



OPEN Diurnal variations in gait parameters among older adults with early-stage knee osteoarthritis: insights from wearable sensor technology

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Knee osteoarthritis (KOA) significantly impairs mobility in older adults. Understanding its impact on gait dynamics throughout the day is crucial for optimizing management strategies. This study aimed to explore diurnal variations in gait parameters among older adults with KOA using an in-shoe motion sensor (IMS) system equipped with accelerometers and gyroscopes. In this cross-sectional observational study, 19 older adults clinically diagnosed with early-stage KOA (Kellgren–Lawrence grades 1 or 2) participated. Key gait parameters were measured using an IMS system during morning (6:00 AM–11:59 AM) and afternoon (12:00 PM–5:00 PM) sessions. The IMS, placed bilaterally in the participants' shoes, continuously collected gait data during normal daily activities over a 24-hour period. Participants were instructed to walk for at least 10 min in each session. Data were analyzed using descriptive statistics, and paired t-tests or Wilcoxon signed-rank tests were applied to identify significant differences between sessions. Statistical significance was set at $p < 0.05$. The study included 19 participants (11 females, 8 males) with an average age of 71.4 ± 4.2 years. Walking speed decreased significantly from 1.06 ± 0.14 m/s in the morning to 0.99 ± 0.16 m/s in the afternoon ($p = 0.028$). Similarly, the maximum dorsiflexion angle decreased from $20.34^\circ \pm 2.98^\circ$ to $18.80^\circ \pm 3.01^\circ$ ($p = 0.024$), and the maximum plantar flexion angle decreased from $63.40^\circ \pm 5.84^\circ$ to $60.79^\circ \pm 5.77^\circ$ ($p = 0.017$) in the afternoon. Other parameters such as foot height, peak swing angular velocity, and maximum speed during the swing phase also showed significant reductions in the afternoon. Conversely, the roll angle of heel contact increased from $4.60^\circ \pm 2.62^\circ$ to $5.53^\circ \pm 3.12^\circ$ ($p = 0.026$), and stance time and pushing time increased significantly in the afternoon. Significant diurnal variations in gait parameters among older adults with KOA highlight the importance of considering time of day when evaluating gait and planning interventions. Wearable sensor technology enables continuous, objective gait monitoring in real-world settings, facilitating personalized and time-sensitive approaches for managing KOA.

Keywords Knee osteoarthritis, Older adults, In-shoe motion sensor, Wearable sensor technology, Gait parameters, Diurnal variations

Osteoarthritis (OA) is a prevalent multifactorial chronic condition that affects joints and its tissues¹. OA affects approximately 7% (over 500 million individuals) of the global population^{2,3}. Among the susceptible joints, the knee, hand, and hip joints are particularly prone to developing OA⁴. The disease is particularly disabling when the knees are affected. Knee OA (KOA) accounts for almost four-fifths of the global burden of OA⁵. Global estimates suggest that 9.6% of men and 18.0% of women aged 60 years and older experience symptomatic OA⁶. Various factors, such as advanced age, prior knee injuries, high body mass index (BMI), female sex, genetic predisposition, sedentary lifestyle, repetitive movements, participation in high-impact sports, joint misalignment, and instability significantly increase the risk of KOA development^{7,8}. Clinical manifestations of OA encompass joint functional limitations, stiffness, pain, swelling, joint deformity, and difficulties in walking or running, and probably other symptoms^{1,9,10}.

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Walking ability serves as a crucial indicator of overall health and reflects an individual's functional capacity for daily activities. It has also been proposed to be a 'vital sign'^{11,12}. Recent research has established significant correlations between various walking parameters such as gait speed and sedentary time and important health outcomes including mortality, morbidity, and quality of life^{13,14}. Parameters, such as speed and other spatiotemporal aspects, are pivotal for characterizing gait under a wide range of pathological conditions. In patients with KOA, pain often leads to compensatory mechanisms aimed at minimizing joint loading and resultant discomfort¹⁵. KOA can limit walking tolerance, leading to subsequent health consequences^{16,17}. Compared to individuals without KOA, those with KOA exhibit distinct gait characteristics, including smaller knee extension excursion and moment, larger knee flexion excursion and moment, and larger knee adduction excursion and moment when walking at self-selected gait speeds^{15,18,19}. Furthermore, temporal- and distance-related gait factors, including velocity, cadence, and stride length, are frequently diminished in KOA^{20,21}. Additionally, parameters such as stance phase flexion–extension, peak vertical ground-reaction forces, and loading rates are notably reduced in patients with KOA^{22–24}. Meanwhile, stride and double support times were increased in patients with OA²¹.

A deeper understanding of gait is important to quantify the pathomechanics of patients with KOA. In general, noninvasive gait data on KOA in the literature have been obtained from specialized gait laboratories equipped with camera-based motion capture systems and force plates integrated into the floor²⁵. However, this conventional gait analysis method requires specialized locomotion facilities as well as expensive equipment, setup, and post-processing. Furthermore, the moving area and gait cycles of the observed patients have limitations^{26,27}. Recent technological advancements have ushered in the development of reliable and cost-effective wearable sensors for gait analysis, facilitating their application beyond laboratory confines and during routine activities, thereby offering objective insights into individual gait patterns²⁸. A real-world gait analysis can enhance clinical assessments and influence associated interventions regardless of laboratory environmental constraints.

When assessing gait parameters across individuals, researchers and clinicians need to consider the reliability and influence of diurnal variation, which refers to changes that occur throughout the day, on the outcomes being assessed. Circadian rhythms influence physical, mental, and behavioral changes, including the regulation of alertness and fatigue²⁹. Moreover, they affect various physiological processes and cognitive and motor functioning^{30–33}. In addition to circadian rhythms, extrinsic factors that interact with the time of day, such as fatigue, exercise, mood, sleep, and food intake, also influence cognitive and motor function³³, collectively contributing to diurnal rhythm. As cognitive and motor functions are integral to postural control and gait^{33–36}, time of day may also affect gait. Evidence supports diurnal variations in lower limb muscle contractility and fatigability³⁷, postural control³⁸, and gait parameters such as walking speed and stride length during short-duration treadmill walking near the walk–run transition speed in healthy young adults³⁹. Conversely, another study observed that healthy young adults did not exhibit diurnal variations in stride time variability outcomes during continuous, over ground walking⁴⁰. Additionally, analysis of the spatial and temporal variables of footstep patterns revealed that patients with multiple sclerosis (MS) exhibited a symmetric, short-stepped, slow gait with minimal variability in performance from the morning to afternoon. The finding that self-rated fatigue significantly increased from the morning to afternoon, while walking patterns remained consistent, suggests that the mechanisms governing locomotion differ from those that regulate perceived fatigue⁴¹.

Analyzing gait offers crucial insights into the impact of ambulatory biomechanics on OA progression and aids in crafting effective therapeutic strategies. The aim of this study is to explore diurnal variations in daily life gait patterns of individuals with early KOA using an in-shoe motion sensor system (IMS), focusing on how gait characteristics vary between morning and afternoon sessions.

Materials and methods

Study design

This was a cross-sectional observational study designed to investigate diurnal variations in gait parameters among older adults with early-stage KOA.

Participants

A total of 30 older Japanese adults were recruited for this study between September 1 and September 19, 2023. Participants were recruited through a recruiting firm that engaged with the local community in Kyoto, Japan. The inclusion criteria for the study were as follows: participants had to be aged 60 years or older and have a clinical diagnosis of early-stage KOA, defined as Kellgren–Lawrence (KL) grades 1 or 2, confirmed through weight-bearing bilateral knee radiographs in accordance with the European League Against Rheumatism (EULAR) criteria^{42,43}. Additionally, participants needed to have the ability to walk outdoors independently without the use of assistive devices and to sustain continuous walking for at least 6 min without requiring rest. Furthermore, daily use of smartphones was required to facilitate data synchronization.

The exclusion criteria for the study were as follows: individuals with cognitive impairments or an inability to provide informed consent. Participants with a history of other musculoskeletal or neurological disorders that could affect gait, as well as those who had undergone prior lower limb surgery, were also excluded. Additionally, individuals with cardiac or respiratory conditions that limited physical activity or those reliant on walking aids were not eligible to participate in the study.

Out of 30 volunteers, 11 were excluded due to meeting exclusion criteria or inability to complete the study protocol, resulting in a final sample of 19 participants (11 females, 8 males). The majority of participants had bilateral KOA; unilateral involvement was present in two participants.

Ethics statement

Approval: This study was approved by the Medical Ethics Committee of Kyoto University, Japan (approval number: R3664-3).

Accordance: This study was conducted in accordance with ethical standards and guidelines.

Consent: Written informed consent was obtained from all participants prior to study initiation.

Equipment

Measurements were performed using the IMS (A-RROWG, NEC Corporation, Tokyo, Japan)⁴⁴. The IMS is a device used for gait analysis and is powered by an ARM Cortex-M4F MCU (nRF52832) from Nordic Semiconductor, featuring a 64-MHz CPU, 64-KB RAM, and 512-KB ROM. The IMS consists of an inertial measurement unit (IMU) incorporating a triaxial accelerometer and gyroscope (Bosch BMI160) for precise motion tracking. The accelerometer has a full-scale measurement range of ± 16 g with a sensitivity of approximately 0.488 mg/LSB, while the rate gyroscope has a full-scale range of $\pm 2000^\circ/\text{s}$ with a sensitivity of approximately 0.061 $^\circ/\text{s}$ /LSB. The device is powered by an ARM Cortex-M4F microcontroller and includes data storage and real-time clock capabilities. The IMS was mounted bilaterally on custom insoles placed under the foot arch to minimize discomfort. The sampling rates were 3.125 and 100 Hz in the standby mode and during measurement, respectively. Measurements were conducted from 6:00 AM to 10:00 PM at 2-minute intervals, during which the IMS actively collected data for the first minute before entering the power-saving sleep mode. In the active phase, vibration detection and stable walking discrimination were performed to measure the walking parameters for three steps within the ensuing 9 s, and the average value was recorded if detection was successful. Stable-gait identification involves determining whether the detected motion is consistent with level walking. The device monitors repetitive walking patterns by counting the number of instances in which posterior acceleration exceeds 3.5 g, a threshold typically observed during foot contact in normal walking. If this threshold is reached three times within a 5-second window, the device confirms stable walking and proceeds to measure gait parameters over the subsequent 9 s. Activities such as ascending or descending stairs, walking on uneven terrain, or frequent directional changes generally do not produce these stable repetitive acceleration patterns and are therefore automatically filtered out by the system's algorithm⁴⁴. If fewer than two steps were measured, then the attempt was considered a failure. Up to three measurement attempts were allowed per minute; if all attempts failed, no data were recorded for that interval. The device calculates 16 vital gait parameters as follows: walking speed, stride length, maximum (peak) dorsiflexion angle, maximum (peak) plantar flexion angle, foot height, Lateral foot displacement, toe-in/toe-out angle, roll angle of heel contact, roll angle of toe-off, cadence, stance time, swing time, pushing time, peak swing angular velocity, maximum speed during the swing phase, and foot clearance. These parameters have either been validated for accuracy in previous studies or have been calculated from validated parameters^{44–46}. Table 1 presents the definition of each parameter disclosed by the NEC Corporation.

Procedure

The data collection took place from 1st September to 19th September 2023. On the first day, participants were visited at Kyoto University, Japan, where anthropometric measurements were conducted and they were introduced to the IMS-equipped insoles. Participants who donned their everyday attire and comfortable walking shoes had their demographic and physical measurements (height, weight, and foot size) taken before the insertion of the IMS-equipped insoles. To prevent discomfort while walking, the IMS was mounted on an insole placed under the foot arch. A 5-meter gait confirmed that the patient had no difficulty or pain during gait. The protocol required participants to engage in normal daily activities for a minimum of 24 h, exclusively

Parameters	Definition	Units
Walking speed	Stride length divided by stride time	m/s
Stride length	Distance between successive points of initial contact of the same foot	cm
Maximum dorsiflexion direction angle	peak foot-sole angle in the dorsiflexion direction	degrees (°)
Maximum plantar flexion angle	peak foot-sole angle in the plantarflexion direction	degrees (°)
Foot height	Maximum vertical height of the foot during the swing phase	cm
Lateral foot displacement	Displacement of the foot in the medial-lateral direction during the swing phase	cm
Toe-in/toe-out angle	Average foot adduction/abduction angle during the swing phase	degrees (°)
Roll angle of heel contact	Angle of the foot about the longitudinal axis at heel contact	degrees (°)
Roll angle of toe-off	Angle of the foot about the longitudinal axis at toe-off	degrees (°)
Cadence	Number of steps per minute	steps/min
Stance time	Duration the foot is in contact with the ground during the gait cycle	seconds (s)
Swing time	Duration the foot is not in contact with the ground during the gait cycle	seconds (s)
Pushing time	Time from heel-off to toe-off during stance phase	seconds (s)
Peak swing angular velocity	Peak angular velocity of the foot during the swing phase	degrees/second
Maximum speed during swing phase	Maximum forward speed of the foot during the swing phase	m/s
Foot clearance	Minimum vertical distance between the foot and ground during swing phase	cm

Table 1. Definition of the parameters.

wearing the IMS outdoors and carrying a smartphone for data collection. Participants were instructed to walk for more than 10 min in each of the morning and afternoon sessions. Specific care instructions were provided to prevent damage to the IMS from water exposure. Gait data were collected using IMS placed in both the left and right shoes. Data were collected in the morning (6:00 AM–11:59 AM) and afternoon (12:00 PM–5:00 PM) sessions to accurately reflect the variance in daily gait behaviors. This timeframe was selected to ensure consistency across participants and to account for practical considerations related to participant schedules and data quality. The number of attempts recorded varied across participants and between morning and afternoon sessions, reflecting natural differences in daily activity patterns. This variability is expected in real-world settings where daily routines differ among individuals, influencing the frequency of stable walking attempts captured by the IMS.

Data analysis

The analysis commenced with a rigorous data-cleaning phase in which outliers were identified and excluded using the interquartile range method, with particular emphasis on the walking speed parameter to maintain data fidelity. Subsequent segmentation delineated the data into morning (6:00 AM–11:59 AM) and afternoon (12:00 PM–5:00 PM) sessions, facilitating a temporal comparison of the gait metrics. Data from both limbs were used in analysis. Descriptive statistics, including mean, standard deviation (SD), and range, were computed for each time segment to encapsulate the variability and central tendency of the gait parameters. Statistical integrity was further ensured through normality assessments for each parameter, guiding the use of either t-tests or Wilcoxon signed-rank tests, based on distribution characteristics, to ascertain significant temporal variations in gait dynamics. All statistical analyses were conducted using SPSS version 27 with a significance threshold of $p < 0.05$. Given the exploratory nature of this study, corrections for multiple comparisons (e.g., Bonferroni correction) were not applied to avoid an increased risk of Type II errors, which could obscure potentially meaningful findings.

Results

Table 2 presents the details of participant demographics and KOA severity. This study included 19 participants with an average age of 71.42 years (SD=4.15). The average weight and height of the participants were 58.89 kg (SD=10.62) and 1.55 m (SD=0.12), respectively, resulting in a mean BMI of 24.37 (SD=1.34). The sex distribution was relatively balanced, with 57.89% of the participants being female ($n = 11$) and 42.11% being male ($n = 8$). The average NRS pain score among participants was 2.34 ± 1.42 , with a range from 0 to 5. The median score was 2.0, indicating mild to moderate pain levels in this cohort. The severity of KOA was classified using the Kellgren–Lawrence grade, showing a distribution of 5.26% in grade 0 ($n = 2$), 52.63% in grade 1 ($n = 20$), 42.11% in grade 2 ($n = 16$).

In our study, we observed significant variations in gait parameters between the morning and afternoon sessions among older adults with KOA. Table 3 presents the comparison of gait parameters between morning and afternoon sessions. Walking speed decreased from an average of 1.06 (0.14) m/s in the morning to 0.99 (0.16) m/s in the afternoon ($p = 0.028$). The maximum dorsiflexion angle reduced from $20.34^\circ \pm 3.82^\circ$ to $18.80^\circ \pm 4.35^\circ$ ($p = 0.024$), and the maximum plantar flexion angle decreased from $63.40^\circ \pm 5.66^\circ$ to $60.79^\circ \pm 6.98^\circ$ ($p = 0.017$). Foot height slightly decreased from 12.99 ± 1.34 cm to 12.39 ± 1.63 cm ($p = 0.029$). The roll angle of heel contact increased from $4.60^\circ \pm 3.62^\circ$ to $5.53^\circ \pm 3.41^\circ$ ($p = 0.026$). Cadence decreased from 115.64 ± 10.2 steps/min to 112.40 ± 11.1 steps/min ($p = 0.018$). Stance time increased from 0.65 ± 0.05 s to 0.68 ± 0.06 s ($p = 0.003$). Pushing time increased from 0.211 ± 0.03 s to 0.229 ± 0.04 s ($p = 0.002$). The peak swing angular velocity and maximum speed during the swinging phase were reduced, with p-values of 0.006 and 0.023, respectively. Figure 1 shows the distribution of several key gait parameters that varied significantly between the time segments.

Characteristics		Mean (SD)
Age		71.42 (4.15)
Weight (kg)		58.89 (10.62)
Height (m)		1.55 (0.12)
BMI (Kg/m ³)		24.37 (1.34)
NRS Pain Score		2.34 (1.42)
		Count (%)
Sex	Female	11 (57.89%)
	Male	8 (42.11%)
KOA severity	KL grade 0	2.0 (5.26%)
	KL grade 1	20.0 (52.63%)
	KL grade 2	16.0 (42.11%)

Table 2. Summary of participant demographics and knee osteoarthritis severity. SD, standard deviation; BMI, body mass index; NRS, Numeric Rating Scale; KOA, knee osteoarthritis; KL, Kellgren–Lawrence grade. The KL grade 0 classification corresponds to the unaffected limbs of the two participants with unilateral KOA included in the final analysis.

Parameters	Morning mean (SD)	Afternoon mean (SD)	Significance (p-value)	Effect size (Cohen's d/effect size r)
Walking speed	1.06 (0.14) m/s	0.99 (0.16) m/s	0.028*	0.57
Stride length	109.69 (8.3) cm	104.76 (9.4) cm	0.064	0.45
Maximum dorsiflexion angle	20.34° (3.82)	18.80° (4.35)	0.024*	0.38
Maximum plantar flexion angle	63.40° (5.66)	60.79° (6.98)	0.017*	0.41
Foot height	12.99 (1.34) cm	12.39 (1.63) cm	0.029*	0.40
Lateral foot displacement	3.94 (1.33) cm	3.75 (0.98) cm	0.418	0.19
Toe-in/toe-out angle	15.51° (7.9)	15.97° (8.2)	0.398	-0.06
Roll angle of heel contact	4.60° (3.60)	5.53° (3.41)	0.026*	0.51
Roll angle of toe-off	-2.20° (5.49)	-1.92° (4.48)	0.461	-0.06
Cadence	115.64 (10.2) steps/min	112.40 (11.1) steps/min	0.018*	0.54
Stance time	0.65 (0.05) s	0.68 (0.06) s	<0.01**	-0.45
Swing time	0.398 (0.02) s	0.400 (0.02) s	0.570	-0.01
Pushing time	0.211 (0.03) s	0.229 (0.04) s	<0.01**	-0.53
Peak swing angular velocity	485.08°/s (57.79)	452.89°/s (48.87)	<0.01**	0.63
Maximum speed during swinging phase	3.50 m/s (0.27)	3.32 m/s (0.33)	0.023*	0.57
Foot clearance	20.4 (1.66) cm	19.8 (2.24) cm	0.13	0.28

Table 3. Comparison of gait parameters between morning and afternoon sessions in participants with knee osteoarthritis. SD, standard deviation. * $p < 0.05$. ** $p < 0.01$.

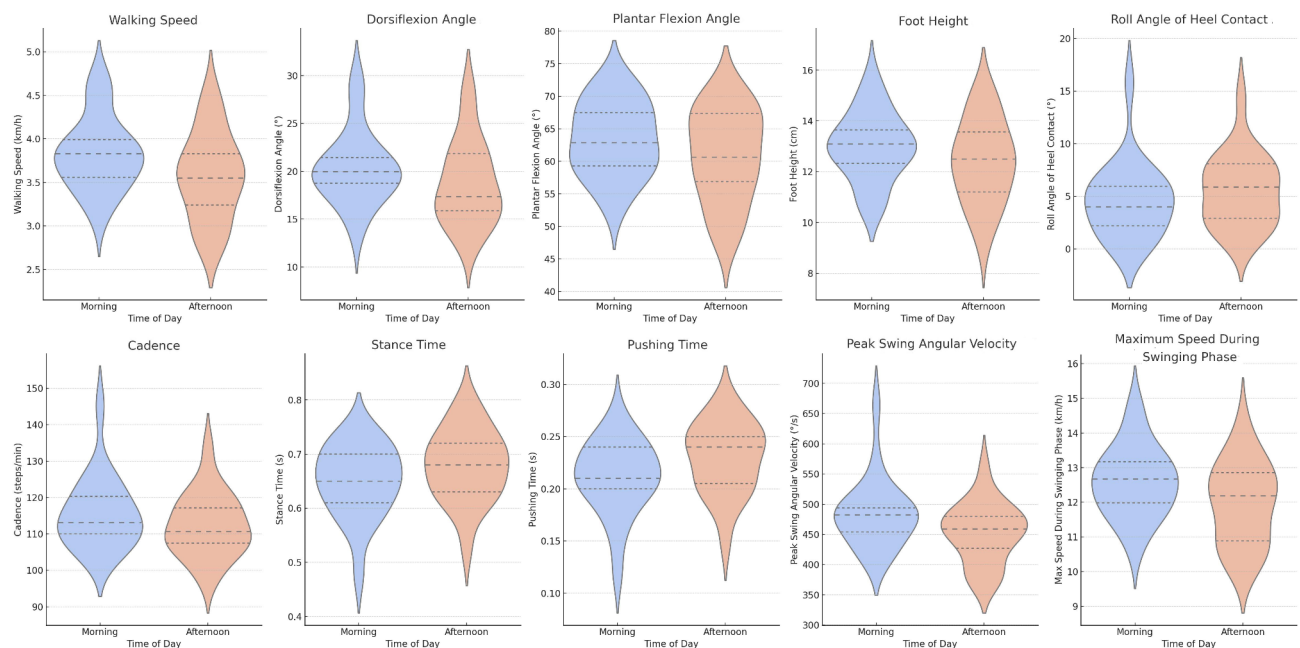


Fig. 1. The violin plots visualize the distribution of walking speeds, maximum dorsiflexion angle, maximum plantar flexion angle, foot height, roll angle of heel contact, cadence, stance time, pushing time, peak swing angular velocity and maximum speed during swinging phase of patients with knee osteoarthritis in the morning and afternoon sessions, providing insights into the data's distribution, mean values, variability, and comparison between different times of the day.

The other gait parameters examined did not indicate significant differences between the morning and afternoon sessions. Stride length changed from 109.69 ± 8.3 cm in the morning to 104.76 ± 9.4 cm in the afternoon ($p = 0.064$). Lateral foot displacement slightly varied from 3.94 ± 1.33 cm to 3.75 ± 0.98 cm ($p = 0.418$). Toe-in/toe-out angle experienced a minor adjustment from $15.51^\circ \pm 7.9^\circ$ to $15.97^\circ \pm 8.2^\circ$ ($p = 0.398$). The roll angle of the toe-off shifted from $-2.20^\circ \pm 5.49^\circ$ to $-1.92^\circ \pm 4.48^\circ$ ($p = 0.461$). Swing time changed from 0.39 ± 0.02 s to 0.40 ± 0.03 s ($p = 0.570$). Lastly, foot clearance slightly changed from 20.4 ± 1.66 cm to 19.8 ± 2.24 cm ($p = 0.132$).

Discussion

This study investigated diurnal variations in gait parameters among older adults with early-stage KOA using wearable sensor technology during normal daily activities. Our findings revealed significant decreases in walking speed, maximum dorsiflexion and plantar flexion angles, foot height, peak swing angular velocity, and maximum speed during the swing phase from morning to afternoon. Additionally, increases in the roll angle of heel contact, stance time, and pushing time were observed in the afternoon. These results suggest that time of day influences gait patterns in individuals with KOA.

The observed decrease in walking speed of approximately 0.07 m/s from morning to afternoon, while statistically significant, represents a relatively small change. However, even modest reductions in walking speed can be clinically meaningful in older adults, as slower gait speed has been associated with increased risk of adverse health outcomes¹³. However, it is important to note that the observed changes in walking speed may also be partially attributed to differences in the types of activities performed during the morning and afternoon. For example, participants may have engaged in more physically demanding or rushed activities in the morning, while afternoon sessions could have been influenced by accumulated fatigue or more relaxed routines. Therefore, caution should be exercised in attributing these changes solely to diurnal effects, as both activity-related factors and environmental factors may have influenced the results. The decreases in maximum dorsiflexion and plantar flexion angles indicate reduced ankle range of motion in the afternoon, which may reflect fatigue or joint stiffness accumulating throughout the day⁴⁷.

The increase in stance time and pushing time suggests that participants spent more time in the stance phase and during push-off in the afternoon. Prolonged stance time may indicate a cautious gait pattern, possibly due to increased joint discomfort or fatigue⁴⁸. The increased roll angle of heel contact could reflect alterations in foot placement strategies to compensate for knee discomfort.

These gait changes may be attributed to several factors. Diurnal fluctuations in pain and stiffness are common in KOA, with some patients experiencing worsening symptoms as the day progresses⁴⁹. Fatigue accumulating from daily activities may also contribute to altered gait patterns in the afternoon⁵⁰. Additionally, circadian rhythms can influence muscle strength and proprioception, potentially affecting gait⁵¹.

Our findings align with previous research indicating that gait parameters can vary throughout the day in individuals with musculoskeletal conditions⁴¹. However, studies in healthy adults have shown inconsistent results regarding diurnal variations in gait, suggesting that such variations may be more pronounced in populations with underlying joint pathology⁵².

It is important to consider whether the observed changes are clinically significant. While some changes, such as a 1.5° decrease in dorsiflexion angle, may seem small, they could cumulatively impact joint loading and mobility over time. Understanding these variations can inform the timing of gait assessments and interventions. For instance, while morning sessions, when gait is more stable, may enhance rehabilitation efficacy, training during symptom-prone periods can improve real-life functional capacity. A balanced approach, targeting both favorable and challenging times, may maximize therapeutic outcomes^{53,54}.

This study has several strengths that contribute to its scientific and clinical relevance. The use of wearable in-shoe motion sensors enabled continuous gait monitoring in real-world settings, enhancing ecological validity and providing insights into natural diurnal variations in gait parameters. By analyzing 16 distinct gait variables, including spatiotemporal and kinematic measures, the study offers a comprehensive assessment of gait dynamics in older adults with early-stage KOA. Additionally, the findings establish preliminary reference values for diurnal gait variations, which can serve as benchmarks for future research. As an exploratory investigation, this study lays the groundwork for future longitudinal studies and intervention trials aimed at understanding and optimizing gait performance in individuals with KOA. Several limitations should be acknowledged. The sample size was relatively small, which may limit the generalizability of the findings. The study population consisted of older adults with early-stage KOA; thus, results may not be applicable to those with advanced KOA. Additionally, we did not include a control group of age-matched individuals without KOA, which would help isolate the effects of KOA from age-related changes in gait. Furthermore, while BMI was not a controlled variable in this study, the BMI range of participants (21.5–26.5) suggests limited representation of individuals with higher BMI. Therefore, the potential influence of BMI on gait parameters should be considered when interpreting the results, particularly in relation to generalizability to populations with higher BMI. Another limitation is the grouping of unilateral and bilateral KOA participants, as these groups may exhibit distinct compensatory gait patterns that could introduce variability in the findings⁵⁵. Finally, it is important to note that multiple comparisons were conducted across 16 dependent variables without applying corrections like the Bonferroni adjustment, which could increase the risk of Type I errors. Although we considered statistical corrections to control for familywise error rates, we chose not to apply them due to the exploratory nature of this study and the interrelatedness of gait parameters. Applying stringent corrections could increase the risk of Type II errors, potentially obscuring meaningful diurnal variations in gait characteristics. Nonetheless, we acknowledge this as a limitation, and future studies with larger sample sizes may benefit from implementing appropriate statistical adjustments to balance the risks of Type I and Type II errors.

Factors such as frailty and sarcopenia, common in older adults, could also influence gait parameters and diurnal variations^{56,57}. While we collected data on KOA duration and pain levels, we did not assess muscle strength or fatigue objectively, which could provide further insights into the observed gait changes. Future studies with larger datasets should consider including a control group of age-matched individuals without KOA, distinguishing between unilateral and bilateral KOA participants, and adopting narrower time frames to provide more granular insights into diurnal gait variations.

Implications for KOA management

Understanding diurnal variations in gait parameters can inform the development of personalized management strategies for KOA. For example, interventions aimed at improving ankle mobility and strength may help mitigate the reductions in dorsiflexion and plantar flexion observed in the afternoon. Additionally, scheduling therapeutic exercises or activities during times of the day when gait patterns are more favorable could enhance intervention efficacy. While addressing functional limitations under less favorable conditions may improve the ability to function effectively during daily activities when they are most challenged. Wearable sensors provide valuable feedback that can guide these personalized approaches.

Conclusions

This study demonstrated significant diurnal variations in gait parameters among older adults with early-stage KOA, highlighting the influence of time of day on gait patterns in this population. The findings highlight the importance of considering diurnal variations when evaluating gait and planning interventions for individuals with KOA. Wearable sensor technology offers a valuable tool for capturing real-world gait data, facilitating personalized and time-sensitive approaches to KOA management.

Data availability

The data set used during the current study is available from the corresponding author upon reasonable request.

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Author contributions

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Declarations

Competing interests

The authors declare no competing interests.

Additional information

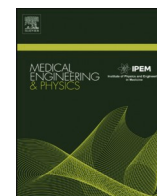
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Influence of surface type on outdoor gait parameters measured using an In-Shoe Motion Sensor System

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ABSTRACT

The objectives of this study were to measure outdoor gait parameters using an In-Shoe Motion Sensor System (IMS) and evaluate how different types of surfaces affect various gait dynamics. Accurate outdoor gait data are crucial for effective fall risk assessment because surface irregularities and tripping hazards often result in falls during walking. An IMS was used in this study to collect spatiotemporal, spatial, and foot parameters from 27 healthy adults walking on indoor asphalt, soil, and grass surfaces. Data were recorded during a 6-minute walk test, with measurements taken every 2 min and analyzed using the Statistical Package for the Social Sciences. The results showed significant differences in foot clearance, heel height, and gait cycle across surfaces. Walking on grass significantly increased foot height, swing time, and roll angle of heel contact. These findings may help develop interventions to prevent falls.

1. Introduction

Walking plays a crucial role in daily life, enabling mobility and independence [1]. Daily walking occurs in various surface environments, from smooth indoor corridors to uneven outdoor terrain. Traditional gait analysis has been conducted using 3D motion capture systems in controlled settings, such as laboratory corridors or treadmills [2,3].

However, the advent of Inertial Measurement Units (IMUs) has revolutionized gait analysis by allowing the measurement of walking parameters in more natural outdoor environments, closely resembling daily life conditions [4–6]. IMUs enable easy and continuous movement tracking, which provides a comprehensive understanding of gait dynamics across different surfaces. For instance, a study comparing indoor corridors with outdoor interlocking block pavements reported an increased ankle angle during outdoor walking [7]. However, these studies primarily focused on the differences between indoor and outdoor conditions and did not consider comparisons among multiple surfaces. Another study using IMUs measured the gait on sidewalks, dirt paths, gravel, grass, and wood chips, highlighting the increased effort required on uneven terrain owing to higher energy cost [8]. Despite these advancements, studies that comprehensively compare temporal, spatial, and spatiotemporal gait parameters across various outdoor surfaces

using IMUs are lacking.

Evaluating diverse gait parameters is essential for assessing overall daily activity levels, the extent of functional recovery post-injury, fall risk, and general endurance [9–12]. Specifically, a detailed gait analysis can reinforce rehabilitation strategies and fall prevention measures. Real-world conditions, where different surface types significantly affect walking stability and efficiency, necessitate such evaluations. Foot height and minimum toe clearance are crucial parameters often associated with tripping risk and walking stability [13,14]. Previous studies have shown that these parameters indicate the effect of overall gait efficiency and safety on individual adaptation to uneven surfaces. The roll angle is a critical indicator of the tarsal joint lock that can evaluate the rigidity and flexibility of the ankle, thereby influencing the kinetic chain of the lower limb [15]. Temporal parameters, such as cadence, stance time, and swing time, provide insights into the rhythm and balance of walking, which are essential for identifying potential gait impairments and designing appropriate interventions [16,17].

Therefore, this study aimed to evaluate and examine the effect of various walking surfaces on a wide range of gait parameters using IMUs. We sought to provide a detailed understanding of the influence of surface conditions on walking patterns by comparing gait dynamics on different surfaces—laboratory corridors, asphalt, dirt, and grass.

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We hypothesized that compared with indoor walking, outdoor walking exhibits larger spatial parameters, such as joint angles, and slower temporal parameters, such as walking speed and stance time. We expect the effects to be more pronounced in the order asphalt < dirt < grass, with grass among outdoor surfaces showing the most significant deviations owing to its uneven nature.

2. Materials and methods

2.1. Study design

This study utilized a cross-sectional design to investigate the research objectives.

2.2. Setting

The analysis of data was conducted at the facilities of Kyoto University (Kyoto, Japan).

2.3. Ethics statement

The study received approval from the Ethics Committee of the Graduate School and Faculty of Medicine, Kyoto University (approval no R3664–3). All participants gave written informed consent before participating in the study.

2.4. Participants

Participants were recruited from Kyoto University students. Eligibility criteria included individuals aged 18 years or older who could walk independently for at least 6 min without a break. Participants with neurological, orthopedic, cardiac, or respiratory conditions affecting mobility or walking were excluded.

2.5. Equipment

An In-shoe Motion Sensor System (IMS; A-RROWG, NEC Corporation, Tokyo, Japan) was employed for gait measurement. The IMS includes critical components such as an ARM Cortex-M4F microcontroller unit (nRF52832) by Nordic Semiconductor, featuring a 64 MHz CPU, 64 KB RAM, and 512 KB ROM. For precise movement tracking, it incorporates a Bosch IMU (BMI160), which measures 3-axis acceleration and angular velocity. The IMU has an accelerometer full-scale range of ± 16 g with a sensitivity of approximately 0.488 mg/LSB, a gyroscope full-scale range of $\pm 2000^\circ/\text{s}$ with a sensitivity of approximately $0.061^\circ/\text{s/LSB}$, and a sampling rate of 100 Hz. Additional components include an ABLIC EEPROM (S-24C32C, 32 K-bit) for data storage and an EPSON real-time clock (RX8130CE) for timekeeping.

The system captures gait parameters every 2 min, assessing both left and right sides. The device remains in sleep mode to conserve power and activates upon detecting motion. Specifically, the device uses a built-in high-g detection function to sense motion likely to be walking when the vertical acceleration exceeds 2.25g. Upon detecting such motion, the device's microcontroller unit (MCU) wakes up, and the IMU's sampling rate is increased to 100 Hz. It measures three steps, either consecutive or non-consecutive, within 9 s of detecting walking initiation. If fewer than three steps are recorded, the system considers it a measurement failure. If more than three steps are recorded, only the first three steps are averaged. This process is repeated up to three times within one minute. If all attempts fail, the measurement is marked as a failure, and no data is stored. Stable-gait identification is performed by counting instances where posterior acceleration exceeds 3.5g, typically during foot contact in normal walking. If three such instances are detected within 5 s, the device confirms stable walking and proceeds to measure gait parameters. If this threshold is not met, the device returns to sleep mode to conserve power.

The system automatically calculates 16 parameters validated in previous studies [18–20], including walking speed, stride length, maximum dorsiflexion angle, maximum plantar flexion angle, foot height, circumduction, toe-in/toe-out angle, heel contact roll angle, toe-off roll angle, cadence, stance time, swing time, pushing time, peak swing angular velocity, maximum speed during the swing phase, and foot clearance. Table 1 defines these parameters as outlined by the IMU developer.

2.6. Procedures

Before the measurements began, participants were asked to wear their regular clothing and footwear, and their age, height, weight, and foot size were documented. Insoles containing the IMS were then placed inside their shoes, followed by a 5-meter walk test to confirm that they experienced no pain or discomfort while walking. The study assessed walking parameters on four surface types commonly encountered in daily life: a laboratory corridor, asphalt, dirt, and grass (Fig. 1). These surfaces were selected to represent a variety of walking conditions, ranging from the controlled environment of an indoor corridor to outdoor surfaces like smooth asphalt and more uneven terrain such as dirt and grass, which were used as baseline experimental conditions. The order of surface testing was randomized using Microsoft Excel's random function. Measurements on asphalt, dirt, and grass were only conducted on dry days with no standing water. Participants were instructed to walk back and forth along a 30-meter path at a comfortable, everyday pace for 6 min, carrying a smartphone in their pocket to collect data. Observers were positioned at each end of the route to monitor normal walking and ensure participant safety, providing time updates every minute. All measurements were completed on the same day, with rest breaks allowed upon participant request.

2.7. Statistical analysis

The average and standard deviation of the walking parameters over six minutes were calculated for each surface type. Statistical analyses were conducted using the Statistical Package for the Social Sciences (SPSS) version 22.0 (IBM Corp., Armonk, NY, USA). The normality of the data distribution was assessed using the Shapiro–Wilk test. Parametric

Table 1
Definition of measured parameters.

Parameters	Definition
Walking speed	Stride length divided by stride time
Stride length	Distance between two consecutive ground contacts of the same foot
Maximum (peak) dorsiflexion angle	Peak foot-sole angle in dorsiflexion
Maximum (peak) plantar flexion angle	Peak foot-sole angle in plantarflexion
Foot height	Maximum vertical height of the midfoot (sensor placement) during the foot trajectory
Circumduction	Displacement in the medial–lateral direction during the swing phase
Toe-in/toe-out angle	Average adduction/abduction angle during the swing phase
Roll angle of heel contact	Plantar roll angle when the heel touches the ground
Roll angle of toe-off	Plantar roll angle when the toe leaves the ground
Cadence	Number of steps per minute
Stance time	Amount of time that the foot is on the ground during the gait cycle
Swing time	Amount of time that the foot is off the ground during the gait cycle
Pushing time	Time between the heel lift and toe leaving the ground
Peak swing angular velocity	Peak angular velocity of the foot during the swing phase
Maximum speed during swinging phase	Maximum forward speed of the swinging leg during the swing phase
Foot clearance	Maximum heel height of the foot trajectory

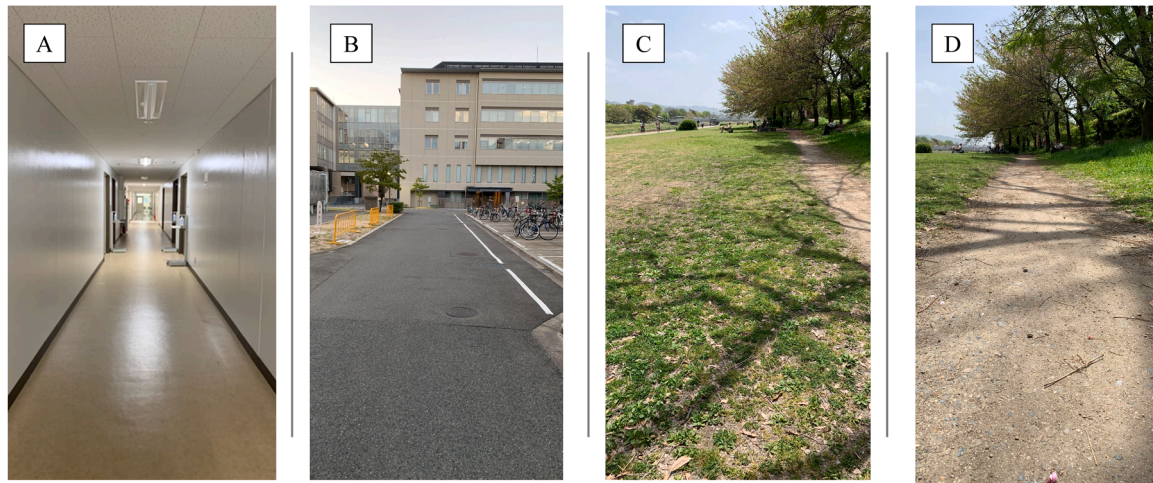


Fig. 1. (A) Laboratory corridor; (B) Asphalt, (C) Dirt, (D) Grass.

data were analyzed using repeated measures ANOVA. We first checked the assumption of sphericity for the repeated-measures ANOVA using Mauchly's test of sphericity [21]. If Mauchly's test indicated that the assumption of sphericity was met ($P \geq 0.05$), we proceeded with the standard repeated measures ANOVA. However, if Mauchly's test violated the sphericity assumption ($P < 0.05$), we applied the Greenhouse-Geisser correction to adjust the degrees of freedom for the F-tests. Significant differences between conditions were explored using a post hoc test with Bonferroni corrections. Non-parametric data were analyzed using the Friedman test, and significant differences were examined using the Bonferroni post-hoc test. To examine the interaction between sex and surface type on gait parameters, a two-way repeated-measures ANOVA was conducted. In this analysis, sex was treated as a between-subject factor and surface type as a within-subject factor. The potential effects of sex, surface type, and their interaction were evaluated, and partial η^2 was calculated as an effect size for all main and interaction effects. The significance level was set at $P < 0.05$.

3. Results

Table 2 shows the characteristics of the study participants. The study analyzed data from 27 participants, consisting of 10 males and 17 females. The average age, height, weight, and Body Mass Index (BMI) of the participants was 21.1 years, 164.4 cm, 57.3 kg, and 21.1 kg/m², respectively. No dropout or adverse events were reported during the study period.

Table 3 presents the gait parameters measured over six minutes across the four different surface types: laboratory, asphalt, dirt, and grass. This table provides the mean and standard deviation (SD) of various gait parameters. The foot height was highest on grass (17.38 $\pm$ 1.59 cm) compared to other surfaces. The roll angle of heel contact

Table 2
Characteristics of the study participants ($n = 27$).

Characteristics	Mean $\pm$ SD
Age, years	21.1 $\pm$ 2.3
Height (cm)	164.4 $\pm$ 6.5
Weight (kg)	57.3 $\pm$ 9.3
BMI, kg/m ²	21.1 $\pm$ 2.3
Foot size	24.2 $\pm$ 1.4
	Frequency (%)
Sex	
Male	10 (37.0)
Female	17 (63.0)

BMI: body mass index (kg per m squared); SD: standard deviation.

Table 3
Mean and standard deviation of gait parameters on different surface types.

	gait on lab corridor	gait on asphalt	gait on dirt	gait on grass
	mean $\pm$ SD	mean $\pm$ SD	mean $\pm$ SD	mean $\pm$ SD
Walking speed (km/h)	5.07 $\pm$ 0.59	5.26 $\pm$ 0.75	5.26 $\pm$ 0.58	5.16 $\pm$ 0.56
Stride length (cm)	144.02 $\pm$ 14.80	148.22 $\pm$ 17.64	147.31 $\pm$ 14.65	148.44 $\pm$ 12.59
Maximum (peak) dorsiflexion angle (degree)	29.97 $\pm$ 5.10	31.20 $\pm$ 4.96	30.61 $\pm$ 4.22	29.37 $\pm$ 3.55
Maximum (peak) plantar flexion angle (degree)	77.79 $\pm$ 6.15	78.92 $\pm$ 10.20	78.84 $\pm$ 5.80	78.04 $\pm$ 6.29
Foot height (cm)	16.13 $\pm$ 1.55	16.44 $\pm$ 2.25	16.43 $\pm$ 2.02	17.38 $\pm$ 1.59
Circumduction (cm)	3.72 $\pm$ 1.72	3.64 $\pm$ 2.03	3.38 $\pm$ 1.39	3.45 $\pm$ 2.29
Toe-in/toe-out angle (degree)	11.84 $\pm$ 7.00	12.54 $\pm$ 6.45	12.36 $\pm$ 6.59	11.55 $\pm$ 6.23
Roll angle of heel contact (degree)	5.66 $\pm$ 5.43	4.55 $\pm$ 4.89	4.33 $\pm$ 5.37	3.17 $\pm$ 5.12
Roll angle of toe-off (degree)	-1.44 $\pm$ 3.97	-0.52 $\pm$ 3.53	-0.73 $\pm$ 3.67	-0.89 $\pm$ 2.82
Cadence (steps/min)	116.52 $\pm$ 4.54	117.34 $\pm$ 5.76	118.33 $\pm$ 4.87	115.10 $\pm$ 5.65
Stance time (s)	0.63 $\pm$ 0.03	0.62 $\pm$ 0.05	0.61 $\pm$ 0.04	0.63 $\pm$ 0.04
Swing time (s)	0.403 $\pm$ 0.012	0.404 $\pm$ 0.016	0.406 $\pm$ 0.012	0.415 $\pm$ 0.014
Pushing time (s)	0.20 $\pm$ 0.02	0.20 $\pm$ 0.02	0.19 $\pm$ 0.02	0.20 $\pm$ 0.02
Peak swing angular velocity (degree/s)	557.37 $\pm$ 72.70	572.91 $\pm$ 87.15	582.36 $\pm$ 74.59	567.16 $\pm$ 84.49
Maximum speed during swinging phase (km/h)	16.11 $\pm$ 1.38	16.50 $\pm$ 1.72	16.23 $\pm$ 1.32	16.10 $\pm$ 1.09
Heel clearance (cm)	24.67 $\pm$ 1.97	24.94 $\pm$ 3.16	24.97 $\pm$ 2.48	25.77 $\pm$ 2.01

showed a decreasing trend from the laboratory surface (5.66 $\pm$ 5.43°) to the grass (3.17 $\pm$ 5.12°). Stance time was relatively consistent across surfaces, with slight variations, ranging from 0.61 $\pm$ 0.04 s on dirt to 0.63 $\pm$ 0.04 s on laboratory and grass. Similarly, swing time showed minimal variation, ranging from 0.40 $\pm$ 0.01 s in the laboratory to 0.41 $\pm$ 0.01 s on dirt and grass. Heel clearance was highest on grass (25.77 $\pm$ 2.01 cm) and lowest in the laboratory setting (24.67 $\pm$ 1.97 cm), indicating differences in foot elevation between surfaces.

Table 4 presents the results of the repeated-measures ANOVA and Friedman test for various gait parameters across different surface types.

Table 4

Comparison of gait parameters between different surface types.

	Repeated measures ANOVA			Friedman test	Significances
	Mauchly's test	Sphericity assumption	Greenhouse-Geisser correction		
Walking speed (km/h)	0.407	0.099			
Stride length (cm)	0.412	0.151			
Maximum (peak) dorsiflexion angle (degree)	0.074	0.102			
Maximum (peak) plantar flexion angle (degree)	<0.001		0.647		
Foot height (cm)				<0.001	*
Circumduction (cm)				0.77	
Toe-in/toe-out angle (degree)				0.123	
Roll angle of heel contact (degree)				0.004	*
Roll angle of toe-off (degree)				0.412	
Cadence (steps/min)	0.465	<0.001			*
Stance time (s)	0.766	<0.001			*
Swing time (s)	0.011		<0.001		*
Pushing time (s)	0.099	0.006			*
Peak swing angular velocity (degree/s)				<0.001	*
Maximum speed during swinging phase (km/h)	0.302	0.426			
Heel clearance (cm)				<0.001	*

The following parameters showed significant differences:

Regarding the spatial parameters, significant differences were found in foot height ($P < 0.001$), roll angle of heel contact ($P = 0.004$), and heel clearance ($P < 0.001$) across different surfaces.

Concerning the temporal parameters, significant differences were observed in cadence, stance time, swing time, and pushing time. Repeated-measures ANOVA showed significant differences in cadence ($P < 0.001$), stance time ($P < 0.001$), swing time ($P < 0.001$ with Greenhouse-Geisser correction), and pushing time ($P = 0.006$) across the different surfaces.

For the spatiotemporal parameters, significant differences were observed in the peak swing angular velocity ($P < 0.001$).

Other gait parameters did not show significant differences across the surfaces, including walking speed, stride length, maximum dorsiflexion angle, and maximum plantar flexion angle.

Table 5 presents the post-hoc test results for the significant gait parameters identified in the repeated-measures ANOVA and Friedman test, comparing the different surface types: laboratory, asphalt, dirt, and grass.

Table 5

Post hoc test results.

	lab vs asphalt	lab vs dirt	lab vs grass	asphalt vs dirt	asphalt vs grass	dirt vs grass
Bonferroni	P-value	P-value	P-value	P-value	P-value	P-value
Foot height (cm)	0.019	0.178	<0.001	1	0.137	0.014
Roll angle of heel contact (degree)	0.375	0.037	0.003	1	0.652	1
Cadence (steps/min)	1	0.047	0.218	0.842	0.007	<0.001
Stance time (s)	1	0.004	1	0.209	0.464	0.001
Swing time (s)	1	1	<0.001	1	<0.001	<0.001
Pushing time (s)	1	0.143	1	0.544	0.402	0.004
Peak swing angular velocity (degree/s)	0.011	0.001	1	1	0.078	0.011
Heel clearance (cm)	0.016	0.137	<0.001	1	0.586	0.104

3.1. Comparison with laboratory surface

For the spatial parameters, significant differences were observed in foot height, and heel clearance between the laboratory surface and grass ($P < 0.001$). The roll angle of the heel contact was significantly lower on the grass surface than on the laboratory surface ($P = 0.003$).

Significant differences were noted in the stance, swing, and pushing times regarding the temporal parameters. The stance time differed significantly between the laboratory and dirt surfaces ($P = 0.004$). The swing time was significantly longer on the grass than on the laboratory surface ($P < 0.001$). The pushing time was not significantly different between the laboratory and the other surfaces.

3.2. Comparison among asphalt, dirt, and grass surfaces

Regarding spatial parameters, significant differences in foot height were found between dirt and grass surfaces ($P = 0.014$).

Concerning temporal parameters, the stance and swing times demonstrated significant differences between the three surfaces. Significant differences were observed in cadence between asphalt and grass ($P = 0.007$) and between dirt and grass ($P < 0.001$). The stance time was also significantly different between dirt and grass surfaces ($P < 0.001$). The swing time and pushing time were significantly different between asphalt and grass ($P < 0.001$ and $P = 0.003$, respectively) and between dirt and grass ($P < 0.001$ and $P = 0.004$, respectively).

Table 6 presents the mean and standard deviation of the gait parameters for males and females under different surface conditions. Foot height, toe-in/out angle, and heel clearance notably differed between sexes. Specifically, foot height on grass was the highest for both males and females, with males exhibiting slightly higher values overall.

Table 7 presents the results, showing that no significant interaction effects between sex and surface type were observed for any of the analyzed parameters. However, significant main effects of sex were identified for foot height ($F = 19.656$, $p < 0.001$, partial $\eta^2 = 0.159$), toe-in/out angle ($F = 7.440$, $p = 0.007$, partial $\eta^2 = 0.067$), and heel clearance ($F = 25.372$, $p < 0.001$, partial $\eta^2 = 0.196$). This suggests that the effect of sex on gait dynamics remains consistent regardless of the surface type.

4. Discussion

4.1. Summary of findings

This study investigated the gait parameters of healthy adults while walking on four different surfaces: an indoor laboratory corridor, outdoor asphalt, outdoor dirt, and outdoor grass. The primary aim of this

Table 6
Mean and standard deviation of gait parameters by sex across surface conditions.

	Sex	Lab (Mean ± SD)	Paved (Mean ± SD)	Dirt (Mean ± SD)	Grass (Mean ± SD)
Walking speed (km/h)	Male	4.98 ± 0.51	5.23 ± 0.48	5.22 ± 0.47	5.12 ± 0.43
	Female	5.12 ± 0.66	5.28 ± 0.9	5.29 ± 0.67	5.19 ± 0.65
Stride length (cm)	Male	142.18 ± 13.12	148.49 ± 12.71	147.76 ± 12.69	148.96 ± 11.4
	Female	145.04 ± 16.33	148.07 ± 20.67	147.06 ± 16.38	148.16 ± 13.85
Maximum (peak) dorsiflexion angle (degree)	Male	28.75 ± 4.48	30.24 ± 3.91	30.04 ± 3.63	29.28 ± 3.26
	Female	30.64 ± 5.56	31.73 ± 5.62	30.92 ± 4.7	29.42 ± 3.89
Maximum (peak) plantar flexion angle (degree)	Male	75.76 ± 4.94	79.51 ± 2.92	77.28 ± 4.91	76.77 ± 5.60
	Female	78.92 ± 6.75	78.59 ± 12.91	79.71 ± 6.36	78.75 ± 6.87
Foot height (cm)	Male	16.92 ± 1.09	17.57 ± 1.10	17.65 ± 1.40	18.24 ± 1.22
	Female	15.69 ± 1.66	15.81 ± 2.55	15.75 ± 2.08	16.91 ± 1.65
Circumduction (cm)	Male	3.72 ± 1.50	3.35 ± 1.51	3.23 ± 1.53	2.52 ± 1.21
	Female	3.72 ± 1.92	3.8 ± 2.35	3.46 ± 1.39	3.96 ± 2.67
Toe-in/out angle (degree)	Male	14.91 ± 4.70	14.47 ± 5.07	14.82 ± 3.82	13.2 ± 4.10
	Female	10.14 ± 7.77	11.47 ± 7.17	11 ± 7.64	10.64 ± 7.25
Roll angle of heel contact (degree)	Male	8.45 ± 3.80	5.93 ± 2.77	6.75 ± 3.25	6 ± 4.57
	Female	4.11 ± 5.82	3.79 ± 5.80	2.99 ± 6.05	1.6 ± 4.98
Roll angle of toe-off (degree)	Male	−1.16 ± 3.79	−0.45 ± 3.08	−1.47 ± 3.28	−2.03 ± 2.74
	Female	−1.6 ± 4.27	−0.56 ± 3.93	−0.31 ± 4.0	−0.25 ± 2.81
Cadence (steps/min)	Male	115.86 ± 2.65	116.39 ± 2.99	116.97 ± 3.18	113.72 ± 2.64
	Female	116.88 ± 5.46	117.86 ± 7.0	119.08 ± 5.66	115.87 ± 6.87
Stance time (s)	Male	0.63 ± 0.02	0.62 ± 0.02	0.62 ± 0.03	0.64 ± 0.02
	Female	0.62 ± 0.04	0.61 ± 0.05	0.6 ± 0.04	0.62 ± 0.05
Swing time (s)	Male	0.4 ± 0.01	0.4 ± 0.01	0.4 ± 0.01	0.42 ± 0.01
	Female	0.4 ± 0.01	0.4 ± 0.02	0.41 ± 0.01	0.41 ± 0.02
Pushing time (s)	Male	0.2 ± 0.01	0.2 ± 0.01	0.2 ± 0.01	0.21 ± 0.02
	Female	0.2 ± 0.02	0.2 ± 0.03	0.19 ± 0.02	0.2 ± 0.03
Peak swing angular velocity (degree/s)	Male	539.38 ± 60.36	568.56 ± 58	560.87 ± 55.26	541.7 ± 65.11
	Female	567.36 ± 80.5	575.32 ± 103.5	594.3 ± 84.41	581.3 ± 94.45
Maximum speed during swinging phase (km/h)	Male	16.15 ± 1.10	16.56 ± 1.38	16.69 ± 1.28	16.31 ± 1.0
	Female	16.09 ± 1.58	16.47 ± 1.97	15.98 ± 1.33	15.98 ± 1.18
Heel clearance (cm)	Male	25.81 ± 1.40	26.68 ± 1.39	26.6 ± 1.62	27.06 ± 1.36
	Female	24.04 ± 2.04	23.97 ± 3.55	24.06 ± 2.51	25.05 ± 2.04

study was to evaluate and examine the effect of various walking surfaces on a wide range of gait parameters using IMUs. A comparison between the laboratory and outdoor surfaces revealed that foot height and heel clearance were higher on the outdoor surfaces, with significant differences observed between the laboratory and asphalt and between the laboratory and grass. The roll angle of the heel contact was lower on the

Table 7
Effects of sex and surface on gait parameters.

	Factor	F Value	P Value	Partial η^2 (Effect Size)
Walking speed	Intercept	6550.613	<0.001	0.984
	Sex	0.432	0.512	0.004
	Surface	0.592	0.622	0.017
	Sex × Surface (Interaction)	0.023	0.995	0.001
Stride length	Intercept	9155.988	<0.001	0.989
	Sex	0.006	0.94	0
	Surface	0.557	0.645	0.016
	Sex × Surface (Interaction)	0.082	0.97	0.002
Maximum (peak) dorsiflexion angle	Intercept	11,090.061	<0.001	0.991
	Sex	1.257	0.265	0.012
	Surface	0.263	0.852	0.008
	Sex × Surface (Interaction)	0.364	0.779	0.01
Maximum (peak) plantar flexion angle	Intercept	4360.739	<0.001	0.977
	Sex	1.445	0.232	0.014
	Surface	0.66	0.579	0.019
	Sex × Surface (Interaction)	0.174	0.914	0.005
Foot height	Intercept	9193.254	<0.001	0.989
	Sex	19.656	<0.001	0.159
	Surface	2.334	0.078	0.063
	Sex × Surface (Interaction)	0.214	0.887	0.006
Circumduction	Intercept	334.757	<0.001	0.763
	Sex	1.936	0.167	0.018
	Surface	0.326	0.807	0.009
	Sex × Surface (Interaction)	0.693	0.558	0.02
Toe-in/out angle	Intercept	376.321	<0.001	0.783
	Sex	7.44	0.007	0.067
	Surface	0.139	0.936	0.004
	Sex × Surface (Interaction)	0.139	0.936	0.004
Roll angle of heel contact	Intercept	98.145	<0.001	0.486
	Sex	13.409	<0.001	0.114
	Surface	1.037	0.379	0.029
	Sex × Surface (Interaction)	0.279	0.841	0.008
Roll angle of toe-off	Intercept	7.539	0.007	0.068
	Sex	0.701	0.404	0.007
	Surface	0.276	0.842	0.008
	Sex × Surface (Interaction)	0.538	0.657	0.015
Cadence	Intercept	48,719.447	<0.001	0.998
	Sex	2.56	0.113	0.024
	Surface	1.68	0.176	0.046
	Sex × Surface (Interaction)	0.066	0.978	0.002
Stance time	Intercept	23,171.502	<0.001	0.996
	Sex	2.912	0.091	0.027
	Surface	1.251	0.295	0.035
	Sex × Surface (Interaction)	0.072	0.975	0.002
Swing time	Intercept	91,453.454	<0.001	0.999
	Sex	0.068	0.794	0.001
	Surface	4.529	0.005	0.116
	Sex × Surface (Interaction)	0.363	0.78	0.01
Pushing time	Intercept	9174.946	<0.001	0.989
	Sex	2.092	0.151	0.02
	Surface	1.339	0.266	0.037
	Sex × Surface (Interaction)	0.102	0.959	0.003
Peak swing angular velocity	Intercept	4941.664	<0.001	0.979
	Sex	2.798	0.097	0.026
	Surface	0.449	0.719	0.013
	Sex × Surface (Interaction)	0.196	0.899	0.006

(continued on next page)

Table 7 (continued)

	Factor	F Value	P Value	Partial η^2 (Effect Size)
Maximum speed during swinging phase	Intercept	13,244.087	<0.001	0.992
	Sex	1.127	0.291	0.011
	Surface	0.423	0.737	0.012
	Sex $\times$ Surface (Interaction)	0.286	0.835	0.008
Heel clearance	Intercept	12,848.3994	<0.001	0.992
	Sex	25.372	<0.001	0.196
	Surface	1.114	0.347	0.031
	Sex $\times$ Surface (Interaction)	0.24	0.868	0.007

outdoor surfaces, with significant differences between the laboratory surface and the grass. The swing time tends to be longer on the outdoor surfaces, with significant differences between the laboratory and the grass. Among the outdoor surfaces, grass showed higher foot height, cadence, stance time, and swing time than dirt and asphalt. Our hypothesis stated that, compared with indoor walking, outdoor walking would exhibit larger spatial parameters, such as joint angles, and slower temporal parameters, such as walking speed and stance time. Among the outdoor surfaces, we expected the effects to be more pronounced in the order asphalt < dirt < grass, with grass showing the most significant deviation owing to its uneven nature. These results partially supported our hypothesis. As hypothesized, outdoor walking showed higher spatial parameters, specifically foot height and heel clearance, than indoor walking. In addition, the roll angle of the heel contact was lower on outdoor surfaces, particularly grass, which aligns with our prediction of greater deviations on uneven surfaces. Temporal parameters, such as swing time, were also longer on outdoor surfaces, especially on grass, supporting our hypothesis. However, our hypothesis that walking speed and stance time would be slower on outdoor surfaces than on indoor surfaces was not fully supported. Although the stance time showed some differences, the walking speed did not exhibit significant deviations across the surfaces. Thus, the hypothesis was validated, confirming several aspects; however, some elements were not fully supported by the data.

4.2. Novelty and significance of using IMS

One of the fundamental innovations of this study is the comprehensive measurement of clinically important gait parameters using IMUs. IMUs can measure a wide range of gait parameters and allow accurate and continuous movement tracking in real-world environments, thereby facilitating a comprehensive understanding of gait dynamics across different surfaces. In a previous study, IMUs were attached to the waist, thighs, and shanks to measure the gait on paved roads, sidewalks, cobblestones, grass, and inclines [22]. However, that study focused solely on the hip and knee joint movements, excluding the foot from the analyses. In contrast, the present study utilized IMUs to collect a variety of detailed gait parameters in outdoor environments and observed significant differences in foot-related parameters, such as foot height and heel clearance. Additionally, we found significant differences in the roll angle of the heel contact, which has not been previously evaluated. Moreover, temporal parameters such as cadence, stance time, and swing time also showed significant differences. Thus, the use of IMUs to measure a wide range of parameters enhances the significance of the study by better reflecting real-life conditions.

Another innovative aspect of this study was the use of an IMU system capable of long-duration measurements. The IMU used in this study was a thin device inserted into a dedicated insole, requiring no special operation from the participants once installed. This allows for continuous, long-term measurements in daily life without the need for specific instructions or interventions [23]. Moreover, this capability makes the

system particularly applicable to long-term evaluations of populations with medical conditions. Our findings contribute to a deeper understanding of the factors that influence gait dynamics in daily life and offer valuable insights for interpreting data from future long-term evaluations.

4.3. Interpretation of results

The observed significant increases in foot height, stance time, and swing time, as well as the significant decrease in cadence while walking on grass, can be interpreted as consistent, interrelated changes. With the increase in foot height without changes in other parameters, such as stride length, the trajectory of the foot during the swing phase may be extended, leading to an increased swing time. Additionally, cadence is calculated from the number of steps per minute; therefore, if the stance and swing time increase, the number of steps per minute decreases, resulting in a lower cadence. A previous study also confirmed that compared to other environments, walking on grass increases the energy cost of walking and reduces walking efficiency; similar trends were observed in this study [8].

These results can be attributed to several factors. The significant increase in foot height and swing time may be because the height of the grass obstructs the forward foot movement during the swing phase, thus increasing the risk of tripping and falling, necessitating a higher foot lift to avoid these obstacles. Uneven surfaces were associated with an increased risk of tripping [24] and foot height [25], and we observed similar trends on the grass surfaces in this study.

The extension of stance time and decreased cadence may also be from efforts to improve walking stability on uneven surfaces. A previous study conducted on treadmills in laboratory settings has reported that ankle inversion and eversion angle changes and muscle activity increase around the ankle are necessary to maintain balance on irregular surfaces, leading to a reduction in walking speed [2]. Although we did not observe a significant decrease in walking speed in this study, there was a trend towards slower walking on grass in the outdoor environment and a confirmed significant reduction in the roll angle of the heel contact, indicating ankle eversion. These findings suggest that the participants may have been adjusting their gait to maintain balance on an uneven grass surface, consistent with previous research [2,26,27].

In a previous study, walking on unstable rocky surfaces decreased walking speed and stride length while showing larger external rotation of the ankle on unstable surfaces [28]. However, this tendency was not observed in the present study. The differences in the results could be attributed to several factors: previous studies compared flat surfaces, urethane mats, and unstable rocky surfaces, whereas this study compared laboratory corridors, paved roads, dirt, and grass; previous studies simulated various environments in a laboratory setting, whereas this study used real outdoor environments; and previous studies employed motion capture cameras and pressure measurement systems, whereas this study utilized IMUs. These methodological and environmental differences may have influenced the trends observed in our results.

Furthermore, this study found significant main effects of sex on foot height, toe-in/out angle, and heel clearance; however, no interaction effects between sex and surface type were observed for any gait parameter. These results suggest that while sex significantly influences specific spatial gait parameters, this influence does not vary across different surface conditions. The absence of interaction effects indicates that the effect of sex on these parameters remains consistent regardless of the walking surface.

4.4. Limitations

This study had several limitations. The sample consisted solely of young adults, limiting the generalizability of the findings to older adults or those with walking impairments. This study did not measure

electromyography (EMG) or confirm whether these trends were more pronounced in older adults with a higher risk of falling; therefore, strong conclusions cannot be drawn. Additionally, the measurements were conducted under dry weather conditions, preventing assessing the impact of wet or slippery conditions on the gait parameters.

4.5. Future research directions

Future studies should include a broader range of participants, such as older adults and individuals with mobility impairment, to better understand how different populations adapt to various walking surfaces. Research under different weather conditions, such as rain or snow, would also provide a more definitive understanding of how environmental factors influence gait. Long-term data collection using IMS can reveal variations in walking patterns over time and under different daily conditions.

4.6. Practical applications

The results of this study provide practical insights for designing rehabilitation and exercise programs. Ankle sprains and chronic ankle instability can restrict ranges of motion during ankle inversion and eversion [29,30]. Monitoring the roll angle of the heel contact while walking on uneven surfaces, such as grass, using IMUs can help compare injured and noninjured limbs. This approach facilitates the evaluation of whether functional recovery extends to activities of daily living and enables training tailored to real-world conditions. Walking training on various surfaces can enhance functional recovery in patients with spinal cord injury [30]. By incorporating surfaces such as grass and other uneven terrains into training programs and monitoring improvements in foot height, toe-in/out angle, and roll angle of heel contact, therapists can use these parameters as indicators to improve balance and stability.

Furthermore, the application of IMUs in real-time feedback systems is a promising avenue for enhancing rehabilitation outcomes. Providing immediate feedback on gait parameters such as the roll angle of heel contact, foot height, and heel clearance enables users to adjust their walking patterns, improve stability, and reduce fall risk. For instance, wearable IMU systems can alert users when their gait deviates from the recorded daily walking patterns, prompting corrective actions. This could be particularly beneficial for older adults or individuals recovering from lower limb injuries.

Finally, these findings support physical therapists and exercise professionals in creating more effective and diverse training environments that mimic real-world walking conditions. The use of IMUs in daily settings offers valuable long-term and real-time feedback for monitoring and improving gait, thereby reducing the risk of falls and enhancing overall mobility.

5. Conclusion

Walking on grass significantly alters gait parameters, increasing foot height, swing time, and roll angle of the heel contact. These changes were likely due to the high and uneven nature of grass surfaces. The use of IMUs in this study provided detailed real-world gait data. Understanding these differences is crucial for developing strategies for improving walking stability and preventing falls in various environments.

Declaration of generative AI and AI-assisted technologies in the writing process

During the preparation of this study, the authors used ChatGPT4/OpenAI to improve language and readability. After using this tool/service, the authors reviewed and edited the content as required and took full responsibility for the publication.

Data statement

The data dataset generated during the current study are available from the corresponding author upon reasonable request, after approval by the data access committee.

Ethical approval

The study received approval from the Ethics Committee of the Graduate School and Faculty of Medicine, Kyoto University (approval no R3664–3).

CRediT authorship contribution statement

Hiroki Shimizu: Conceptualization, Methodology, Formal analysis, Investigation, Data curation, Writing – original draft, Writing – review & editing, Visualization. **Kyoma Tanigawa:** Conceptualization, Methodology, Investigation, Methodology, Writing – review & editing. **Anuradhi Bandara:** Conceptualization, Methodology, Investigation, Methodology, Writing – review & editing. **Shinichi Kawamoto:** Conceptualization, Methodology, Investigation, Methodology, Writing – review & editing. **Shota Suzuki:** Conceptualization, Methodology, Investigation, Methodology, Writing – review & editing. **Momoko Nagai-Tanima:** Supervision, Project administration, Writing – review & editing. **Tomoki Aoyama:** Conceptualization, Methodology, Resources, Writing – review & editing, Supervision, Project administration, Funding acquisition.

Declaration of competing interest

The authors declare the following financial interests/personal relationships which may be considered as potential competing interests:

Tomoki Aoyama reports equipment, drugs, or supplies was provided by NEC Corporation. If there are other authors, they declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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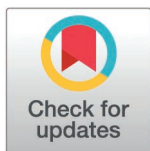
RESEARCH ARTICLE

The effects of high-heeled shoes on gait parameters in healthy adult women

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Abstract

Background

Walking is essential in daily life, and footwear type significantly affects walking patterns. High-heeled shoes increase the risk of knee osteoarthritis and falls in women. Traditional studies often use treadmills or unfamiliar footwear, which may not reflect daily walking. This study investigated the impact of high-heeled shoes on walking parameters in healthy adult women using in-shoe motion sensors.

Methods

Seventeen healthy adult women without pain during walking participated. They walked for 6 minutes along a 30-meter corridor wearing high-heeled shoes and sneakers. Walking data were recorded using an in-shoe motion sensor system every 2 minutes. The average of three valid consecutive steps was calculated automatically. Statistical analysis compared the mean walking parameters between the high-heel and sneaker groups.

Results

The high-heel group showed significantly reduced foot clearance, stride length, peak plantar angle in dorsiflexion, and peak plantar angle in plantarflexion, and significantly greater toe-out angle in spatial parameters. Spatiotemporal parameters revealed significantly reduced walking speed and maximum swing phase speed in the high-heel group. No significant differences were observed in temporal parameters between the groups.

Conclusions

Since this study collected gait data under conditions similar to daily life, it provides data suitable for practical applications and may contribute to future research

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evaluating everyday gait. Additionally, future studies should include a broader range of participants and incorporate measurement devices capable of capturing hip and knee joint movements, providing a more comprehensive evaluation of the effects of high-heeled shoes on gait in healthy adult women.

Introduction

Walking is a crucial means of transporting the body safely and efficiently [1]. Additionally, it has beneficial effects on both physical and mental health, enhances human quality of life (QOL), and plays a significant role in the prevention and treatment of various diseases [2]. As such, walking is indispensable in daily life. It is also known that walking is influenced by various external factors [3].

One of these external factors is footwear. Shoes come in various types and have multiple functions, including protecting the feet from external environments, assisting walking functions, and adapting to foot deformities. These functions are related to the occurrence of various conditions such as falls, osteoarthritis, and back pain [4]. Numerous studies have investigated the impact of shoes on walking parameters [5,6]. For example, research comparing walking barefoot and wearing athletic shoes by Shakoor et al. found that wearing athletic shoes decreases step length, increases the number of steps, and reduces joint angles of the lower limbs [5]. Thus, research on how shoes affect walking is of great importance [7].

One type of shoe that warrants attention is high-heeled shoes. High-heeled shoes are characterized by heels higher than the forefoot, narrow toe boxes, and curved soles [8]. High-heeled shoes have been widely worn by women worldwide for centuries, with 37% to 69% of women wearing them regularly and for an average of 1–8 hours per day [8]. Despite their regular use, high-heeled shoes are associated with medical conditions, such as osteoarthritis, and can cause frequent falls [9]. The incidence of such injuries doubled from 7.1% to 14.2% between 2002 and 2012 [10]. Therefore, it is crucial to investigate the impact of high-heeled shoes on the walking parameters of healthy adult women [7]. Several previous studies have examined this effect [9,11–13]. For instance, it has been shown that wearing high-heeled shoes does not change the temporal walking parameters such as stance time and cadence [11]. Additionally, a study by Barkema et al. noted that the higher the heel height, the greater the maximum knee adduction moment, indicating a relationship between heel height and knee osteoarthritis [9].

Research on the effects of high-heeled shoes on gait parameters has significantly progressed. However, many evaluations have been conducted using a treadmill with participants walking at a constant speed [14]. Treadmill walking occurs under flat, steady-speed conditions, which differs from everyday walking environments [15]. For example, because a treadmill maintains a fixed speed, participants cannot walk at their natural pace, potentially suppressing their typical gait patterns [16,17]. Additionally, on a treadmill, individuals often need to adjust their posture and gait mechanics to match the treadmill's motion in order to maintain visual and balance cues, which is known to influence gait biomechanics [15]. In certain studies, participants wore

unfamiliar shoes that may not accurately reflect their usual gait patterns in daily life [11,14]. Shoes that are unfamiliar or uncomfortable for an individual often do not fit well with the person's foot shape or gait style, potentially affecting the gait biomechanics [18]. Specifically, when the shoe's size or shape does not align with the foot, it may cause foot pain and fatigue, while also reducing stability and increasing the risk of falls during walking [19]. Therefore, conditions different from those in daily life may alter an individual's natural gait patterns and may not accurately reflect habitual walking [11].

As a more suitable evaluation method, Inertial Measurement Units (IMUs) can be used. IMUs can estimate three-dimensional movement from angular velocity and acceleration [20,21]. Due to their low cost, small size, and ease of use, they have been widely used to evaluate daily life walking in recent years [22]. By using the In-Shoe Motion Sensor system (IMS), which measures the acceleration and angular velocity of the feet using IMUs and calculates walking parameters, it is possible to obtain walking data comparable to daily life more accurately than conventional studies [23].

Therefore, this study aimed to evaluate the impact of high-heeled shoes on gait parameters in healthy adult women using the IMS system. Unlike previous studies that typically used treadmill walking at a constant speed or required participants to wear unfamiliar shoes, this study allowed participants to walk in their own footwear that fitted their feet and that they were accustomed to, under conditions similar to those of daily life. This approach enabled us to capture a more natural gait pattern, closely resembling everyday walking, and to more accurately reflect the effects of high-heeled shoes on gait parameters.

Materials and methods

Study design

This was an intervention study.

Participants

The participants in this study were healthy adult women enrolled at Kyoto University. The inclusion criteria were individuals aged 18 years and older who could walk for at least 6 minutes without rest. To minimize the influence of unfamiliarity with high-heeled shoes on the results, we selected participants with extensive experience wearing high heels in various settings, such as work, social events, shopping, or formal occasions, based on previous studies [14]. The exclusion criteria were individuals with neurological or orthopedic conditions that could affect walking function. The purpose of the study was thoroughly explained to the participants, and their consent was obtained prior to measurement.

Ethics statement

This study was approved by the Kyoto University Medical Ethics Committee (Approval number: R3664-3). Prior to study initiation, written informed consent was obtained from all participants in accordance with the ethical standards and guidelines.

Equipment

The IMS System (A-RROWG, NEC Corporation, Tokyo, Japan) was used for the measurements. The IMS is a device used for gait analysis and consists of key components including an ARM Cortex-M4F MCU (nRF52832 by Nordic Semiconductor). This component features a 64 MHz CPU, 64 KB of RAM, and 512 KB of ROM. Additionally, it includes a Bosch IMU (BMI160) for precise motion tracking, an ABLIC EEPROM (S-24C32C, 32K-bit) for storage, and an Epson real-time clock (RX8130CE). This IMS maintains measurement accuracy by applying real-time correction through two algorithms: the Three-dimensional zero-velocity-update (3D ZUPT) algorithm and the On-line stride segmentation algorithm. The on-line stride segmentation algorithm enables stable foot-flat detection using the plantar angle, ensuring precise stride segmentation. Additionally, the 3D ZUPT algorithm effectively compensates for drift errors, allowing for accurate stride parameter computation within the IMS device [24]. Due to the implementation of these algorithms, no manual calibration

of the IMU was conducted in this study. This device performs gait measurements at 2-minute intervals, capturing gait parameters on both sides. The system detects the gait cycle during flat walking, operates for 1 min, and then enters sleep mode to conserve energy. During this 1-minute period, it detects vibrations and identifies stable walking patterns, and upon detecting the start of a gait cycle, it measures the parameters for three steps within the following 9 s. These steps do not need to be consecutive; if the three steps are successfully measured within a specified time, the average value is recorded. If fewer than three steps were measured, the average value was not recorded. Additionally, this measurement process can be repeated up to three times within a 1-minute window; if all three attempts fail, the measurement for that period is considered unsuccessful, and no data is recorded [24–26].

The following 16 parameters are calculated: stride length, maximum (peak) dorsiflexion angle, maximum (peak) plantar flexion angle, foot clearance, foot height, circumduction, toe-in/out angle, roll angle of heel contact, roll angle of toe-off, walking speed, peak swing angular velocity, maximum speed during swinging phase, stance time, swing time, pushing time, and cadence. Table 1 below, disclosed by NEC Corporation, categorizes the definitions of each parameter into spatial, spatiotemporal, and temporal parameters. These parameters were validated in previous studies for accuracy or were derived from validated parameters [24–26]. High-heeled shoes were defined as “footwear with a heel elevated higher than the forefoot, a narrow toe box, and a curved sole” [11], while sneakers were defined as “light footwear that has a top made of cloth and a bottom made of rubber.” Based on previous studies that used shoes with a heel height of 3 cm or more as high-heeled shoes [27,28], the present study defined high-heeled shoes as those with a heel height of 3 cm or greater. Accordingly, the habitual shoes used in this study refer to the footwear that the participants considered to be the ones they wore most frequently during daily activities, such as commuting and shopping.

Procedure

The measurements were conducted in a 30-meter corridor at the Faculty of Human Health Sciences building, at Kyoto University. First, we verified whether the shoes worn by the participants met the aforementioned criteria. Subsequently, we measured the age (years), height (cm), body mass(kg), size of the shoes(cm), heel height (cm) of the high-heeled shoes, and mass (g) of each shoe. Prior to data collection, participants walked 5 m with insoles equipped with an IMS inserted into

Table 1. Definitions of each walking parameter.

Classification	Parameters	Definition
Spatial parameters	Stride length (cm)	Distance from the starting point to end point of the foot trajectory
	Maximum (peak) dorsiflexion angle (degree)	Peak foot-sole angle in dorsiflexion
	Maximum (peak) plantar flexion angle (degree)	Peak foot-sole angle in plantarflexion
	Foot clearance (cm)	Maximum heel height of the foot trajectory
	Foot height (cm)	Maximum vertical height of the foot trajectory
	Circumduction (cm)	Displacement in the medial–lateral direction during the swing phase
	Toe-in/out angle (degree)	Average adduction/abduction angle during the swing phase
	Roll angle of heel contact (degree)	Plantar roll angle when the heel touches the ground
Spatiotemporal parameters	Roll angle of toe-off (degree)	Plantar roll angle when the toe leaves the ground
	Walking speed (km/h)	Stride length divided by stride time
	Peak swing angular velocity (degree/s)	Peak angular velocity of the foot during the swing phase
Temporal parameters	Maximum speed during swinging phase (km/h)	Maximum forward speed of the swinging leg during the swinging phase
	Stance time(s)	Amount of time that the foot is on the ground during the gait cycle
	Swing time(s)	Amount of time that the foot is off the ground during the gait cycle
	Pushing time (s)	Time between the heel contact and toe leaving the ground.
	Cadence (steps/min)	Number of steps per min

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their shoes to ensure that they experienced no discomfort during walking. For data collection, Participants placed a smart-phone, used for data recording, in either pocket of their pants, as there were no restrictions on which side to use. They were instructed to begin walking from one end of the corridor, turn around at the opposite end, and continue this back-and-forth walking for 6 min. The measurements were conducted twice: once in sneakers and once in high-heeled shoes, with the order of conditions determined using Excel's random function. Participants were given the option to take a break between trials upon request. The observer stood in the middle of the pathway during the test to ensure that the participants walked normally and without safety issues, providing updates on the remaining time every minute. If no data were collected, the 6-minute walking test was repeated. In this study, real-life conditions refer to natural walking situations that closely resemble daily life, without the use of treadmills or experimental footwear. Natural gait refers to an individual's inherent walking pattern under conditions without external constraints or artificial instructions, and without any intentional modification of their gait.

Data analysis

Statistical analyses were performed using the JMP PRO16 software (SAS Institute). First, the Shapiro-Wilk test was conducted to determine whether each dataset followed a normal distribution. For data that were normally distributed, a paired t-test was used, because the same participants were measured under different conditions (high-heeled shoes and sneakers). For data that did not follow a normal distribution, the Wilcoxon signed-rank test, suitable for paired datasets, was applied. The group wearing high-heeled shoes was referred to as the "high-heeled group" (HG) and the group wearing sneakers as the "sneaker group" (SG). Data are presented as mean standard deviation $\pm$ standard deviation (SD). The effect size was evaluated using Cohen's d to assess the magnitude of the differences between groups, with thresholds defined as Small (0.2–0.49), Medium (0.5–0.79), and Large (≥ 0.8) [29]. The statistical significance level for this study was set at less than 5%.

Results

A. Physical characteristics (Table 2)

There were 17 participants in the study, with no dropouts or adverse events reported. The mean values were as follows: age, 23.4 years; height, 161.78 cm; body mass, 53.46 kg; BMI, 20.38; size of shoes, 24.15 cm; mass of high-heeled shoes, 201.76 g; mass of sneakers, 265.00 g; heel height of high-heeled shoes, 4.93 cm (Table 1).

B. Comparison of walking parameters between high-heeled and sneaker shoes (Table 3)

From the Table 3, in the comparison between the high-heeled group and the sneaker group, spatial parameters showed that the high-heeled group had a significantly shorter stride length, reduced peak dorsiflexion angle of the foot, reduced

Table 2. Basic information of participants.

Characteristics	Mean $\pm$ SD(N=17)
Age(years)	23.40 $\pm$ 1.23
Height(cm)	161.78 $\pm$ 6.14
Body Mass(kg)	53.46 $\pm$ 5.61
BMI(kg/m ²)	20.38 $\pm$ 1.09
Size of shoes(cm)	24.15 $\pm$ 1.01
Mass of heel shoes(g)	201.76 $\pm$ 37.12
Mass of sneakers(g)	265.00 $\pm$ 59.48
Heel height of heel shoes(cm)	4.93 $\pm$ 1.17

SD, standard deviation; BMI, body mass index.

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Table 3. Comparisons of the mean values of each parameter in each category.

Classification	Parameters	HG	SG	P-value	Effect Size (Cohen's d)
Spatial parameters	Stride length (cm)	130.42±18.14	143.88±16.09	0.0006*	-1.09
	Maximum (peak) dorsiflexion angle (degree)	24.21±5.25	29.88±5.13	0.0017*	-0.94
	Maximum (peak) plantar flexion angle (degree)	62.33±11.15	78.31±6.49	<0.0001*	-1.47
	Foot clearance (cm)	21.17±3.31	24.05±2.15	0.0005*	-1.09
	Foot height (cm)	13.33±2.40	15.70±1.76	0.0023*	-0.90
	Circumduction (cm)	4.10±1.58	3.72±2.01	0.5397	0.15
	Toe-in/out angle (degree)	12.05±8.82	10.14±7.97	0.0476*	0.54
	Roll angle of heel contact (degree)	5.21±4.86	3.71±5.70	0.2343	0.30
	Roll angle of toe-off (degree)	-2.05±6.17	-1.36±4.52	0.5178	-0.17
Spatiotemporal parameters	Walking speed (km/h)	4.61±0.68	5.08±0.66	0.0005*	-1.09
	Peak swing angular velocity (degree/s)	531.64±95.96	565.64±75.54	0.0932	-0.45
	Maximum speed during swinging phase (km/h)	15.15±1.65	15.97±1.59	0.0085*	-0.75
Temporal parameters	Stance time(s)	0.62±0.05	0.62±0.04	0.8271	-0.06
	Swing time(s)	0.41±0.04	0.40±0.01	0.3225	0.03
	Pushing time (s)	0.19±0.03	0.20±0.02	0.6805	-0.13
	Cadence (steps/min)	117.22±6.41	116.92±5.69	0.8030	0.06

SD, standard deviation. *p<0.05.

Effect size (Cohen's d): Small = 0.2–0.49, Medium = 0.5–0.79, Large ≥ 0.8.

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peak plantar flexion angle of the foot, and significantly greater toe angle compared to the sneaker group. Regarding spatiotemporal parameters, the high-heeled group exhibited significantly reduced walking speed and maximum speed during swinging phase. There were no significant differences observed between the groups in temporal parameters.

Discussion

This study aimed to evaluate the effect of high-heeled shoes on gait parameters in healthy adult women using an IMS system. In previous studies, participants often wore unfamiliar shoes and measurements were commonly conducted on a treadmill at a constant speed, which may not accurately reflect natural gait patterns in real-life settings. In contrast, this study utilized a compact and user-friendly IMS to collect gait data while participants wore their own familiar shoes. This approach allowed for accurate collection in a manner closer to daily life conditions, providing foundational data for assessing the effects of footwear under more realistic circumstances.

As a result, in terms of spatial parameters, the high-heeled group showed significantly shorter stride length, reduced maximum dorsiflexion angle, reduced maximum plantar flexion angle, and significantly greater toe-out angle compared to the sneaker group, with effect sizes ranging from moderate to high. Additionally, in spatiotemporal parameters, the high-heeled group exhibited significantly reduced walking speed and maximum speed during the swinging phase, with large effect sizes. However, no significant differences were observed between the groups in temporal parameters.

Regarding spatial parameters such as walking speed, stride length, and swing speed, all were significantly reduced in the high-heeled group, with large to moderate effect sizes: Cohen's d = -1.09 for stride length and walking speed, and Cohen's d = -0.75 for swing speed. These results strongly suggest that wearing high-heeled shoes has a substantial impact on propulsion and walking efficiency. Generally, wearing high heels narrows the base of support during walking, prompting individuals to adopt a safer and more stable gait pattern [30,31]. In a previous study [24], the RMSE for stride length and walking speed based on measurements taken over three steps was reported as 0.042 and 0.071, respectively. In contrast, the differences between the HG and SG in the present study were 0.1346 for stride length and 0.47

for walking speed, both of which substantially exceeded the reported RMSE values. Therefore, these differences are considered to exceed the range of measurement error and can be interpreted as practically meaningful and statistically significant group differences. Additionally, when wearing high heels, the reduced base of support during walking leads to increased muscle activity to maintain balance, which in turn tends to increase energy expenditure. To minimize this increase in energy expenditure, it has been shown that walking speed and stride length are reduced as a strategy [32,33]. In addition to these stable and efficient walking strategies, previous studies have shown that wearing high heels brings the ground reaction force closer to the ankle joint, and the curved shape of the shoe sole causes the ankle to remain in a plantarflexed position [11]. These factors can shorten the length of the triceps surae muscle and Achilles tendons. Consequently, compared to sneakers or barefoot walking, ankle plantarflexion moments decrease, and the effect of the stretch-shortening cycle on the muscle-tendon complex is reduced [34,35]. The stretch-shortening cycle is a process in which muscles and tendons quickly stretch and then shorten, store, and release elastic energy, and plays a crucial role in efficient force production [35]. However, the shortened state of the muscle-tendon complex induced by wearing high heels is thought to impede this function, leading to decreased energy efficiency and a weaker push-off force. As a result, the calculated parameters were significantly different between the two groups.

Similarly, foot clearance and foot height were significantly reduced in the high-heeled group. A very large effect size for foot clearance (Cohen's $d = -1.09$) and a large effect size for foot height (Cohen's $d = -0.90$) strongly suggest that foot clearance was more reduced in the high-heeled group. In a previous study [24], the RMSE for foot height was reported to be 0.017. In contrast, the difference between the HG and SG observed in this study was 0.0464, which substantially exceeds the RMSE. Therefore, the difference in foot clearance can be considered to exceed the margin of measurement error and represents a practically and statistically significant intergroup difference. As noted above, wearing high-heeled shoes is suggested to reduce the push-off force of the ankle. Additionally, previous studies investigating the role of the ankle in walking have shown that ankle push-off contributes to up to 50% of the power needed for leg swinging during the terminal stance and early swing phases [36–38]. This reduction in ankle push-off may decrease the leg swing speed, which is essential for the double-pendulum effect during walking [39]. In the double-pendulum effect, an increased hip swing speed results in the relative lifting of the lower leg, consequently increasing foot clearance. This occurs as the hip and knee joints work in coordination, with the hip's movement influencing the lower leg's motion. Specifically, a rapid change in the hip angle causes the lower leg to rise, allowing the foot to clear obstacles before making contact with the ground. In the heel group, however, a reduction in leg swing speed led to insufficient knee flexion in the double-pendulum mechanism, causing the lower leg to remain relatively low and potentially resulting in reduced foot clearance.

In fact, prior research comparing joint angles in the sagittal plane under different heel heights at a constant walking speed found that while hip flexion during the swing phase remained unaffected by heel height, the knee flexion angle significantly decreased as heel height increased [40–42]. Moreover, a decrease in the foot clearance height can increase the risk of tripping over small obstacles or uneven surfaces while walking, potentially leading to falls [43]. High-heeled shoes also tilt the toes downward and elevate the heel, reducing the contact area of the foot with the ground, which may further destabilize balance. This combination of reduced foot clearance and a limited base of support likely exacerbates the risk of falls [14].

Thus, the high-heeled group showed significantly reduced stride length, reduced peak dorsiflexion angle of the foot, reduced peak plantar flexion angle of the foot, reduced walking speed, and swing phase maximum velocity, while the toe angle during the swing phase was significantly greater. This change demonstrated a moderate effect size (Cohen's $d = 0.54$), suggesting that individuals wearing high heels may adjust foot positioning to enhance stability. A greater toe angle indicates a larger ankle abduction angle. Studies investigating the influence of heel height on the anterior-posterior joint moments of the reduced limb have shown that as heel height increases, knee internal rotation moments increase, potentially increasing the load on the medial compartment of the knee [9,44]. Conversely, walking with toes pointing outward reduces knee internal rotation moments and decreases the load on the medial compartment [45]. In this study, our

data suggest that the high-heeled group adopted a walking strategy with toes pointing outward to reduce the load on the medial compartment of the knee. Additionally, since knee joint movements during stance and toe direction were not measured in this study, further experiments are necessary. In a previous study [24], the RMSE for toe angle during the swing phase was reported to be 2.071. In contrast, the difference between the HG and SG observed in this study was 1.91, which is approximately the same as the RMSE. Therefore, it is possible that the observed difference in toe angle may not be entirely free from the influence of measurement error.

There are limitations in this study; the participants were limited to healthy adult women from the same university, which restricts the generalizability of our findings. To apply these results to other age groups, sexes, or individuals with specific health conditions, further studies with more diverse participants are necessary. Second, the sample size in this study was limited to 17 participants, which poses constraints on statistical robustness and the power to detect effects. In particular, to adequately assess the impact of individual differences, it is essential to include a larger number of participants. Future studies should increase the sample size to enhance statistical robustness and improve the power to detect effects. Third, we were unable to capture the movements of the hip and knee joints, making it challenging to fully assess how these joints affect gait mechanics, particularly balance and posture control. Specifically, it remains unclear to what extent high-heeled shoes stress on these joints under near-daily conditions and how this stress might increase the risk of falls or lead to long-term joint disorders. Future research should incorporate devices that measure the movements of these joints to provide a more comprehensive evaluation of the impact of different footwear types on gait and balance. Finally, for the toe angle, the difference between groups was comparable to the RMSE, making it difficult to conclusively determine that the observed difference was statistically meaningful. Therefore, future studies should adopt more precise measurement techniques to verify the reliability of this difference.

Conclusions

This study investigated the effects of high-heeled shoes on walking parameters in healthy adult women using the IMS. As a result, in terms of spatial parameters, the high-heeled group showed significantly shorter stride length, reduced maximum dorsiflexion angle, reduced maximum plantar flexion angle, and significantly greater toe-out angle compared to the sneaker group, with effect sizes ranging from moderate to high. Additionally, in spatiotemporal parameters, the high-heeled group exhibited significantly reduced walking speed and maximum speed during the swinging phase, with large effect sizes. However, no significant differences were observed between the groups in temporal parameters. These findings suggest that wearing high-heeled shoes may have an impact on gait instability and efficiency, potentially increasing the risk of falls and musculoskeletal strain. Future studies should include a more diverse population, such as older adults or individuals with joint disorders, and incorporate motion capture systems or wearable sensors that can measure reduced limb joint dynamics. These additional studies would allow for a comprehensive understanding of how footwear affects not only the ankle but also hip and knee joint movements across a wider population. Because this study collected gait data under conditions similar to those of daily life, it provides data suitable for practical applications and may contribute to future research focused on evaluating gait in everyday scenarios.

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