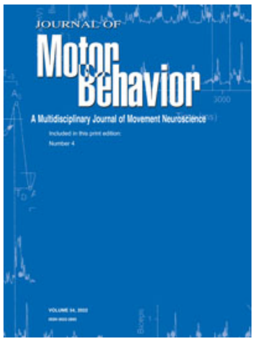

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RESEARCH ARTICLE

Walking Speed Alters Barefoot Gait Coordination and Variability

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ABSTRACT. Using the dynamic system approach, we examined the pattern and variability of inter-joint coordination in barefoot and shod walking in 20 women at three speeds: SLOW, FAST, and comfortable walking speed (CWS). We found that barefoot and shod walking used different coordination strategies to cope with increasing walking speed. As walking speed increased, ankle-knee coordination patterns between shod and barefoot became less different ($p < 0.00001$), and ankle-hip coordination patterns became more different ($p < 0.001$). Compared to shod, barefoot walking had significantly lower coordination variability in mid stance of knee-hip at CWS and FAST and late swing of ankle-hip at SLOW and CWS with medium effect (effect size 0.61–0.74). Future research should investigate the connection between the decreased coordination variability and joint tissue stress to understand the impact of barefoot walking on the lower extremity joints.

Keywords: barefoot walking, gait coordination, coordination variability

INTRODUCTION

Many features of modern footwear, especially athletic shoes, are designed to provide comfort and protection during walking and running. For instance, a cushioned sole of athletic shoes absorb the impact of heel strike, a force that is a possible cause of musculoskeletal injury (Collins & Whittle, 1989; Lafortune & Hennig, 1992; Whittle, 1999). It is not clear, however, if shoes have a protective effect throughout stance. A systemic review article (Franklin et al., 2015) reported that, while barefoot walking produced significantly lower ground reaction force (GRF) at heel strike and significantly lower ankle moment at early stance compared to shod walking, ankle and knee moments and GRF at late stance were significantly higher in the barefoot condition. What does seem clear is that walking barefoot produces a different pattern of GRF and joint moments compared to shod walking. Barefoot walking also changes the lower limb kinematics (Morio et al., 2009; Zhang et al., 2013).

The change of joint moment and adjustment of kinematics in the lower extremity can also impact neuromuscular control and movement dynamics of locomotion (Biewener & Daley, 2007). From the dynamic system view of human movement, walking and running requires coordination among different joints that must be integrated into functional units rather than acting alone (Robertson et al., 2013). Kurz and Stergiou (2004) estimated the shank-foot coordination pattern and found that it was more out-of-phase in barefoot than shod running in the sagittal and frontal planes. Gruber et al. (2011) found no significant difference, however, in shank-foot coordination between barefoot and shod running. Walking and running have their own distinct neuromuscular control characteristics, especially during the stance phase (Cappellini et al., 2006), so the coordination pattern observed in barefoot running should not be utilized to infer barefoot walking. Recently, Romer et al. (2019) examined the thigh-shank and shank-foot coordination during barefoot walking and found that the shank-foot coordination was more out-of-phase in some regions of the stride cycle compared to shod walking, but no quantitative test of the effect of footwear on coordination pattern was performed. More importantly, most related studies examined the coordination pattern of barefoot walking in a self-selected comfortable walking speed (CWS) only. People, however, often walk in a non-CWS. The ability to walk fast is considered a functional vital sign (Middleton et al., 2015). A recent meta-analysis study (Fukuchi et al., 2019) showed that walking speed had a strong effect on spatiotemporal(e.g., stride length), kinematic(e.g., joint range of motion) and kinetics(e.g., GRF) parameters of gait, but neither barefoot walking nor gait coordination were examined. Walking speed influenced inter-joint coordination differently in young and old participants, indicating that different neuromuscular control strategies to cope with the increasing walking speed were used (Chiu & Chou, 2012). Slower than CWS changed the inter-joint coordination pattern during the swing phase for shod walking (Little et al., 2019), while Wang et al. (2017) found that the peak knee stress exhibited a quadratic growth as walking speed increased when barefoot. Both footwear and walking speed constrains locomotion (Sparrow & Newell, 1998), but it remains unknown whether or not one constraint (walking speed) interacts with the other constraint (footwear) in the neuromuscular control of gait. By varying footwear and walking speed and observing the change of coordination pattern, better insight into the neuromuscular control of barefoot walking is possible.

Of note, Romer et al. (2019) also found lower thigh-shank coordination variability but higher shank-foot coordination variability in barefoot walking than in shod walking. Coordination variability has been examined extensively to assess the neuromuscular control of

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locomotion in orthopedic (Bonacci et al., 2020; Cunningham et al., 2014; Desai & Gruber, 2021; Hamill et al., 2012) and neurological (Perrin et al., 2018; Socie & Sosnoff, 2013) disorders. The movement variability is considered an intrinsic property of any biological system (Bernshteĭn, 1967) and reflects the ability of the motor system to reliably perform a motor task under different locomotion conditions (Stergiou & Decker, 2011), such as different footwear. Compared to shod walking, barefoot walking significantly increase the variability of impact GRF (Broscheid & Zech, 2016). Although minimalist shoes are designed to mimic the effect of bare feet, barefoot walking has significantly higher stride length variability than walking with minimalist shoes (Petersen et al., 2020). If minimalist shoes are an intermediate type between normal athletic shoes and bare feet in terms of the thickness and hardness of the sole of shoes, barefoot walking may also have higher movement variability than normal athletic shoes. Such an inference, however, should not be made without empirical assessment. Also, the variability of gait parameters such as stride length or single joint kinematics are not equivalent to coordination variability. Because coordination variability in Romer et al.'s study (2019) was estimated across the entire stride cycle, it is also unclear whether or not coordination variability changes if it is estimated within each phase of the stride cycle.

The alteration of movement dynamics has often been associated with health conditions (Stergiou & Decker, 2011). It has been suggested that modern footwear is mismatched with the functional environment from which human's foot evolved, thus may lead to various pathological conditions (Lieberman, 2013; Sichting et al., 2020). Although there is limited evidence to support the health-related outcomes of barefoot walking over shod walking (Hollander et al., 2017), the examination of coordination pattern and coordination variability can provide insight into the neuromuscular control in relation to footwear change.

This study builds on the previous work by examining the pattern and variability of coordination of ankle-knee, knee-hip, and ankle-hip in barefoot and shod walking at three self-selected walking speed categories (SLOW, CWS, FAST). While previous studies estimated gait coordination using continuous relative phase (CRP) (Kurz & Stergiou, 2004; Romer et al., 2019), CRP can only estimate segmental coordination (Lamb & Stöckl, 2014). Consequently, CRP is difficult to associate with previous insights, as segmental kinematics in barefoot are infrequently reported in the literature. Gait coordination in our study, thus, is estimated between joints using the vector coding method. We characterize the difference of coordination patterns between shod and barefoot at each speed category within the same the subject and examine the change in the coordination pattern as walking speed varies. Coordination variability is estimated for early stance, mid stance, late stance, and late swing gait phases. We hypothesize that 1) walking speed changes the difference of coordination patterns between barefoot and shod walking, and 2) coordination variability estimated in early stance, mid stance, late stance, and late swing phase is significantly different between barefoot and shod condition in the three speed categories.

METHODS

Subjects

Twenty college-aged female subjects (height: 167.3 ± 6.3 centimeter; weight: 62.4 ± 8.6 kilogram; leg length: 79.8 ± 3.8 centimeter) volunteered to participate in this study. All were habitually shod and free from lower limb injury and signed an informed consent form approved by the Institution Review Board of the University of Washington.

Instrument and Experimental Protocol

The experiment was conducted in the Human Motion Analysis Lab (HMAL) at the University of Washington. Reflective markers were placed on both left and right sides, as appropriate, on anatomical landmarks (greater trochanter of femur, lateral femoral condyle, medial femoral condyle, tibial tuberosity, lateral malleolus and medial malleolus of the ankle, Achilles tendon insertion on the calcaneus, and between the 2nd and 3rd metatarsal heads). The 3 D trajectories of the markers were captured with six high-speed cameras (Qualisys, Inc, Sweden) with a sampling frequency of 100 Hz. The floor surface in the lab was industrial grade, uncushioned vinyl tile.

Subjects were instructed to walk straight on an 8-meter walkway while shod and barefoot at three speed categories: SLOW, CWS, and FAST. Subjects wore their own athletic shoes in shod conditions. Since our primary purpose is to study gait coordination in barefoot walking and compare it with gait coordination in preferred daily-wearing athletic shoes, no restriction on the specific type of athletic shoes was enforced. Speed category was not randomized, and the sequence was SLOW-CWS-FAST. Subject completed all the shod trials for three speed categories first and then completed barefoot trials. Subjects completed trials under verbal instruction of each speed category. The verbal instruction for SLOW was "walk at a stroll. You have nowhere to be and are enjoying yourself. It is a pleasant day, and you have good companionship." For CWS: "walk at your normal pace. You have somewhere to go, but you are not in a hurry." For FAST: "You are in a hurry—like you are late for your bus— but you have to maintain your speed for 5 minutes. You are walking as fast as you can, but not running, and not so fast that you can not stop within a stride." The verbal instruction was delivered to subjects consistently.

The subjects were given sufficient time to familiarize themselves with the protocol before any change of footwear or speed category. Walking speed was calculated based on each extracted stride cycle.

Data Processing and Statistical Analysis

Single Joint Kinematics

Each subject performed ten successful trials for each footwear condition and each speed category, so each subject completed 60 walking trials in total (two footwear conditions × three speed categories × ten trials = 60 trials in total). A trial was considered "successful" for this analysis if at least one stride cycle of the left and of the right limb was fully captured. Due to the limited capture volume, the number of fully captured stride cycles in one successful trial varied from 1 to 3. To maintain consistency in the analysis, only the first-fully captured stride cycle of the left and right limb was extracted. The first-fully captured stride cycle was approximately in the middle of the capture volume and there were approximately 1–2 strides before and 2–3 strides after the first-fully captured stride cycle, depending on each subject's walking speed and stride length. A stride cycle was defined using the two consecutive lowest positions of the ipsilateral heel markers (the point of Achilles tendon insertion on the calcaneus) in the vertical axis as initiation and termination points of the stride. Marker positions were utilized to estimate ankle, knee, and hip joint angles. The knee angle was determined as the relative angle between the thigh and shank segment using markers on the greater trochanter, lateral femoral condyle, and lateral malleolus of the ankle. The ankle angle was determined as the relative angle between the shank and foot segment using markers on tibial tuberosity, Achilles tendon insertion at the calcaneus, and $2^{nd}/3^{rd}$ metatarsals head. The hip angle was defined as the absolute angle with respect to the bottom vertical axis (270° in 2D cartesian coordinate system) originating from the greater trochanter (Robertson et al., 2013). Joint angles were calibrated using the measurements obtained from a quiet standing trial, which was performed before the walking trials. A 0° was considered as the neutral position for the ankle joint and hip joint, and the full extension for the knee joint. The data was smoothed with a second-order low-pass Butterworth filter with a cutoff frequency determined as six times the stride frequency (number of strides per second) of the corresponding stride cycle (Kirtley, 2006). For instance, the average stride frequency in this analysis was approximately 0.8, 1, and 1.1 strides per second for SLOW, CWS, and FAST, respectively, so the corresponding cutoff frequency was 4.8 Hz, 6 Hz, and 6.6 Hz. Each stride cycle was then interpolated from the original time domain to the 100% stride cycle with a cubic spline. Initial contact and maximal ankle dorsiflexion/plantarflexion, hip flexion/extension, and knee flexion/extension were extracted from the full stride cycle.

Inter-Joint Coordination

Ankle, knee, and hip joint angles in the sagittal plane were utilized to estimate inter-joint coordination. Three couples were selected: ankle-knee, knee-hip, and ankle-hip. Inter-joint coordination was calculated using the vector coding method (Sparrow et al., 1987). First, a phase plane, in which two joint angles were plotted against each other, was constructed. The distal joint was placed on the horizontal axis and the proximal joint was placed on the vertical axis. The coordination pattern was indicated by the coupling angle (unit: degree), which was calculated between the directional vector of the trajectory and the right horizontal axis in the counterclockwise direction for all time increments.

The difference of inter-joint coordination pattern between shod and barefoot conditions at each speed category was calculated with the cross-correlation coefficient (CCC) and root-mean-square-difference (RMSD). The CCC measures the difference in the spatiotemporal evolution of coordination patterns, whereas RMSD measures the magnitude differences between the coordination patterns(Chiu & Chou, 2012). A CCC value that is close to 1 and RMSD that is close to 0 suggests a less different coordination pattern between shod and barefoot walking. A more different coordination pattern is indicated when CCC is lower and RMSD is higher. The two-sided permutation test for symmetry was utilized for hypothesis testing ($\alpha = 0.05$). When a global effect of walking speed was detected, a follow-up pairwise permutation test was conducted with Bonferroni adjustment to identify effects between speed categories (i.e., SLOW vs. CWS, CWS vs. FAST).

To understand what specific coordination patterns were responsible for any change between the two footwear conditions due to walking speed, we further categorized the coordination pattern of the inter-joint couple(s) with a global effect of walking speed. The method of categorization was based on the work of Chang and colleagues (Chang et al., 2008). First, we averaged the coupling angle time series across all walking trials and all subjects in the same speed category and each footwear condition with a circular mean (Freedman Silvernail et al., 2018) and then categorized the coupling angle time series into four distinct regions: 1) in-phase, 2) anti-phase, 3) phase of the proximal joint and 4) phase of the distal joint. Taking the ankle-knee couple as an example, in-phase indicates that the ankle dorsiflexes while the knee flexes or the ankle plantarflexes while the knee extends; anti-phase indicates that the ankle dorsiflexes while the knee extends, or the ankle plantarflexes while the knee flexes; ankle phase indicates that the ankle rotates in the sagittal plane while the knee stays relatively motionless; knee phase indicates that the knee rotates while ankle stays relatively motionless. The detailed definition of each categorized region and the criteria for categorization have been described by Chang and colleagues (Chang et al., 2008) and other (Robertson et al., 2013).

Coordination Variability

To determine the coordination variability within each subject, we first established cycle-to-cycle variability for each time step across all stride cycles (for the same footwear condition and speed category). Because the coupling angle was circular, angular deviation (AD), which was equivalent to the standard deviation in linear statistics, was utilized to estimate coordination variability(Miller et al., 2010). The arithmetic mean was then used to average the AD data in early stance (initial heel strike + loading response: 1–12%), mid stance (12–31%), late stance (31–50%), and late swing phases (87–100%) of the stride cycle (Perry et al., 2010). We used the permutation test for symmetry to test the difference in mean between shod and barefoot walking in each walking speed category. For the comparisons that were significantly different, we determined Cohen's *d* effect size (ES) (small effect = 0.2; medium effect = 0.5; large effect = 0.8). Because AD data in this study was asymmetrically distributed (right-skewed), the data was log-transformed before the ES calculation (Botta-Dukát, 2018). No significant difference in CCC, RMSD, and AD was detected between the left and right sides based on the permutation test for symmetry, so the data were pooled, and the average value was used for hypothesis testing. All data analyses, including joint kinematics and vector coding, were conducted in MATLAB Version R2020a (The MathWorks Inc., Natick, MA, USA) with custom programs. Statistical analysis was performed in R (Version 3.6.0).

RESULTS

Walking Speed and Single Joint Kinematics

The average walking speed was 0.91 ± 0.18 m/s (shod) vs. 0.94 ± 0.17 m/s (barefoot) for SLOW, 1.35 ± 0.20 m/s (shod) vs. 1.38 ± 0.20 m/s (barefoot) for CWS, and

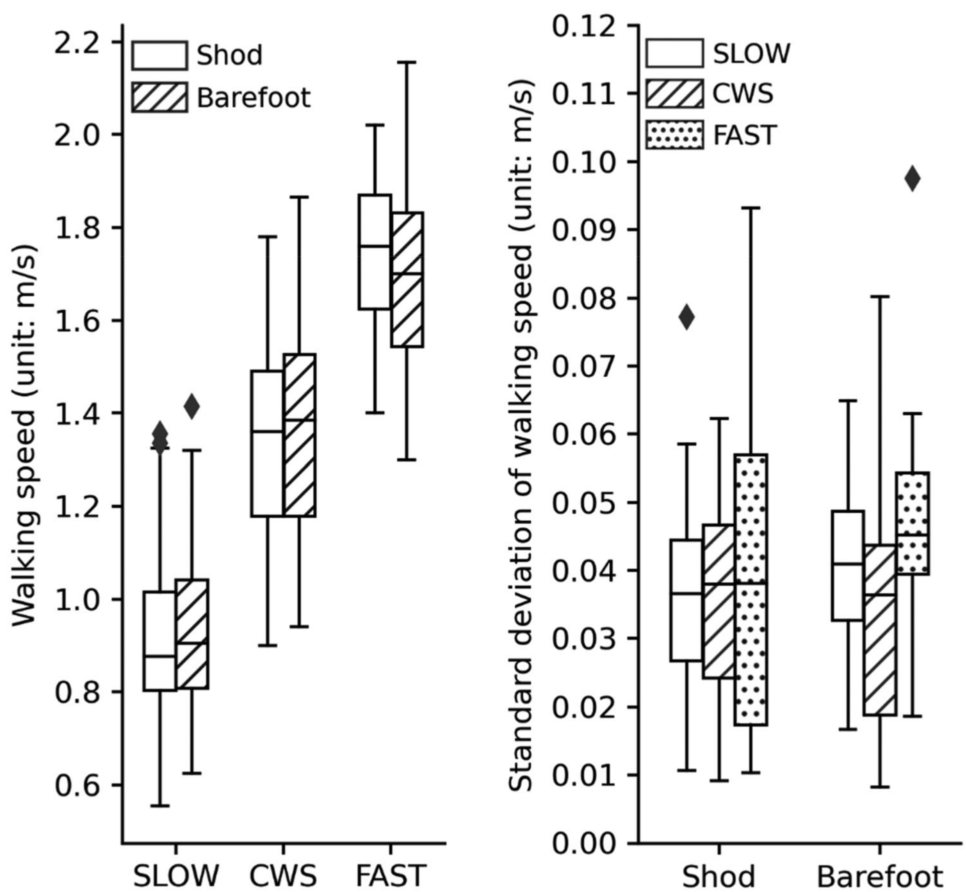

FIGURE 1. Summary of walking speed. Left: walking speed at each test condition; Right: standard deviation of walking speed at each test condition within the same subject; ◆: outlier.

TABLE 1. Initial contact and maximal value extracted from single joint kinematics.

	Shod			Barefoot		
	SLOW (0.91 m/s)[1]	CWS (1.35 m/s)[1]	FAST (1.74 m/s)[1]	SLOW (0.94 m/s)[1]	CWS (1.38 m/s)[1]	FAST (1.70 m/s)[1]
Init. cont.[2]						
Ankle	−0.8°	2.5°	7.4°	−9.7°	−7.9°	−5.8°
Knee	−0.6°	−0.7°	0.3°	0.1°	1.7°	3.4°
Hip	31.4°	37.5°	43.1°	32.4°	39.5°	44.1°
Maximum						
Dorsiflex.[3]	11.3°	11.1°	10.1°	9.1°	8.3°	8.1°
Plantarflex.[4]	−13.1°	−18.7°	−22.1°	−16.9°	−21.8°	−22.1°
Knee flex.[5]	52.4°	55.6°	55.9°	49.9°	52.9°	55.9°
Knee ext.[6]	−2.6°	−2.7°	−2.5°	−2.9°	−3°	−2.9°
Hip flex.[7]	19°	21.8°	23.8°	19.1°	21.7°	23.8°
Hip ext.[8]	−17.2°	−20°	−23.6°	−17.3°	−20.2°	−23.6°

[1]Mean walking speed at different test conditions. [2]Initial contact. [3]Dorsiflexion. [4]Plantarflexion. [5]Knee flexion. [6]Knee extension. [7]Hip flexion. [8]Hip extension. °: degree.

1.74 ± 0.17 m/s (shod) vs. 1.70 ± 0.20 m/s (barefoot) for FAST. The full description of walking speed at each test condition was summarized in Figure 1. All subjects attained significantly different speeds for each walking speed category ($p < 0.00001$). No significant difference between shod and barefoot walking speed was identified at any speed category nor were the walking speeds of left and right strides different (all p's > 0.05). Initial contact and maximum of ankle, knee, and hip angle were summarized in Table 1.

The Difference of Coordination Patterns between Shod and Barefoot

A global effect of walking speed on the difference of inter-joint coordination pattern between two footwear conditions was detected in the ankle-knee ($p < 0.00001$) and the ankle-hip ($p < 0.001$) but not in the knee-hip couple (Figure 2). We, therefore, categorized the coordination pattern of ankle-knee (Figure 3) and ankle-hip (Figure 4). Knee-hip coordination plot is not shown, as there was no global effect of walking speed on knee-hip coordination. For the ankle-knee couple, pairwise comparison indicated that a significant effect was detected in the comparison between SLOW and CWS (CCC & RMSD: $p < 0.001$) and between CWS and FAST (CCC & RMSD: $p < 0.001$). For the ankle-hip couple, the pairwise comparison indicated that a significant effect was detected in the comparison between SLOW and CWS (CCC & RMSD: $p < 0.001$) but not in the comparison between CWS and FAST.

Coordination Variability

For the ankle-hip couple, AD in the late swing phase at SLOW ($p < 0.005$; ES: 0.67) and CWS ($p < 0.05$; ES: 0.74) was significantly lower in the barefoot than in the shod condition; for the knee-hip couple, the AD in the mid stance phase during CWS ($p < 0.05$; ES: 0.64) and FAST ($p < 0.05$; ES: 0.61) was significantly lower in the barefoot than in the shod. No significant differences were identified in other comparisons (Table 2).

DISCUSSION

This study examined the pattern and variability of inter-joint coordination of barefoot and shod walking at different walking speeds. We hypothesized that walking speed would change the difference of coordination patterns between barefoot and shod walking. We found that walking speed significantly affected the coordination pattern of ankle-knee and ankle-hip but not knee-hip. Specifically, ankle-knee coordination patterns between barefoot and shod conditions became less different as walking speed increased (CCC was higher and RMSD was lower as walking speed increased); ankle-hip coordination patterns between barefoot and shod conditions, on the other hand, became more different as walking speed increased (CCC was lower and RMSD was higher as walking speed increased) (Figure 2).

Our report is the first, to the best of our knowledge, to identify this phenomenon (i.e., as walking speed increases, ankle-knee pattern becomes less different and ankle-hip pattern becomes more different between shod and barefoot walking). Our result suggests that barefoot

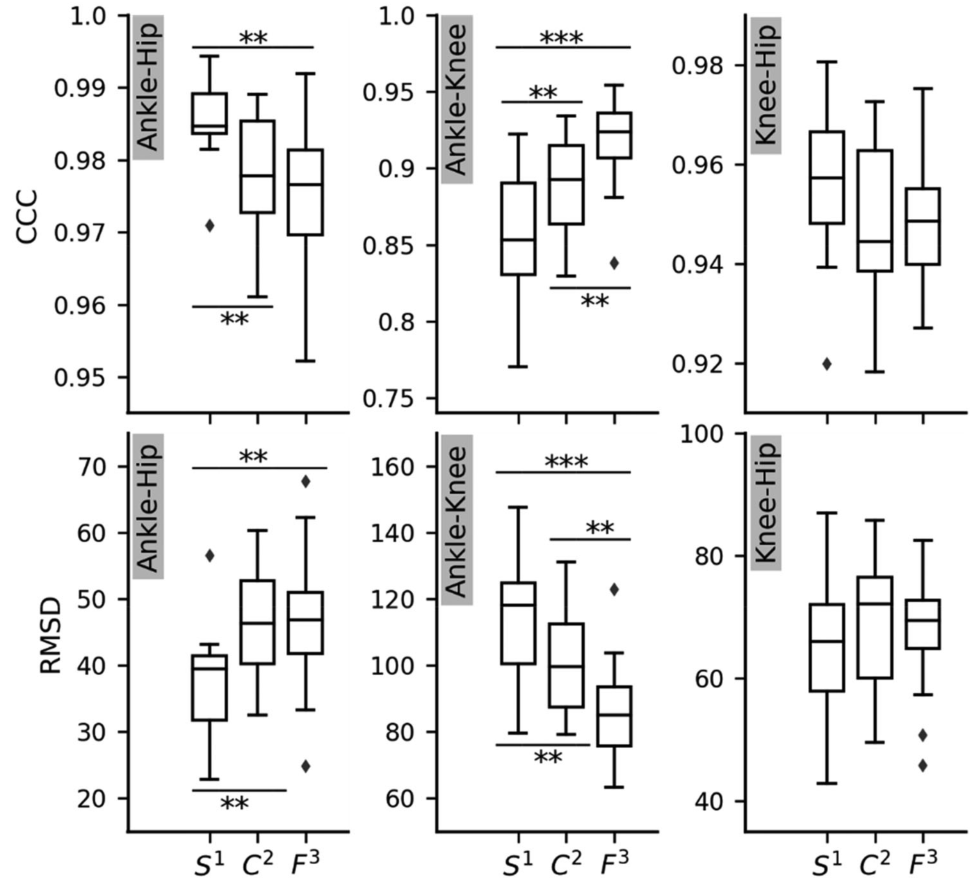

FIGURE 2. differences of coordination pattern in ankle-hip, ankle-knee, and knee-hip at three speed categories. [1]S: SLOW; [2]C:CWS; [3]F: FAST. The top row is the plots of CCC, the bottom row is the plots of RMSD. ***: $p < 0.00001$; **: $p < 0.001$; ◆: outlier.

walking and shod walking adopt different coordination strategies to cope with the increased walking speed (Figure 2). For ankle-hip couple, we found that barefoot condition was primarily in the ankle phase region during the late swing phase(≈ 87–100% stride cycle) (Figure 4), indicating that ankle motion was more dominant in the late swing, regardless of walking speed; the shod condition, in contrast, clearly demonstrated a pattern where the coordination extended more phase regions as walking speed increased, with no single-phase region being dominant (Figure 4). Ankle-knee coordination plot (Figure 3) demonstrated a similar pattern, though less noticeable than ankle-hip coordination, in the late swing phase of the stride cycle.

Wallace et al. (2018) found barefoot walking had a significantly smaller impact force than walking in sandals and suggested that the attenuation of impact force was initiated by better exteroception provided by bare feet. Using continuous relative phase (CRP), Kurz and Stergiou (2004) estimated the foot-shank coordination pattern in barefoot running and found that the shank-foot coordination in stance phase was more out-of-phase in barefoot compared to shod. They (Kurz & Stergiou, 2004) suggested that the more out-of-phase coordination pattern in shank-foot modulated the impact force based on the perception of impact, which may be obtained through the mechanoreceptor of the foot given that they contribute to motor planning and movement control (Ackerley & Kavounoudias, 2015; Kennedy & Inglis, 2002; Lane et al., 2019). Experimental studies showed that people had better sensory function when barefoot (Robbins et al., 1988; 1995). Individuals may adopt other movement strategies without the cushioning of athletic shoes to help absorb impact force at heel strike. The plantar fat pad is the natural shock absorber that dissipates the impact stress generated during walking (Campanelli et al., 2011). The ankle joint is more plantarflexed in the initial heel strike of barefoot walking than shod walking at CWS (Chard et al., 2013; Oeffinger et al., 1999). Our study confirmed this

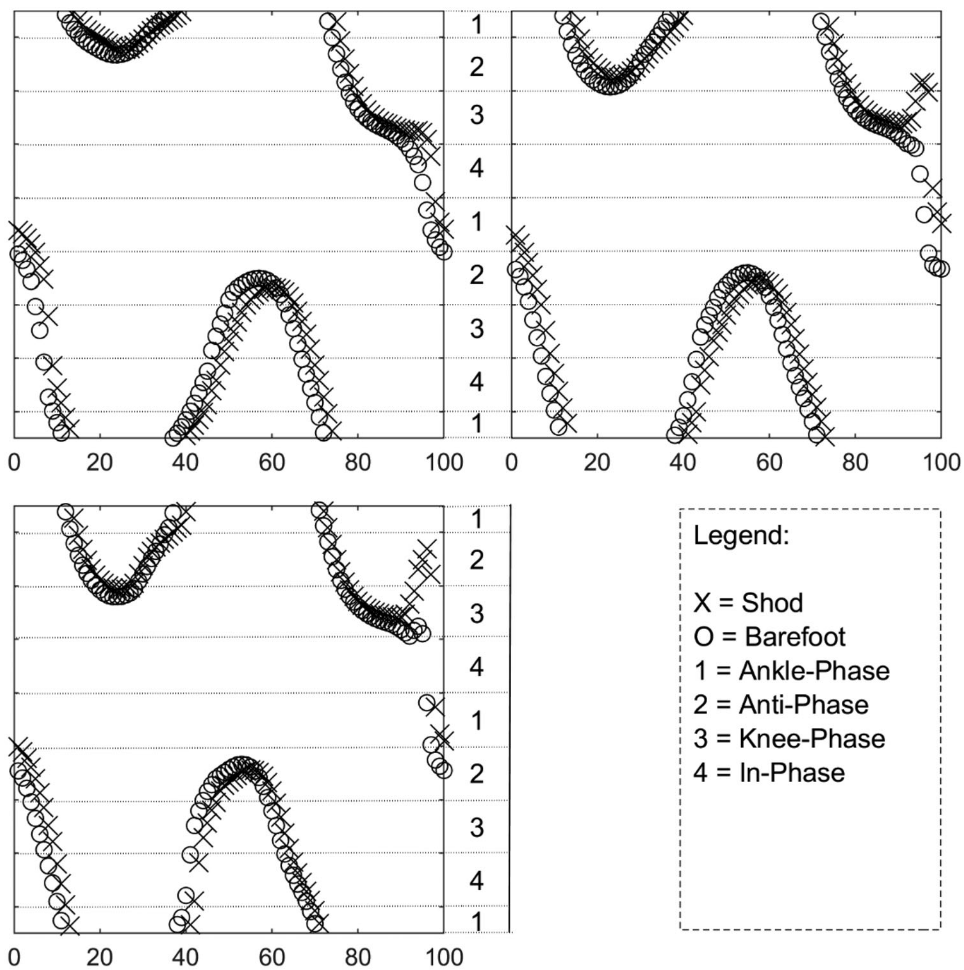

FIGURE 3. Coordination pattern of ankle-knee. upper left: SLOW; upper right: CWS; lower left: FAST. Horizontal axis is stride cycle (unit: %). The numerical labels along the vertical axis are the categories of coordination patterns that are separated by the dotted line from each other.

observation and further extended this observation to SLOW and FAST walking speeds (Table 1). A more plantarflexed position at the initial heel strike is expected to increase the contact surface area and thus decrease the stress. In addition, our study also found that coordination variability of ankle-hip in late swing at SLOW and CWS was significantly lower in barefoot with medium effect. Coordination variability of ankle-hip in late swing at FAST was also lower in barefoot than shod; however, no significant difference was identified. Overall, the result of our study strongly suggests that the ankle-dominant coordination pattern in ankle-hip of barefoot during the late swing and the reduction of coordination variability in barefoot may be the consequence of intentional muscular control to prepare for the anticipated impact at the forthcoming heel strike.

The second hypothesis of this study was that coordination variability estimated in early stance, mid stance, late stance, and late swing phase was significantly different between barefoot and shod condition in the three speed categories. In addition to the lower coordination variability of the ankle-hip couple in barefoot at the late swing phase, we also found a significantly lower coordination variability of the knee-hip couple in barefoot during mid stance at both CWS and FAST. No significant differences were identified in other comparisons. Romer et al. (2019) found that barefoot walking significantly lowered the segmental coordination variability of the shank-foot couple but increased the coordination variability of the thigh-shank couple at CWS. The coordination variability in Romer et al.'s study (2019) was estimated across the entire strike cycle, which may not be an

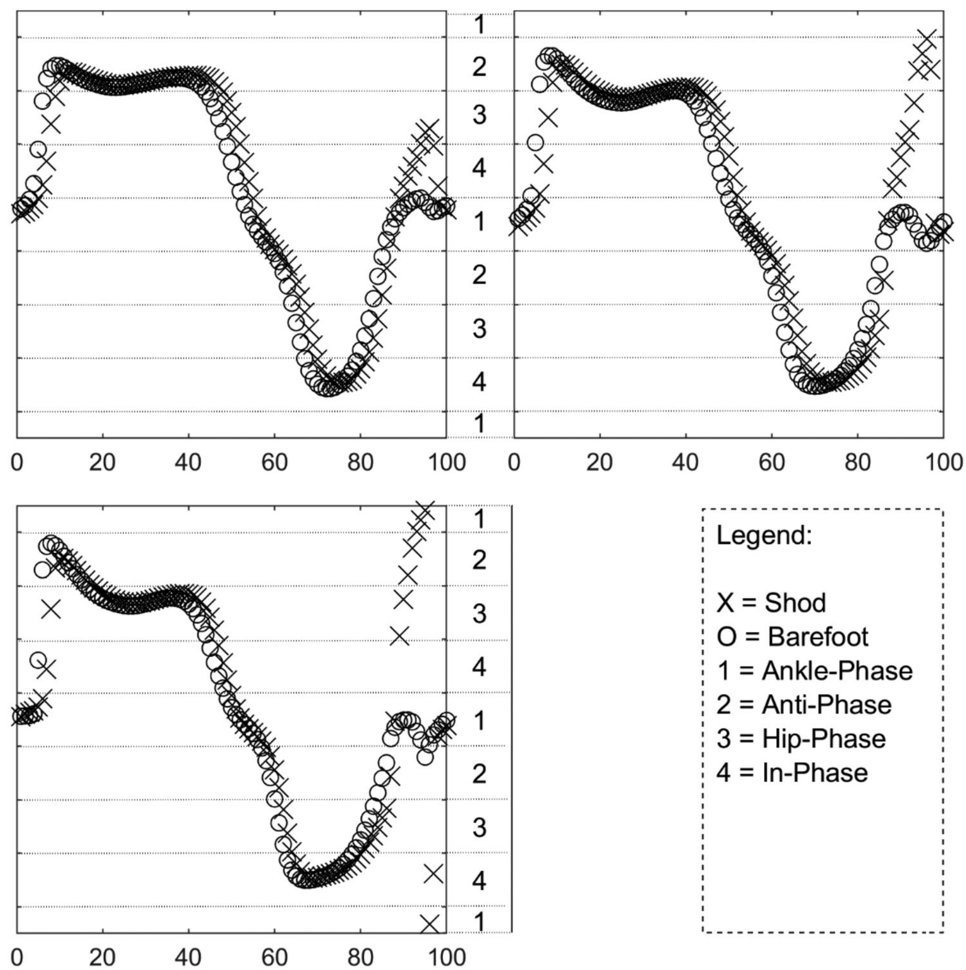

FIGURE 4. Coordination pattern of ankle-hip. upper left: SLOW; upper right: CWS; lower left: FAST. Horizontal axis is stride cycle (unit: %). The numeurcal labels along the vertical axis are the categories of coordination patterns that are separated by the dotted line from each other.

appropriate approach as coordination variability was likely to change across the stride cycle (Robertson et al., 2013). The conflict between the result of our study and that of Romer et al. (2019) may derive from methodological difference.

The reduction of coordination variability of knee-hip during mid stance deserves more attention, as mid stance is a weight-bearing phase and may carry more implications for future study. Khoury-Mireb et al. (2019) examined the gait variability when subjects wore unstable shoe designs, which were utilized to strengthen neuromuscular control. They (Khoury-Mireb et al., 2019) found that the variability of ankle moment was significantly decreased in unstable footwear and suggested that the decrease of variability was likely due to the compensatory strategy to control the dynamic stability of movement. Therefore, the decreased coordination variability in our study may be due to the same compensatory mechanism to control movement stability. Although variability and stability are two coupled phenomenon of human gait, the variability of movement is not entirely equivalent to the stability of movement (Granata & England, 2007). Stergiou and Decker (2011) claimed that the gait stability could only be estimated using nonlinear metrics, such as the Largest Lyapunov Exponent or entropy. The stability of movement is significantly lower in barefoot running than shod running both in the short-term (Ekizos et al., 2017) and long-run (Hollander et al., 2021) as estimated by the Largest Lyapunov Exponent. The change of stability, however, was not identified in barefoot walking (Hollander et al., 2021). These conflicting results indicates that the decrease of coordination variability in our study should not be interpreted as a change of stability during barefoot walking. In orthopedic

TABLE 2. Coordination variability for each inter-joint couple at different speed category.

	SLOW		CWS		FAST	
	Shod (0.91 m/s)[2]	Bf[1] (0.94 m/s)[2]	Shod (1.35 m/s)[2]	Bf [1] (1.38 m/s)[2]	Shod (1.74 m/s)[2]	Bf [1] (1.70 m/s)[2]
Ankle - Knee						
Early Stance	11.6	12.7	10.5	8.7	8.82	7.1
Mid Stance	13.0	13.2	11.7	11.0	11.6	10.0
Late Stance	15.5	16.3	13.6	16.2	16.7	17.4
Late Swing	17.3	16.9	15.2	15.2	14.9	17.0
Knee - Hip						
Early Stance	14.0	13.8	14.5	10.7	11.6	9.1
Mid Stance	8.5	7.8	**6.86**	**5.12***	**5.64**	**4.26***
Late Stance	10.6	11.1	7.5	8.9	8.1	7.1
Late Swing	17.8	19.7	15.5	16.6	15.8	17.7
Ankle - Hip						
Early Stance	11.2	12.4	13.5	12.7	10.7	11.8
Mid Stance	5.15	5.07	4.41	3.96	3.96	3.53
Late Stance	8.26	9.58	7.04	8.13	7.4	7.87
Late Swing	**24.1**	**18.1***	**26.5**	**20.2***	25.5	21.6

[1]Bf: Barefoot. [2]Mean walking speed at different test conditions. *Significant difference between shod and barefoot.

biomechanics, the reduction of coordination variability was considered an unhealthy state related to overuse injury (Hamill et al., 2012). We think that the decreased coordination variability in the mid stance observed in our study was unlikely to be the result of orthopedic conditions, as our subjects were healthy and free from medical conditions that influenced gait at the time of data collection. The reduction of coordination variability was theorized to increase the risk of joint wear and tear in the long term, as highly repetitive movement patterns can induce higher stress in the joint (Kumar et al., 2017). One alternative interpretation of the reduced coordination variability in our study, thus, is that humans accommodate developmentally to wearing shoes, and walking barefoot is, therefore, no longer necessarily "natural" for all people as has been suggested by some (Lieberman, 2012; 2013; Sichting et al., 2020). More research is necessary to understand the relation between the decreased coordination variability and joint stress during barefoot walking.

One of the challenges in this study was how to control the walking speed. A treadmill is a viable option, yet the treadmill can interfere with gait variability (Hollman et al., 2016). One method to control overground walking speed is metronome cueing, where a subject walks overground in a step rate deliberately matching with a predetermined beat. Metronome cueing, however, could decrease the gait variability (Wright et al., 2016). Self-selected walking speed has also been utilized in some gait studies (Chiu et al., 2010; Chiu & Chou, 2012; Hutin et al., 2012; Wang et al., 2021), in which subjects have maximum freedom to select their walking speed. Verbal instruction are commonly utilized in clinical practice of gait training and gait study to control walking speed (Lehman et al., 2005). Our study used verbal instruction to simulate three scenarios in daily life, allowing subjects to self-select their walking speed under each speed condition but also maintaining some control. We found no significant difference in walking speed between barefoot and shod within the same speed category, so the difference of coordination pattern between shod and barefoot within the same speed category is not affected by the walking speed. Also, all subjects walked at significantly different speeds between speed categories as instructed and walking speed variation at each test condition within the same subject is low (Figure 1). Overall, our method of using verbal instruction to control walking speed was successful.

There are several limitations to this study. First, although we purposely controlled the subjects to be female, young, and healthy, as gender (Boyer et al., 2017), age (Callisaya et al., 2010; Chiu & Chou, 2012; Skiadopoulos et al., 2020), and medical condition (Chiu et al., 2010; Moon et al., 2016) can influence coordination pattern and variability of gait, this decision limits the generalizability of this study. Future work should determine if the same effect of walking speed is observed in other populations. Second, although we identified interesting patterns in gait coordination, some of which have not been reported previously, our ability to understand these results was limited due to the lack of other measurements such as EMG or energy consumption. Further analysis is required to understand the relationship of gait coordination to other parameters such as anthropometrics, muscle activation, or energy expenditure.

CONCLUSION

Walking speed influences the difference in the ankle-knee and ankle-hip couples between barefoot and shod walking. As walking speed increases, the ankle-knee coordination pattern becomes less different, but the ankle-hip coordination pattern becomes more different. Compared to shod, barefoot walking has decreased coordination variability in mid stance of knee-hip at CWS and FAST speed and in the late swing of the ankle-hip at SLOW and CWS speed. Future research should investigate the connection between the decreased coordination variability and joint tissue stress to understand the impact of barefoot walking on the lower extremity joints.

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