Gait & Posture 70 (2019) 317–322

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Full length article

Sex differences in lower extremity coordinative variability during running

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ARTICLE INFO

Keywords:
Continuous relative phase
Injury implications
Hip
Knee

ABSTRACT

Background: Differences in coordinative variability have been previously reported between healthy and injured runners. Many running-related injuries have a sex bias, particularly patellofemoral pain (PFP), as female runners are approximately twice as likely to develop PFP compared to males. However, very little is currently known regarding sex differences in coordinative variability during running.

Research Question: Are there sex differences in continuous relative phase (CRP) variability for pelvis-thigh and thigh-shank couplings during the stance phase of running?

Methods: Pelvis, thigh, and shank segment kinematics were collected on 15 female and 15 male subjects during overground running at a self-selected easy pace (2.39–3.56 m/s) using a 10-camera 3D motion capture system. Continuous relative phase (CRP) variability was calculated between the pelvis-thigh and thigh-shank, and averaged during four distinct stance sub-phases. A mixed effects linear model compared CRP variability between sexes at each stance sub-phase.

Results: Compared to males, females displayed significantly lower pelvis-thigh CRP variability in the transverse plane during the loading response phase, and significantly lower thigh-shank CRP variability in the sagittal plane during the loading response and pre-swing phases.

Significance: Lower coordinative variability in females during the loading response for two couplings may provide additional insight into the sex bias for developing certain running-related injuries. However, any injury implications from these results are speculative and should be interpreted with caution.

1. Introduction

Patellofemoral pain (PFP) is the most common running-related injury [1], and is approximately twice as likely to occur in females compared to males [1,2]. Several individual risk factors have been proposed which may help explain the greater incidence in females, including local structural factors [3], low hip strength [3–9] and excessive peak hip adduction [8,10–13] and peak hip internal rotation [8,11,11,12,13] during running. However, most biomechanical studies have focused on variables at discrete time points, ignoring data from most of the gait cycle. Thus, recent studies have begun investigating overuse injuries using a dynamical systems approach, which focuses on coordinative variability between coupled segments or joints rather than quantifying discrete biomechanical variables [14,15].

Several studies have compared coordinative variability between healthy and injured athletes using methods such as continuous relative phase (CRP) and vector coding [16]. Many of these studies have found differences between healthy and injured athletes, however, results have not been consistent. Compared to healthy controls, some studies have found that injured participants displayed lower coordinative variability [15,17–19], while other studies have found injured participants displayed greater coordinative variability [18–24], or no differences in coordinative variability [25–27]. These differences do not appear to be explained by injury region or analysis type [16]. It may be possible that too much or too little coordinative variability are both problematic [14,16], but the exact ranges for ideal variability are still unknown.

Despite the sex bias for PFP, only one study to date has examined sex differences in coordinative variability during running [28]. Because the exact reason for the sex bias in knee injuries is still not well understood [1,29], investigating sex differences in coordinative variability during running could provide additional insight. Based on previous literature linking discrete biomechanical variables to either injury status or increased patellofemoral joint stress, specific couplings of interest include pelvis-thigh coordination in the frontal and transverse planes [8,10–13], as well as thigh-shank coordination in the sagittal [30,31] and frontal planes [32]. In addition, coordination timing has

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https://doi.org/10.1016/j.gaitpost.2019.03.024
Received 14 September 2018; Received in revised form 14 March 2019; Accepted 25 March 2019
0966-6362/

previously been suggested as an important consideration in studying knee joint injuries [33,34], so further comparing coordinative variability at distinct stance sub-phases could provide additional insights into whether any sex-related differences have injury implications.

Therefore, the purpose of this study was to compare coordinative variability for pelvis-thigh (frontal and transverse planes) and thigh-shank couplings (sagittal and frontal planes) during the stance phase of running. It was hypothesized that females would demonstrate less pelvis-thigh frontal plane and transverse plane coordinative variability compared to males, but that there would be no differences in coordinative variability for the other couplings.

2. Methods

2.1. Subjects

All subjects signed an informed consent form approved by the Institutional Review Board of the University of Oregon prior to participating. To be included in the study, subjects needed to be between 18–45 years old [8,11], average running at least 20 miles per week [35], and report no major injuries for at least the previous 6 months [36,37]. A major injury was consistent with the consensus injury definition by Yamato et al. [38] as pain that forces a runner to restrict or stop running for at least seven days, three consecutive training sessions, or requires consultation with a physician. Thirty subjects (15 males and 15 females) were included in this study (Table 1).

2.2. Data collection

Subjects wore their normal training shoes during this study so that new footwear would not alter their normal running gait [39]. However, subjects could not wear minimalist or maximalist shoes. Subjects warmed up at their easy pace on a treadmill for five minutes and were outfitted with 39 reflective markers [40]. The pelvis was defined by two markers on the anterior superior iliac spines, and one at the midpoint between the posterior superior iliac spines. The hip joint center was defined based on anthropometric measurements of ASIS breadth [41]. The femur was defined by two markers at the medial and lateral femoral epicondyles, and one marker mid-thigh in line with the lateral epicondyle and greater trochanter. The shank was defined by two markers at the medial and lateral malleoli, and one marker on the medial shank. All markers except for medial knee and ankle markers were used as tracking markers. This marker set has been shown to be reliable for gait data compared to a cluster-based tracking system [42].

Subjects were given two laps to become accustomed to level-ground running in the lab prior to data collection, and then ran 20 continuous laps of approximately 40-meters in the laboratory at their easy run pace. This pace was measured using the average velocity of the sacral marker within a 10-meter region of the capture volume during each trial, and defined as the pace a subject could comfortably maintain for 30 min while maintaining a conversation. Once each participant's easy pace was established using the last practice lap, only trials within ± 5% of this pace were eligible for inclusion in the analysis. Participants were instructed not to alter their stride to hit one of three force plates (AMTI, Watertown, MA) located in series in this region. Segment kinematics were collected when the participants passed through a straight 10-meter region in the center of the capture volume with a 10-camera motion capture system (Motion Analysis Corp., Santa Rosa CA) sampling at 200 Hz.

Table 1
Subject demographics for females and males.

	Females	Males	*p*-value
Age (years)	27.5 ± 5.8	26.3 ± 8.3	0.648
Height (cm)	166.3 ± 4.3	177.8 ± 7.8	< 0.001
Weight (kg)	59.1 ± 6.0	72.0 ± 11.4	< 0.001
Weekly Mileage (miles)	27.7 ± 9.6	37.0 ± 15.8	0.061
Running Experience (years)	13.4 ± 5.7	11.5 ± 6.5	0.404
Running Speed (m/s)	2.9 ± 0.3	3.1 ± 0.4	0.055

Note: Values represent mean ± standard deviation.

2.3. Data processing

Marker trajectories were identified using Cortex 5.0 motion capture software (Motion Analysis Corp., Santa Rosa, CA) and were smoothed using a low-pass, fourth-order, zero-lag Butterworth filter with an 8 Hz cutoff. Stance phase was defined as the first frame the vertical ground reaction force was greater than or equal to 50 Newtons until the first frame the vertical ground reaction force was less than 50 Newtons [43]. The first five trials per limb where the subject successfully hit the force plate were selected for each subject.

Continuous relative phase (CRP) was selected as the metric of coordination for this study because it incorporates angular velocity in addition to angular position, which may add additional information in regard to injury risk [34]. Segment angles and velocities during stance phase were calculated with respect to a global coordinate system using a custom LabView program (National Instruments, Austin, TX). These values were interpolated to 100% of stance phase and normalized to values between -1 and 1 [44] using the following formulas [15]:

$$\theta_i = \frac{2 * [\theta_i - \mathbf{min}(\theta_i)]}{\mathbf{max}(\theta_i) - \mathbf{min}(\theta_i)} \tag{1}$$

$$\omega_i = \frac{\omega_i}{\mathbf{max}\{\mathbf{max}(\omega_i), \mathbf{max}(-\omega_i)\}} \tag{2}$$

Phase portraits were constructed from these normalized segment angles (θ) (x-axis) and velocities (ω) (y-axis) so that phase angles (φ) could be calculated using the following formula:

$$\varphi = \mathbf{tan}^{-1}(\omega/\theta) \tag{3}$$

CRP was calculated by subtracting the phase angle of the distal segment from the proximal segment for each coupling [15]. All trials for each subject were averaged to create an ensemble CRP curve for each plane of motion. Point-by-point CRP variability was calculated by averaging the standard deviation at each percent of stance phase, for a total of 101 data points [15].

To compare data between sexes, ensemble CRP variability curves were calculated for females and males, respectively. Average CRP variability was calculated for the entire stance phase (all 101 data points) in addition to four distinct stance sub-phases: 0–20% (loading response), 20–50% (midstance), 50–80% (terminal stance), and 80–100% (pre-swing) using the same methods as Hein et al. (2012) [27].

In order to include data from all 60 limbs while accounting for paired limb dependency, data were analyzed by a mixed effects linear model using the methodology of Stewart et al. (2018) [45]. Specifically, in our model, the effect of sex was included as a fixed factor, while dependency between limbs was accounted by inputting limb side (right or left) as a random factor. While running speed was not significantly different between sexes ($p = .055$), it was added to the statistical model (limb-specific) as a covariate to ensure any differences in CRP variability observed were not a product of running speed. The alpha-level for each comparison was set to .05, and effect sizes were calculated using Cohen's *d*. All statistics were calculated using SPSS version 25 (SPSS Inc., Chicago IL), except for effect sizes, which were calculated using G*Power 3.1 (G*Power, Düsseldorf, Germany).

3. Results

No significant differences were observed for CRP variability between the pelvis and thigh in the frontal plane, $p > 0.05$ for all

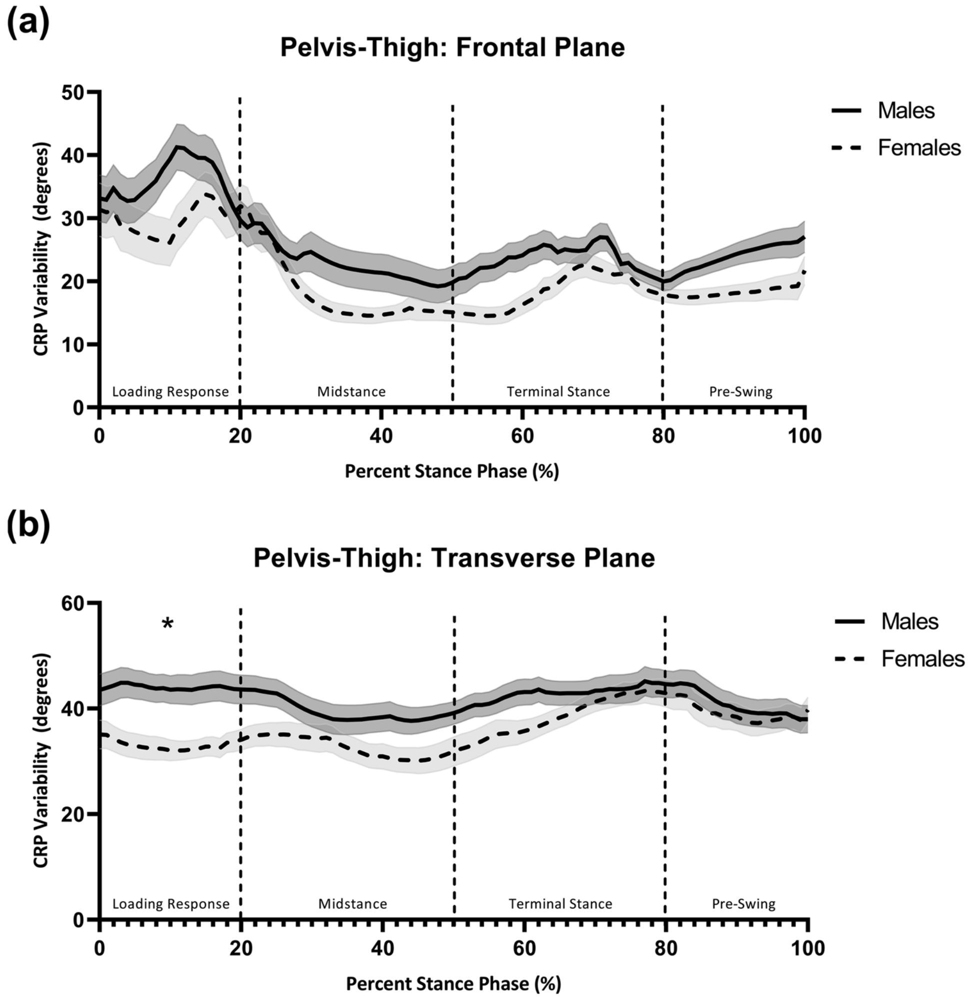

Fig. 1. CRP variability during stance phase between the pelvis and thigh for males and females in the (a) frontal plane and (b) transverse plane. Shaded areas represent standard error, while * indicates a significant difference between sexes.

Table 2
Sex differences in CRP variability for pelvis-thigh couplings.

	Male	Female	*p*-value	Effect Size (*d*)
Pelvis-Thigh Frontal				
Entire Stance Phase	26.0 ± 11.3	20.5 ± 8.3	.083	.555
Loading Response	35.9 ± 21.1	29.8 ± 23.4	.291	.274
Midstance	22.8 ± 19.1	18.1 ± 9.4	.308	.312
Terminal Stance	23.5 ± 12.0	18.4 ± 9.0	.120	.481
Pre-Swing	24.0 ± 13.7	18.2 ± 9.2	.122	.497
Pelvis-Thigh Transverse				
Entire Stance Phase	41.7 ± 13.6	35.8 ± 11.4	.074	.470
Loading Response	44.0 ± 20.3	33.1 ± 14.5	.028	.618
Midstance	39.4 ± 18.0	32.7 ± 16.4	.133	.389
Terminal Stance	42.9 ± 16.7	38.6 ± 14.8	.294	.273
Pre-Swing	40.7 ± 19.3	39.0 ± 18.1	.757	.091

Note: Values represent mean ± standard deviation. All units are in degrees.

comparisons (Fig. 1, Table 2). However, while statistically insignificant, we did observe a moderate effect size comparing CRP variability between sexes when averaged across the entire stance phase (males: 26.0 ± 11.3; females: 20.5 ± 8.3; p = 0.083; d = 0.555). In the transverse plane, significantly less CRP variability between the pelvis and thigh was observed for females during the loading response, with a moderate effect size (males: 44.0 ± 20.3°; females: 33.1 ± 14.5°; p = .028, d = 0.618) (Fig. 1, Table 2).

For CRP variability between the thigh and shank in the sagittal plane, females displayed significantly less variability compared to males during the loading response (males: 12.2 ± 5.2°; females: 8.4 ± 4.7°; p = .006; d = 0.767) and pre-swing phases (males: 13.5 ± 6.2°; females: 10.5 ± 4.1°; p = .032; d = 0.571) with moderate effect sizes (Fig. 2, Table 3). No significant differences between sexes were detected for thigh-shank frontal plane CRP variability for all stance sub-phases (Fig. 2, Table 3).

4. Discussion

The purpose of this study was to examine sex differences in CRP variability between the pelvis-thigh and thigh-shank during running. Compared to males, pelvis-thigh CRP variability was lower in females during the loading response in the transverse plane. Thigh-shank sagittal plane CRP variability was also lower in females during the loading response and pre-swing phases. These results partially supported our hypothesis.

Transverse plane pelvis-thigh CRP variability was significantly lower in females during the loading response (Fig. 1, Table 2), when the body must decelerate the limb and absorb the forces of impact [46]. Previous research has demonstrated that females with PFP run with greater hip internal rotation than healthy controls [8,11–13], which may also increase patellofemoral contact pressure [47]. According to the hypothesis by Hamill et al. [14], low transverse plane CRP variability during the loading response could exacerbate patellofemoral

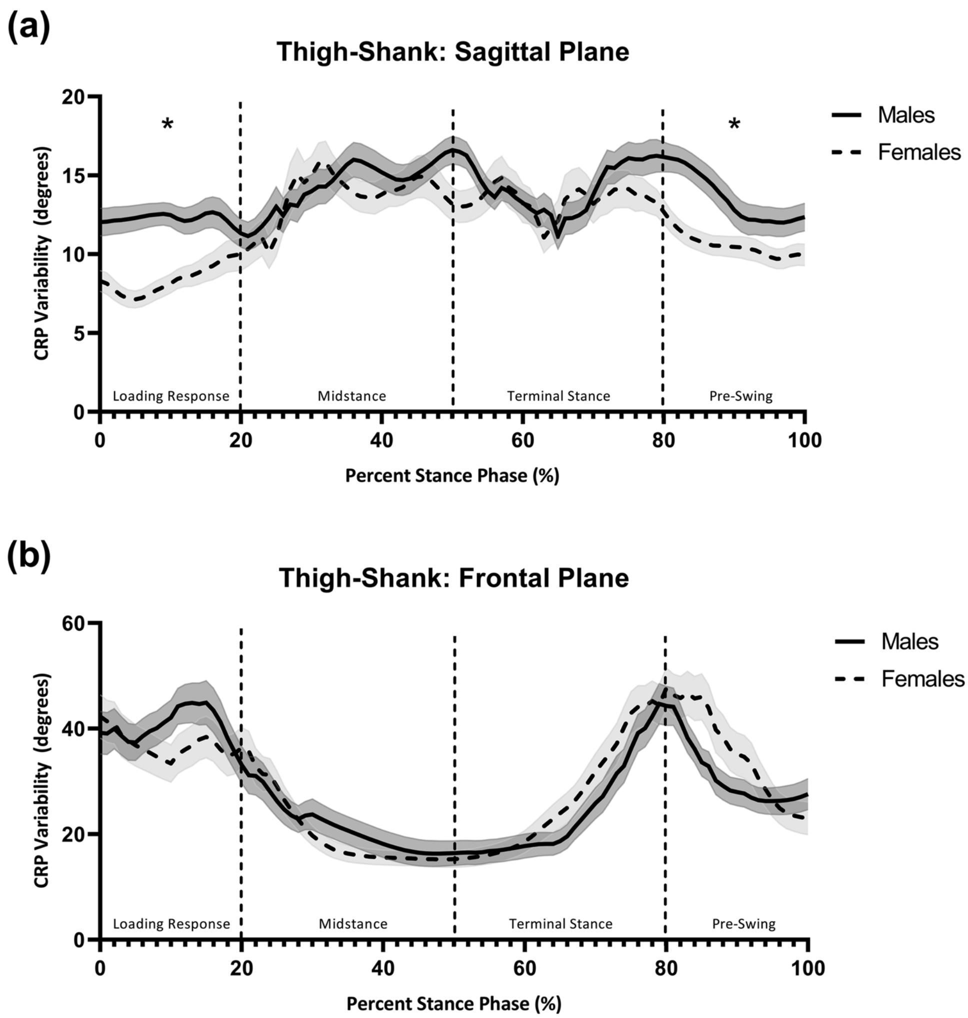

Fig. 2. CRP variability during stance phase between the thigh and shank for males and females in the (a) sagittal plane and (b) frontal plane. Shaded areas represent standard error, while * indicates a significant difference between sexes.

Table 3
Sex differences in CRP variability for thigh-shank couplings.

	Male	Female	*p*-value	Effect Size (*d*)
Thigh-Shank Sagittal				
Entire Stance Phase	13.7 ± 3.8	11.9 ± 5.1	.115	.400
Loading Response	12.2 ± 5.2	8.4 ± 4.7	.006	.767
Midstance	14.4 ± 5.9	13.7 ± 7.4	.678	.105
Terminal Stance	14.3 ± 4.9	13.4 ± 7.5	.591	.142
Pre-Swing	13.5 ± 6.2	10.5 ± 4.1	.032	.571
Thigh-Shank Frontal				
Entire Stance Phase	28.1 ± 12.9	28.7 ± 13.4	.861	.046
Loading Response	40.4 ± 21.1	37.0 ± 24.0	.587	.150
Midstance	21.2 ± 17.6	19.8 ± 10.9	.750	.096
Terminal Stance	24.7 ± 15.2	27.7 ± 18.1	.498	.180
Pre-Swing	30.5 ± 18.0	34.8 ± 25.0	.444	.197

Note: Values represent mean ± standard deviation. All units are in degrees.

contact pressure by further decreasing the surface area available to distribute the high internal forces during this phase of running [14]. While speculative, this finding has potential to further explain the sex bias for PFP.

In this study, females displayed significantly lower sagittal plane CRP variability between the thigh and shank compared to males during the loading response and pre-swing phases (Fig. 2, Table 3). While the peak knee flexion angle is largely responsible for peak patellofemoral joint force [30,48], this angle generally occurs during midstance, where CRP variability was similar between sexes. Low CRP variability may indicate that loading is distributed across a relatively small surface area at the joint [14]. Therefore, low thigh-shank sagittal plane CRP variability in females could indicate decreased contact area, and thus increased contact stress, at the knee during these two phases. However, this is speculative, and more research is needed to determine whether differences in sagittal plane thigh-shank CRP variability have injury implications for females.

While not reaching statistical significance ($p = 0.083$), a moderate effect size was observed ($d = 0.555$) between sexes for frontal plane pelvis-thigh CRP variability, with females displaying less CRP variability than males for the entire stance phase. Previous research suggests that healthy females run with greater peak hip adduction during running compared to healthy males [12], which may decrease contact area and increase contact stress on the lateral patella [49] as well as increase the external knee abduction moment [50]. Low frontal plane CRP variability for this coupling could also exacerbate lateral patellar contact stress and further decrease the surface area available to distribute internal forces, similar to the transverse plane [14]. While this may potentially place females at an even greater risk of developing lateral knee pain during running, the results from this study cannot confirm this hypothesis. However, this finding is worth further investigation due to the observed effect size.

To our knowledge, only one other study has investigated sex differences in coordinative variability during running, as Boyer et al. (2017) compared three different couplings between older and younger males and females using a vector coding approach [28]. The only coupling in common between these studies was between the thigh and shank in the transverse plane, with Boyer et al. finding that younger females displayed more coordinative variability in this coupling. This difference between studies may be due to methodological approaches in quantifying coordinative variability, or differences in participant's age.

Lower coordinative variability in females has been observed previously for a dynamic task. During an unanticipated cutting maneuver, Pollard et al. [51] found that females displayed less coordinative variability than males, specifically between the thigh-shank and hip-knee, using a vector coding approach. These authors hypothesized that lower coordinative variability during cutting in females may underlie the sex bias for sustaining a non-contact ACL injury. This previous study supports the notion that low coordinative variability in females may partially underlie the sex bias for sustaining an injury to the knee.

In addition to possible injury implications, this study demonstrated that males and females do not display the same CRP variability for certain sub-phase couplings. Males and females also display different joint kinematic patterns while healthy [12,52] and while suffering from PFP [32], as well as differences in muscle activation patterns at the hip [52]. Altogether, these findings suggest that males and females should generally be analyzed separately in running biomechanics and injury research.

Many studies have reported greater movement variability in an injured sample compared to healthy controls [18–24]. However, this trend was not consistent across subcategories of injury type, body region, or task performed, with several studies conversely finding lower coordinative variability in the injured athletes compared to their healthy counterparts [15,17–19], or no differences in coordinative variability [25–27]. Most studies comparing coordinative variability between injured and healthy cohorts are cross-sectional in nature, meaning that we cannot determine whether any differences observed were the cause, or result, of injury. There is clearly more work that needs to be done to explain how movement variability may be related to injury, particularly conducting prospective studies. Given the collective results of all previous work, any implications for injury from the sex differences found in this study are purely speculative and should be viewed cautiously.

This study is not without limitations. First, we only investigated coordination within a single plane, and did not look at coordinative variability between different planes of motion (i.e., between transverse plane pelvic motion and frontal plane thigh motion). Previous studies have investigated coordination between planes in adjacent segments for distal couplings, and found differences in coordinative variability between injured and healthy runners [15]. Thus, sex differences in between-plane coordinative variability of adjacent segments should be investigated for future studies. Also, foot strike patterns were not controlled in this study, so it is not clear if coordinative variability is affected by the type of foot strike. Running shoes were also not standardized in this study, which was chosen so that coordinative variability was not affected by introducing novel footwear [39].

In conclusion, females demonstrated lower CRP variability between the pelvis and thigh in the transverse plane during the loading response, and between the thigh and shank in the sagittal plane during the loading response and pre-swing phases. This finding may provide more insight into explaining the sex bias of developing a knee injury; however, this hypothesis is purely speculation. More studies, particularly prospective in nature, are needed comparing coordinative variability between sexes, and between healthy and injured runners, as literature in this area is currently sparse. The sex differences cited in this study also suggest that future such studies on coordinative variability in healthy subjects should probably analyze females and males separately.

Funding

This work was supported in part by the Betty Foster McCue Fellowship for Human Performance and Development, the Marthe E. Smith Memorial Science Scholarship, and the Jan Broekhoff Graduate Scholarship through the University of Oregon.

Conflict of interest

The authors have no conflicts of interest to declare.

Acknowledgments

The authors would like to acknowledge Varneet Brar, Colin Lipps, Hao Tan, and Peter Shen for their assistance with data collection and processing.

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