankle-Knee coupling in older adults on uneven, compared to flat, surfaces (median difference 7.6°, $p = 0.001$, $\boldsymbol{\Delta} = 3.97$), versus young adults (median difference 3.8°, $p = 0.025$, $\boldsymbol{\Delta}= 1.07$) (Fig. 2).

For Knee-Hip coupling, the uneven walkway prompted greater DP values during all gait phases, largely independent of age ($p = 0.017$) (Table 2). An age effect occurred during initial swing ($p = 0.039$), suggesting reduced Knee-Hip variability in older adults, independent of walking surface.

4. Discussion

4.1. Summary

This study investigated differences in lower-limb joint coordination amplitude (MARP) and variability (DP), during gait on flat and uneven walking surfaces, in young and older healthy adults. The uneven walkway prompted more in-phase Ankle-Knee coupling (i.e. more tightly coordinated) in all adults during key aspects of stance and swing phases (Fig. 3), with this response being greater in older adults during initial contact (Fig. 2). For Knee-Hip coupling, a more in-phase walking strategy on an uneven surface was observed in all adults during gait (Fig. 4), with this change being accentuated in older adults during initial contact and loading response (Fig. 2). During swing phase, more in-phase Knee-Hip coupling was observed in older adults, independent of walking surface. Regarding variability, the uneven surface resulted in more variable patterns of coordination during gait, largely independent of age, for both joint pairs. The sole exception was during mid-swing for Ankle-Knee, where a greater change to a variable gait pattern in older adults was observed. More in-phase lower-extremity coupling on the uneven surface during stance in older adults may reflect a more cautious gait pattern, which serves to help reduce the risk of falling. Changes in variability were likely a functional response to an uneven walking surface in all participants, due to a greater challenge to the neuromuscular system.

4.2. Age and surface-related changes in inter-joint coordination (MARP)

Consistent with our hypotheses, the uneven walking surface prompted changes in Knee-Hip and Ankle-Knee coordination, and these differences were greater in older, compared to young adults (Fig. 2). These observations of more in-phase behavior during early stance likely point to a more cautious, en-bloc, gait strategy implemented by older adults. Joint coordination plays an essential role in maintaining a stable gait pattern [32]. Therefore, this strategy may reflect an attempt to uphold a stable walking posture on a challenging surface [32], potentially reducing risk of falls via tripping or loss of balance. During swing phase, more in-phase behavior was also observed in the Knee-Hip (all phases) and Ankle-Knee (mid-terminal swing) joint pairs, independent of age. These behaviors may be indicative of a preparatory motor strategy as an individual plans for the upcoming loading phase on an uncertain surface, and/or considers toe-clearance over elevated bricks. In other words, these 'more cautious' strategies could be entirely functional and reflective of changing gait dynamics with healthy aging.

To our knowledge, this is the first study to examine changes in coordination in healthy adults on an uneven surface, which makes comparisons with other studies difficult.

Comparability is further complicated by differences in methodologies used to quantify coordination in the gait literature, some of which are more susceptible to bias than others. Past works examining lower-limb coordination during gait reported (i) no change in Knee-Hip and Ankle-Knee coordination during obstacle-crossing, when comparing younger and older adults [13]; and (ii) large correlations between slower walking speeds and less coordinated Foot-Shank coupling during late stance phase in older adults [33]. Similar to uneven walkways, both obstacle crossing and slower gait represent a more complex task for the neuromuscular motor system; however, these works present conflicting results, both supporting and refuting the notion that coordination changes with task complexity. Else, others found older, compared to young adults adopted more in-phase coordination during early-mid stance in the Shank-Thigh and Foot-Shank couples; but did not compare these across different walking surfaces [10, 11]. While our findings suggest that an uneven surface prompts more in-phase behavior to a greater extent in older adults, the impact of uneven surfaces on lower-extremity coordination requires further study.

4.3. Age and surface-related changes in inter-joint coordinative variability (DP)

Consistent with our first hypothesis, an uneven surface prompted more variable Ankle-Knee and Knee-Hip coordination (higher DP) across most of the gait cycle (Table 2). Contrary to our second hypothesis, however, greater inter-joint coordinative variability was observed largely independent of age. Given these changes occurred in parallel between younger and older healthy adults, modest increases in lower-limb variability are likely functional and reflective of adaptations to the increased demands of an irregular surface [8]. In line with our second hypothesis, this more variable response was accentuated in older, compared to young adults during mid-swing. The implications of this finding, however, are less clear. Functional motor control is proposed to occur across an 'optimal range' of variability, where excessive variability can reflect an unstable movement pattern [8, 34]. However, when considering our sample (healthy older adults, no history of falls) and past work reporting reduced Ankle-Knee variability during swing phase in fall-prone older adults [9]; categorizing our observations as dysfunctional may not be warranted. Rather, this strategy may simply reflect a more cautious older adult group cognizant of toe clearance on an uneven surface.

While studying kinematic measures of gait variability (e.g. step length) is common, examining intra-limb coordination variability, particularly with respect to walking surfaces, is more novel. Past work linked greater toe-clearance with greater variability of Knee-Hip and Ankle-Knee couples in the leading leg during obstacle crossing in older adults [13]. While we observed a similar behavior on an uneven surface (i.e. more variable lower extremity coupling); with the exception of Ankle-Knee coupling during mid-swing, this occurred mostly independent of age. In-line with our findings, Byrne and colleagues reported increases in Foot-Shank variability for both younger and older adults, in response to more challenging walking condition (asymmetrically loaded leg) [11]. Whereas, Chui et al. linked greater Ankle-Knee variability during stance in fall-prone older adults with poor dynamic balance scores, our data suggest that this response was due to the walking surface [9].

4.4. Limitations

Our findings should be interpreted in light of several limitations. First, our walking environment may not have been sufficiently challenging to generate large changes in neuromuscular control. Our investigation was limited to the sagittal plane; thus, we cannot extrapolate these findings to other planes of movement. Variability was analyzed using four walking trials; there are no measurement standards for variability and recording additional trials may have yielded different results. Some researchers have performed coordination analyses using joint angles [9, 12], while others have preferred reporting segment angles [10, 11]. The two approaches may produce slightly different findings; however, we have chosen the former as joint angle-based analyses are more familiar to the gait analysis community and may aid in wider adoption of coordination metrics.

5. Conclusion

In conclusion, an uneven walking surface prompted more tightly coordinated and more variable patterns of Ankle-Knee and Knee-Hip coupling during walking in healthy adults. These changes were accentuated in older adults during early stance (i.e. more in-phase) and mid-swing phases (i.e. more variable). This response may reflect a cautious gait pattern, or a functional strategy to help mitigate the risk of falls in older adults.

Code and data associated with this investigation are available at:

https://github.com/PhilD001/crp_irregular_surfaces

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Conflict of Interest Disclosure: None.

Table 1.
Analysis of variance (ANOVA) results for Mean Absolute Relative Phase (MARP)

Joint Pair	Gait phase	Surface				Age				Age x Surface	
		Flat	Uneven	p	η^2	Young	Old	p	η^2	p	η^2
Ankle-Knee	Initial Contact	135.3 (39.0)	113.8 (46.1)	**<0.001**	0.347	137.1 (36.7)	111.2 (47.1)	**0.050**	0.111	**0.021**	0.151
	Loading response	87.0 (42.2)	76.5 (42.6)	**0.024**	0.145	93.3 (45.6)	69.6 (35.5)	0.082	0.089	0.332	0.029
	Mid-Stance	93.7 (28.5)	95.5 (29.8)	0.480	0.015	97.0 (34.8)	92.1 (21.4)	0.610	0.008	0.501	0.014
	Terminal Stance	148.6 (21.7)	147.6 (23.0)	0.653	0.006	147.5 (24.3)	148.7 (20.0)	0.871	0.001	0.437	0.018
	Pre-Swing	155.5 (12.8)	155.6 (13.3)	0.965	<0.001	154.6 (14.9)	156.7 (10.6)	0.624	0.007	0.086	0.087
	Initial Swing	164.8 (4.0)	163.9 (3.3)	0.069	0.097	163.0 (3.8)	165.8 (2.9)	**0.012**	0.175	0.642	0.007
	Mid-Swing	138.4 (18.7)	129.7 (19.5)	**<0.001**	0.391	134.3 (21.5)	133.8 (17.3)	0.939	<0.001	0.397	0.022
	Terminal Swing	127.8 (29.8)	102.7 (37.6)	**<0.001**	0.554	127.1 (29.8)	102.6 (38.0)	**0.020**	0.154	0.067	0.098
Knee-Hip	Initial Contact	155.7 (11.8)	146.7 (11.2)	**<0.001**	0.709	152.4 (11.5)	150.0 (13.1)	0.524	0.012	**0.001**	0.290
	Loading response	158.7 (13.2)	151.6 (13.9)	**<0.001**	0.475	158.0 (13.5)	152.1 (13.8)	0.177	0.055	**0.017**	0.162
	Mid-Stance	122.2 (9.3)	117.8 (9.8)	**<0.001**	0.271	122.2 (9.0)	117.7 (10.1)	0.133	0.067	0.099	0.080
	Terminal Stance	68.6 (7.4)	66.7 (5.7)	**0.023**	0.146	68.0 (6.1)	67.3 (7.2)	0.763	0.003	0.534	0.012
	Pre-Swing	51.3 (4.6)	48.7 (4.4)	**<0.001**	0.418	50.5 (3.9)	49.5 (5.3)	0.471	0.016	0.827	0.001
	Initial Swing	44.2 (3.7)	39.6 (3.6)	**<0.001**	0.822	43.1 (4.3)	40.6 (4.1)	**0.040**	0.122	0.985	<0.001
	Mid-Swing	69.3 (6.1)	63.2 (5.7)	**<0.001**	0.731	68.6 (6.6)	63.8 (5.7)	**0.008**	0.194	0.851	0.001
	Terminal Swing	121.7 (7.9)	115.0 (7.6)	**<0.001**	0.693	120.7 (8.2)	115.8 (8.0)	**0.048**	0.114	0.213	0.047

Mean (standard deviation) for all dependent Mean Absolute Relative Phase (MARP) variables for the Knee-Ankle and Knee-Hip joint pairs during each phase of gait. Statistically significant surface, age, and age x surface (A x S) interactions effect p-values (p) are shown, bolded indicates statistical significance of $p < 0.05$)

Table 2.
Analysis of variance results (ANOVA) for Deviation Phase (DP)

Joint Pair		Surface				Age				Age x Surface	
	Gait phase	Flat	Uneven	p	η^2	Young	Old	p	η^2	p	η^2
Ankle-Knee	Initial Contact	15.2 (16.8)	26.6 (18.8)	**<0.001**	0.280	17.2 (15.9)	24.8 (20.3)	0.143	0.064	0.346	0.027
	Loading response	14.6 (10.8)	26.8 (13.5)	**<0.001**	0.380	19.1 (12.1)	23.2 (15.0)	0.128	0.069	0.853	0.001
	Mid-Stance	10.4 (3.8)	20.8 (9.4)	**<0.001**	0.553	16.0 (8.9)	14.9 (8.7)	0.459	0.017	0.503	0.014
	Terminal Stance	6.3 (3.1)	13.8 (8.3)	**<0.001**	0.496	10.5 (8.2)	9.3 (6.0)	0.373	0.024	0.658	0.006
	Pre-Swing	4.1 (2.0)	9.0 (4.3)	**<0.001**	0.595	6.8 (4.2)	6.1 (4.0)	0.364	0.025	0.821	0.002
	Initial Swing	3.4 (1.4)	5.5 (1.9)	**<0.001**	0.484	4.7 (1.8)	4.3 (2.1)	0.358	0.026	0.153	0.061
	Mid-Swing	7.2 (3.3)	14.5 (9.2)	**<0.001**	0.388	9.5 (4.8)	12.3 (9.8)	0.093	0.083	**0.050**	0.111
	Terminal Swing	13.4 (10.6)	23.8 (11.4)	**<0.001**	0.388	16.6 (10.8)	20.5 (13.1)	0.208	0.048	0.876	0.001
Knee-Hip	Initial Contact	3.9 (2.1)	5.9 (3.5)	**0.002**	0.260	5.1 (3.4)	4.9 (3.)	0.997	<0.001	0.694	0.005
	Loading response	4.5 (2.8)	7.1 (4.0)	**<0.001**	0.320	5.9 (4.2)	6.0 (3.6)	0.778	0.002	0.938	<0.001
	Mid-Stance	5.8 (2.4)	8.1 (3.2)	**0.001**	0.273	7.1 (3.2)	6.8 (2.9)	0.831	0.001	0.796	0.002
	Terminal Stance	3.4 (1.3)	4.5 (1.6)	**0.004**	0.225	4.2 (1.6)	3.7 (1.4)	0.114	0.074	0.919	<0.001
	Pre-Swing	2.4 (1.2)	3.2 (1.5)	**0.017**	0.162	2.8 (1.1)	2.7 (1.6)	0.729	0.004	0.416	0.020
	Initial Swing	2.2 (0.9)	2.7 (1.0)	**0.008**	0.195	2.8 (1.0)	2.2 (0.9)	**0.039**	0.123	0.187	0.052
	Mid-Swing	2.6 (1.0)	3.8 (1.5)	**<0.001**	0.425	3.5 (1.3)	3.1 (1.6)	0.446	0.018	0.592	0.009
	Terminal Swing	3.3 (1.9)	4.6 (2.4)	**0.008**	0.194	3.9 (2.3)	4.2 (2.5)	0.497	0.014	0.333	0.028

Mean (standard deviation) for all dependent Deviation Phase (DP) variables for the Knee-Ankle and Knee-Hip joint pairs during each phase of gait. Statistically significant age, surface, and age × surface (A × S) interactions effect p-values (p) are shown, bolded indicates statistical significance of p<0.05).

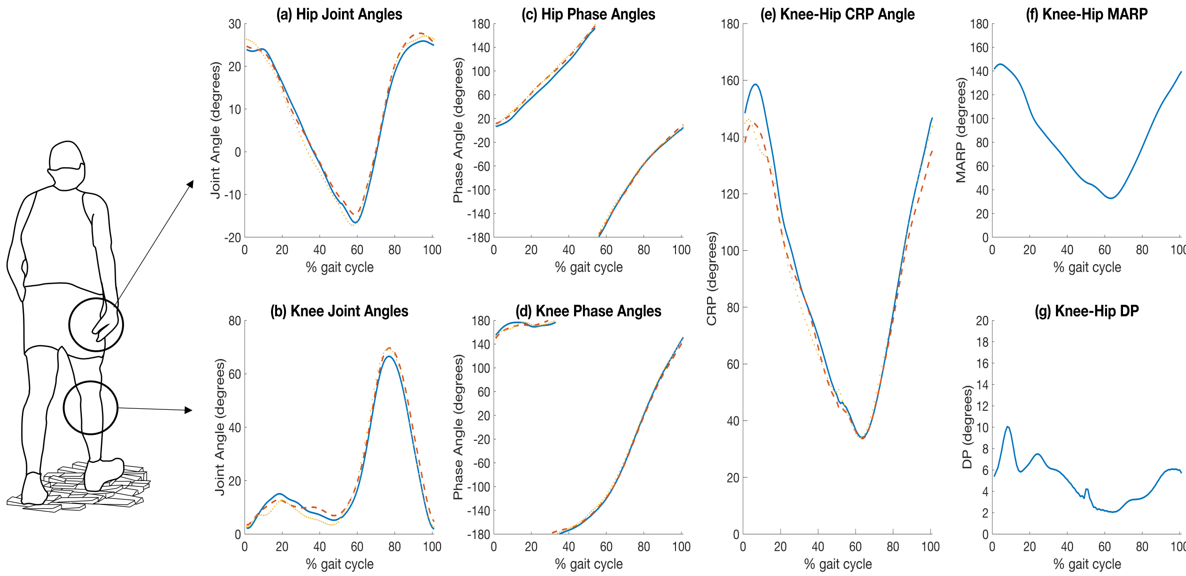

Fig 1. Measurement of hip and knee joint kinematics during representative walking trials (trial 1 solid line; trial 2 dashed line; trial 3 dotted line) on an irregular surface and steps to determine continuous relative phase (CRP) angles. Graphs depict time normalized hip and knee joint angles (a, b); hip and knee phase angles determined using the Hilbert-transform approach (c, d); knee-hip CRP angles on a 0-180 degree scale (e); knee-hip Mean Absolute Relative Phase (MARP), which quantifies CRP amplitude across multiple trials (f); and knee-hip Deviation Phase (DP), which quantifies CRP variability across multiple trials (g).

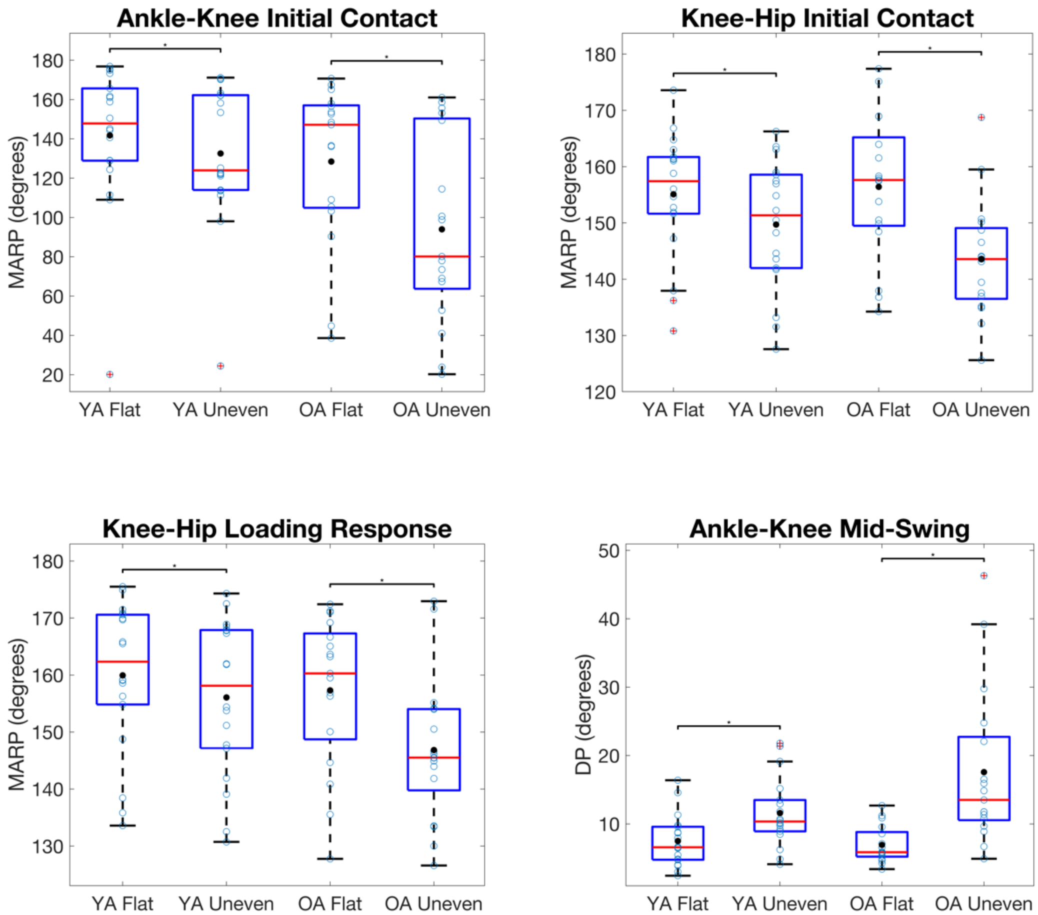

Fig. 2. Summary of Analysis of Variance (ANOVA) age × surface interaction effects for Mean Absolute Relative Phase (MARP) and Deviation Phase (DP) of the Ankle-Knee and Knee-Hip joint pairs comparing Young Adults (YA) and Older Adults (OA) across flat and uneven walkways during gait. Black filled circle denotes mean, while horizontal line shows quartiles and median. Significance bar denotes statistically significant difference ($p < 0.05$).

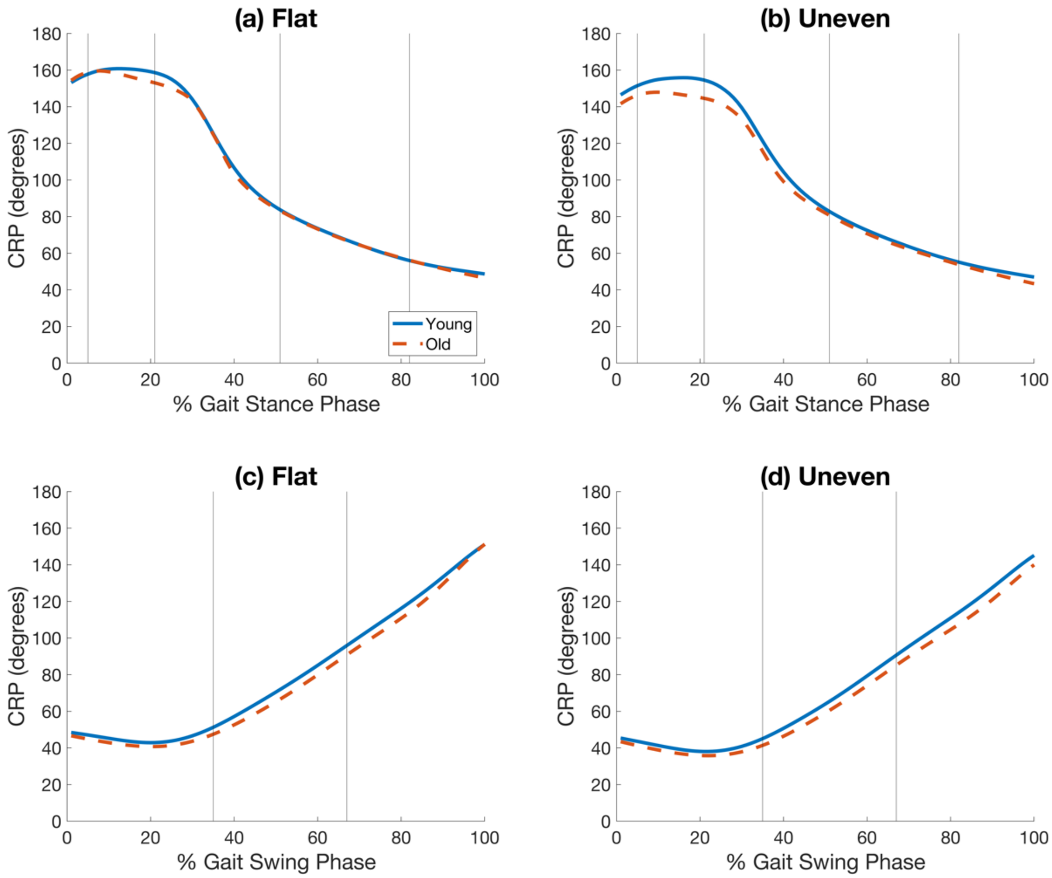

Fig 3. Ensemble averaged Continuous Relative Phase (CRP) curves for the Ankle-Knee joint pair in older and younger adults walking on flat and uneven surfaces, separated into stance and swing phases: (a) flat – stance; (b) uneven – stance; (c) flat – swing; (d) uneven – swing. Vertical lines denote partitioned gait phases for stance (initial contact, loading response, mid-stance, terminal stance, pre-swing) and swing (initial swing, mid-swing, terminal swing), respectively.

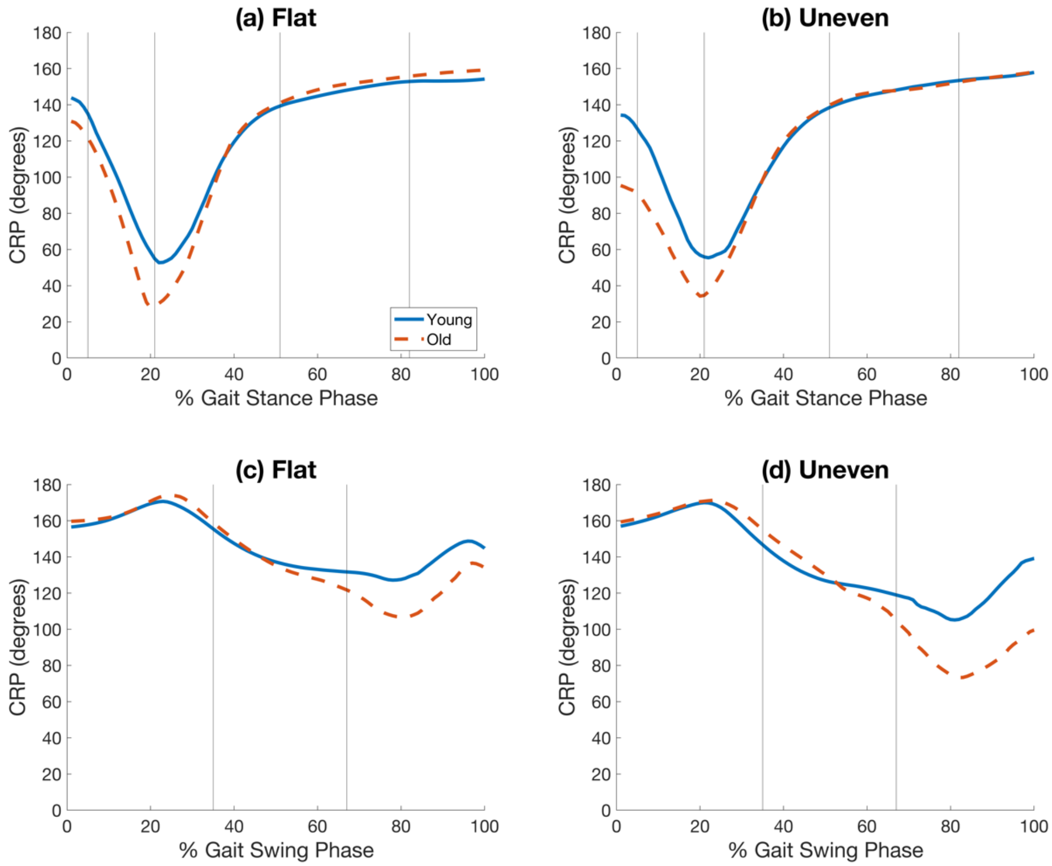

Fig. 4. Ensemble averaged Continuous Relative Phase curves for the Knee-Hip joint pair in older and younger adults walking on flat and uneven surfaces, separated into stance and swing phases: (a) flat – stance; (b) uneven – stance; (c) flat – swing; (d) uneven – swing Vertical lines denote partitioned gait phases for stance (initial contact, loading response, mid-stance, terminal stance, pre-swing) and swing (initial swing, mid-swing, terminal swing), respectively.

References

[1] L.Z. Rubenstein, Falls in older people: epidemiology, risk factors and strategies for prevention, Age Ageing 35 Suppl 2 (2006) ii37-ii41.
[2] M.E. Tinetti, Clinical practice. Preventing falls in elderly persons, N Engl J Med 348(1) (2003) 42-9.
[3] W.C. Hayes, E.R. Myers, J.N. Morris, T.N. Gerhart, H.S. Yett, L.A. Lipsitz, Impact near the hip dominates fracture risk in elderly nursing home residents who fall, Calcif Tissue Int 52(3) (1993) 192-8.
[4] T.E. Jager, H.B. Weiss, J.H. Coben, P.E. Pepe, Traumatic brain injuries evaluated in U.S. emergency departments, 1992-1994, Academic emergency medicine : official journal of the Society for Academic Emergency Medicine 7(2) (2000) 134-40.
[5] W. Li, T.H. Keegan, B. Sternfeld, S. Sidney, C.P. Quesenberry, Jr., J.L. Kelsey, Outdoor falls among middle-aged and older adults: a neglected public health problem, Am J Public Health 96(7) (2006) 1192-200.
[6] R. Burgess-Limerick, B. Abernethy, R.J. Neal, Relative phase quantifies interjoint coordination, J Biomech 26(1) (1993) 91-4. https://www.ncbi.nlm.nih.gov/pubmed/8423174.
[7] J. Hamill, R.E. van Emmerik, B.C. Heiderscheit, L. Li, A dynamical systems approach to lower extremity running injuries, Clin Biomech. 14(5) (1999) 297-308. https://www.ncbi.nlm.nih.gov/pubmed/10521606.
[8] N. Stergiou, R. Harbourne, J. Cavanaugh, Optimal movement variability: a new theoretical perspective for neurologic physical therapy, Journal of neurologic physical therapy : JNPT 30(3) (2006) 120-9.
[9] S.L. Chiu, L.S. Chou, Variability in inter-joint coordination during walking of elderly adults and its association with clinical balance measures, Clin Biomech (Bristol, Avon) 28(4) (2013) 454-8.
[10] J.F. Hafer, K.A. Boyer, Age related differences in segment coordination and its variability during gait, Gait & posture 62 (2018) 92-98.
[11] J.E. Byrne, N. Stergiou, D. Blanke, J.J. Houser, M.J. Kurz, P.A. Hageman, Comparison of gait patterns between young and elderly women: an examination of coordination, Percept Mot Skills 94(1) (2002) 265-80.
[12] S.-L. Chiu, C.-C. Chang, L.-S. Chou, Inter-joint coordination of overground versus treadmill walking in young adults, Gait & posture 41(1) (2015) 316-318.
[13] H.-C. Yen, H.-L. Chen, M.-W. Liu, H.-C. Liu, T.-W. Lu, Age effects on the inter-joint coordination during obstacle-crossing, Journal of biomechanics 42(15) (2009) 2501-2506.
[14] D.S. Marigold, A.E. Patla, Age-related changes in gait for multi-surface terrain, Gait & posture 27(4) (2008) 689-696.
[15] S.B. Thies, J.K. Richardson, J.A. Ashton-Miller, Effects of surface irregularity and lighting on step variability during gait:: A study in healthy young and older women, Gait & posture 22(1) (2005) 26-31.
[16] H.B. Menz, S.R. Lord, R.C. Fitzpatrick, Age-related differences in walking stability, Age and ageing 32(2) (2003) 137-142.
[17] K. Zurales, T.K. DeMott, H. Kim, L. Allet, J.A. Ashton-Miller, J.K. Richardson, Gait Efficiency on an Uneven Surface Is Associated with Falls and Injury in Older Subjects with a Spectrum of

Lower Limb Neuromuscular Function: A Prospective Study, Am J Phys Med Rehabil 95(2) (2016) 83-90.
[18] P.C. Dixon, K.H. Schutte, B. Vanwanseele, J.V. Jacobs, J.T. Dennerlein, J.M. Schiffman, Gait adaptations of older adults on an uneven brick surface can be predicted by age-related physiological changes in strength, Gait Posture 61 (2018) 257-262.
[19] P.C. Dixon, J.J. Loh, Y. Michaud-Paquette, D.J. Pearsall, biomechZoo: An open-source toolbox for the processing, analysis, and visualization of biomechanical movement data, Comput Methods Programs Biomed 140 (2017) 1-10.
[20] M.E. Harrington, A.B. Zavatsky, S.E. Lawson, Z. Yuan, T.N. Theologis, Prediction of the hip joint centre in adults, children, and patients with cerebral palsy based on magnetic resonance imaging, J Biomech 40(3) (2007) 595-602.
[21] Vicon Motion Systems Ltd. Plug-In Gait Manual Version 1.
[22] R. Baker, L. Finney, J. Orr, A new approach to determine the hip rotation profile from clinical gait analysis data, Human Movement Science 18(5) (1999) 655-667.
[23] M.P. Kadaba, H. Ramakrishnan, M. Wootten, Measurement of lower extremity kinematics during level walking, Journal of orthopaedic research 8(3) (1990) 383-392.
[24] J.A. Zeni, Jr., J.G. Richards, J.S. Higginson, Two simple methods for determining gait events during treadmill and overground walking using kinematic data, Gait Posture 27(4) (2008) 710-4.
[25] J.S. Kelso, Dynamic patterns: The self-organization of brain and behavior, MIT press1995.
[26] P. Ippersiel, R. Preuss, S.M. Robbins, The Effects of Data Padding Techniques on Continuous Relative-Phase Analysis Using the Hilbert Transform, J Appl Biomech 35(4) (2019) 247-255.
[27] P.F. Lamb, M. Stockl, On the use of continuous relative phase: Review of current approaches and outline for a new standard, Clin Biomech 29(5) (2014) 484-93.
[28] J. Perry, B. J.M., Gait analysis: Normal and pathological function, 2nd ed., Slack Publishing, Thorofare, USA, 2010.
[29] N. Stergiou, J.L. Jensen, B.T. Bates, S.D. Scholten, G. Tzetzis, A dynamical systems investigation of lower extremity coordination during running over obstacles, Clin Biomech 16(3) (2001) 213-21. https://www.ncbi.nlm.nih.gov/pubmed/11240056.
[30] B. Kirkwood, J. Stern, Essential medical statistics, 2nd ed., Blackwell publishing, Oxford, UK, 2003.
[31] J. Cohen, Statistical Power Analysis for the Behavioral Sciences, 2nd ed., L. Erlbaum1988.
[32] R.E. Van Emmerik, J. Hamill, W.J. McDermott, Variability and coordinative function in human gait, Quest 57(1) (2005) 102-123.
[33] S. Ogaya, A. Iwata, Y. Higuchi, S. Fuchioka, The association between intersegmental coordination in the lower limb and gait speed in elderly females, Gait Posture 48 (2016) 1-5.
[34] J. Hamill, C. Palmer, R.E. Van Emmerik, Coordinative variability and overuse injury, Sports medicine, arthroscopy, rehabilitation, therapy & technology : SMARTT 4(1) (2012) 45.