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WOMEN IN BIOMECHANICS AND CONTROL OF HUMAN MOVEMENT: 2021

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Postural Balance Ability and the Effect of Visual Restriction on Older Dancers and Non-Dancers

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Dance has been suggested to be an advantageous exercise modality for improving postural balance performance and reducing the risk of falls in the older population. The main purpose of this study was to investigate whether visual restriction impacts older dancers and non-dancers differently during a quiet stance balance performance test. We hypothesized higher balance performance and greater balance deterioration due to visual restriction in dancers compared with non-dancers, indicating the superior contribution of the visual channel in the expected higher balance performances of dancers. Sixty-nine (38 men, 31 women, 74 ± 6 years) healthy older adults participated and were grouped into a Greek traditional dance group ($n = 31$, two to three times/week for 1.5 h/session, minimum of 3 years) and a non-dancer control group ($n = 38$, no systematic exercise history). The participants completed an assessment of one-legged quiet stance trials using both left and right legs and with eyes open while standing barefoot on a force plate (Wii, A/D converter, 1,000 Hz; Biovision) and two-legged trials with both eyes open and closed. The possible differences in the anthropometric and one-legged balance parameters were examined by a univariate ANOVA with group and sex as fixed factors. This ANOVA was performed using the same fixed factors and vision as the repeated measures factor for the two-legged balance parameters. In the one-legged task, the dance group showed significantly lower values in anteroposterior and mediolateral sway amplitudes ($p = 0.001$ and $p = 0.035$) and path length measured in both directions ($p = 0.001$) compared with the non-dancers. In the two-legged stance, we found a significant vision effect on path length ($p < 0.001$) and anteroposterior amplitude ($p < 0.001$), whereas mediolateral amplitude did not differ significantly ($p = 0.439$) between closed and open eyes. The dance group had a significantly lower CoP path length ($p = 0.006$) and anteroposterior ($p = 0.001$) and mediolateral sway amplitudes ($p = 0.003$) both in the eyes-open and eyes-closed trials compared with the control group. The superior balance performance in the two postural tasks found in the dancers is possibly the result of the coordinated, aesthetically oriented intersegmental movements, including alternations between one- and two-legged stance phases, that comes with dance. Visual restriction resulted in a similar deterioration of balance performance in both groups, thus suggesting that the contribution of the visual channel alone cannot explain the superior balance performance of dancers.

Keywords: falls prevention, physical activity, visual channel, proprioception, aging, biomechanics, dance exercise

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INTRODUCTION

Approximately 30–60% of people over 65 years of age experience unintentional falls at least once a year as a result of a loss of balance (Gill et al., 2005; Rubenstein, 2006). Falls are usually characterized by high incidence, high susceptibility to injury, and the severity of consequences, thus having a tremendous impact on older adults, with grave social and financial ramifications (Tinetti, 2003). Fall prevention in healthy elders can be achieved with a satisfactory level of physical conditioning (Cadore et al., 2014; Hamed et al., 2018) that targets age-related impairments in balance, strength, power, and neuromotor coordination (Bierbaum et al., 2013; Bohm et al., 2020). A recent meta-analysis concluded that exercise reduces the rate of falls by 23% and the number of older people who experience one or more falls by 15% (Sherrington et al., 2019). Multicomponent (i.e., a combination of endurance, muscle strength, balance exercises, and/or flexibility or coordination training) exercise programs are being recommended for older adults because evidence suggests they are an effective approach to reducing the risk of falling (Baker et al., 2007; Bouaziz et al., 2016).

In recent years, dance has emerged as an advantageous exercise modality for enhancing the postural stability of healthy elders. Dance involves highly coordinated intersegmental movements with body rotations around multiple planes and axes with continuous alternations between one and two-legged stance phases, thus resulting in a continuous need for postural control (Guzmán-García et al., 2011; Douka et al., 2019a). Dance exercise interventions of short (i.e., 8–12 weeks) (Granacher et al., 2012; Sofianidis et al., 2017) or long durations that employ different dance styles, such as traditional/folk or Latin, have been shown to significantly improve the balance performances of healthy old adults (Kattenstroth et al., 2010; Serra et al., 2016). Furthermore, dancing provides dancers with musical experience, acoustic stimulation, and rhythmic motor coordination; for that reason, it is considered a sensory-enriched form of physical activity that can trigger the integration of sensorimotor performance with perceptual abilities in the elderly population (Kattenstroth et al., 2010; Douka et al., 2019b).

During dancing, there is a high need to achieve an artistic result through memorized forms of aesthetic body configurations either on an individual level or in relation to other partners. Since the visual system provides direct information on the position of the body and its perception of orientation with respect to its surroundings (Horak et al., 1989; Horak, 2006), dancers are expected to rely strongly on visual inputs for postural regulation. It has been reported that contemporary dancers show better balance performance with eyes open compared with matched control groups of non-dancers (Golomer et al., 1999; Hugel et al., 1999; Pérez et al., 2014). Postural control relies on a synergistic relationship between the visual and neuromotor systems (Bonnet and Baudry, 2016), with postural control being better when the visual task involves gazing toward specific targets while attempting to preserve upright standing. It can, therefore, be argued that dancers who practice in acquiring visual feedback from their partners to achieve dance's aesthetic result while maintaining their own stability, would show an enhanced postural control that depends on visual information compared to non-dancers. Given that postural stability depends on the integrated sensory information processing from the visual, vestibular, and proprioceptive systems (Horak et al., 1989; Peterka, 2002), the deprivation of visual information might be more pronounced in dancers, who practice learning to dance through visual information. However, whether the visual channel is an important contributor to the advantageous postural balance performance of dancers has not yet been investigated.

Postural balance performance is typically assessed on the basis of the center of pressure displacement during quiet standing on a force platform. These derived center of pressure (CoP) values represent the geometrical location of the reaction force vector on the platform. Furthermore, CoP amplitude-based parameters are typically used as primary outcome measures and have been found to relate with age-related differences in balance performance (Prieto et al., 1996). These measures of static postural control are also typically employed in detecting sex-related differences, with some studies reporting greater CoP sway movement in older women (Kim et al., 2010; Riva et al., 2013; Chen et al., 2019), while others found older men to have increased postural sway compared with women (Era et al., 1997; Masui et al., 2005; Puszczalowska-Lizis et al., 2018).

The goal of this study was to investigate if visual restriction impacts older dancers and non-dancers differently during a quiet stance balance performance test. We hypothesized higher balance performance and greater balance deterioration due to visual restriction in dancers compared with non-dancers, indicating the superior contribution of the visual channel in the expected higher balance performances in dancers.

METHODS

Participants

In this cross-sectional study, a municipal senior club and four clubs instructing Greek traditional dance classes were contacted, with a total of 69 (38 men, 31 women, age: 74 ± 6 years) eligible senior adults volunteering to participate. Inclusion criteria required that the participants were healthy adults that were at least 65 years old. Participants were excluded if they reported a history of neuromuscular diseases, musculoskeletal disorders, cardiovascular or severe systemic diseases, severe arthritis, or if they had been taking any medication for the above diseases in the last 6 months.

The participants were grouped into a control group of non-dancers ($n = 38$, 20 men and 18 women, no systematic exercise history) and an experimental group of dancers ($n = 31$, 18 men and 13 women), whose participants had been exercising with Greek traditional dances at a frequency of two to three times per week for 1.5 h per session and for a minimum of 3 years. The study was approved by the Ethics Committee of the School of Physical Education and Sport Science, National and Kapodistrian University of Athens (approval number: 1152/11-12-2019), and all the participants gave their written informed consent in accordance with the Declaration of Helsinki.

Assessment of Postural Balance Performance

Balance performance was assessed in one-legged and two-legged quiet stance trials. The participants were tested in spacious, quiet rooms with appropriate light and temperature conditions, with a measuring device located in the middle of the room at an approximate distance of 2–3 m between the walls and the participants. During the assessment of the one-legged trials, the participants had their eyes open, stood barefoot with either their left or right leg on a force plate (Wii, A/D converter, 1,000 Hz, 24-bit resolution; Biovision), and maintained a straight body posture with their arms hanging relaxed on their sides. Their gazes were fixed on an imaginary point on the wall 2–3 m in front of them, while their heads were kept parallel to ground level. The order of the starting leg was randomized.

For the two-legged stance trials, the participants were instructed to keep their feet at hip-width apart and stand as motionless as possible with their eyes open, as described above, and eyes closed. In these trials, one researcher was always situated behind the participants for safety reasons. Two trials were performed per visual condition in a randomized order. The duration of every trial in the postural stability measurements was 20 s with 30 s of rest across trials and 1 min of rest between quiet stance conditions. Off-line, the data were filtered using a second bi-directional order digital low-pass Butterworth filter with a 15-Hz cut-off frequency and analyzed with MATLAB custom-made scripts (R2012a, 64 Bit; Mathworks, Natick, MA, United States).

The data were analyzed from the 1st to the 16th second ($\Delta t = 15$ s) of each 20-s trial time. Postural balance performance was determined by the following parameters: (a) CoP path length, defined as the sum of Euclidean distances of adjacent measurement points, and (b) CoP sway amplitude, defined as the range (i.e., from minimum to maximum) of the CoP values in the anteroposterior and mediolateral directions. Body height (Bryant et al., 2005) and mass may affect path length and sway amplitudes; therefore, the determined CoP parameters were normalized to body height and mass, with the normalized values (% of body height per kilogram of body mass) being used for statistical analysis. For the two-legged stance, the average value of the two trials was used for the analysis. On the other hand, for the one-legged stance, the average value of the left and right leg trial was also used for analysis.

Statistical Analyses

All statistical analyses were performed using SPSS Statistics (Version 17.0). The normal distribution of the CoP data was examined by a Kolmogorov–Smirnov test with Lilliefors correction. The statistical testing of normality failed (p values: 0.046–0.007 for the two-legged CoP parameters with eyes open and eyes closed and respective $p = 0.004$ to $p < 0.0001$ for the one-legged parameters with eyes open); however, upon visual inspection with quantile-quantile (Q-Q) plots, the CoP data were normal with slight deviations. A two-way ANOVA with group (non-dancers, dancers) and sex (male, female) as fixed factors was performed to test for possible differences in the anthropometric parameters and the one-legged balance performance parameters. An ANOVA for repeated measures was also performed, with vision (open, closed eyes) as the within-subjects factor and group and sex as between-subjects factors on the two-legged balance performance outcome measures. A Bonferroni-corrected pairwise analysis was conducted in the case of a significant interaction between the factors of vision, group, and sex. The level of significance for all the tests was set at a = 0.05. For the graphical representation of the outcomes, we used boxplots depicting the median and the 5th and 95th percentiles as whiskers.

RESULTS

Age was not significantly different ($p = 0.192$, $\eta_p^2 = 0.026$) between the men and the women, but the men were significantly heavier ($p = 0.032$, $\eta_p^2 = 0.069$), taller ($p < 0.001$, $\eta_p^2 = 0.479$), and had greater body mass indices ($p = 0.017$, $\eta_p^2 = 0.085$) than the women across the groups (**Table 1**). Age was significantly lower in the dancers compared with the non-dancers ($p < 0.001$, $\eta_p^2 = 0.29$), while body mass was significantly higher in the dancers ($p = 0.006$, $\eta_p^2 = 0.11$) compared with the non-dancers. Body height was also significantly higher in the dancers compared with the non-dancers ($p = 0.011$, $\eta_p^2 = 0.095$), while body mass index did not differ between the two groups ($p = 0.144$). There were no significant sex-by-group interaction effects in any of the anthropometric parameters (**Table 1**).

In the one-legged quiet stance condition, a significant main effect of group was found for path length ($p = 0.001$, $\eta_p^2 = 0.281$) and anteroposterior ($p = 0.001$, $\eta_p^2 = 0.292$) and mediolateral ($p = 0.035$, $\eta_p^2 = 0.124$) sway amplitudes of the CoP (**Figure 1**). *Post-hoc* pairwise comparisons showed that the dancers had a significantly smaller path length compared with the non-dancers ($p = 0.001$, **Figure 1**). The anteroposterior and mediolateral CoP sway amplitudes were also smaller in the dancers compared with the non-dancers ($p = 0.001$ and $p = 0.035$, respectively) (**Figure 1**). There was no significant main effect of sex on either path length ($p = 0.079$) or anteroposterior ($p = 0.563$) and mediolateral ($p = 0.208$) sway amplitudes across the groups (**Figure 1**). No significant ($p > 0.05$) sex-by-group interaction effect was found in any of the examined CoP parameter during the one-legged trials.

In the two-legged trials, we found a significant effect of group on path length ($p = 0.006$, $\eta_p^2 = 0.111$), and anteroposterior ($p = 0.001$, $\eta_p^2 = 0.146$) and mediolateral sway amplitudes ($p = 0.003$, $\eta_p^2 = 0.124$), with the *post-hoc* pairwise comparisons showing that the dancers had smaller values in these parameters both in the eyes-open and eyes-closed conditions compared with the non-dancers (**Figure 2**). No significant effect of sex was found for path length ($p = 0.06$) and the anteroposterior ($p = 0.868$) or mediolateral sway amplitudes ($p = 0.147$) of the CoP in the two-legged stance trials. There was a statistically significant main effect of vision on path length ($p < 0.001$, $\eta_p^2 = 0.559$)

TABLE 1 | Anthropometric data for the non-dancer (control) and dancer (dance) groups (means ± SD).

	Control ($N = 38$)		Dance ($N = 31$)	
	Males ($N = 20$)	Females ($N = 18$)	Males ($N = 18$)	Females ($N = 13$)
Age (yr)*	79 ± 6	75 ± 4	70 ± 4	71 ± 4
Body mass (kg)* +	76.2 ± 12.3	72.0 ± 9.4	84.7 ± 8.7	77.8 ± 10.5
Body height (cm)* +	167 ± 8	158 ± 4	174 ± 7	160 ± 5
Body mass index (kg/m^2) +	27.1 ± 3.7	28.9 ± 4.5	28.1 ± 2.4	30.6 ± 3.4

*Statistically significant group effect ($p < 0.05$). +Statistically significant sex effect ($p < 0.05$).

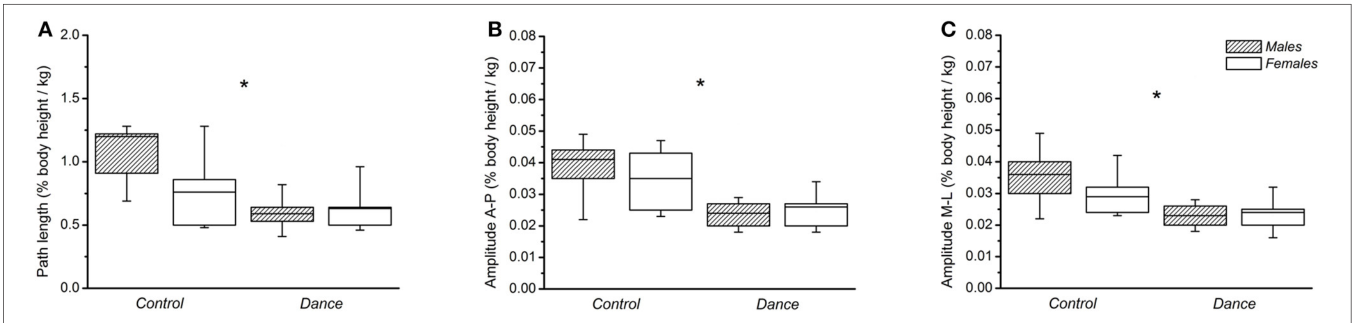

FIGURE 1 | (A) Normalized path length, **(B)** center of pressure (CoP)-amplitude in the anteroposterior direction, and **(C)** CoP-amplitude in the mediolateral direction in the one-legged quiet stance trials for men and women in the non-dancer (control) and dancer (dance) groups. *Statistically significant group effect ($p < 0.05$).

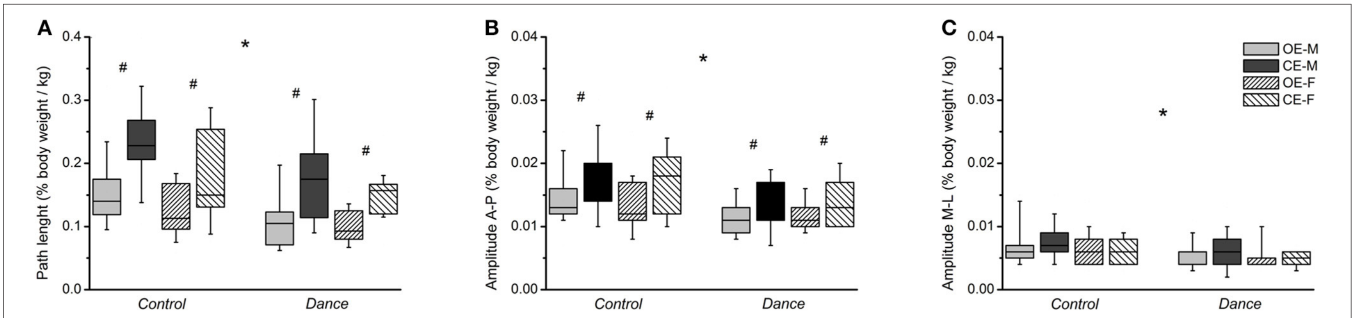

FIGURE 2 | (A) Normalized path length, **(B)** CoP-amplitude in the anteroposterior direction, and **(C)** CoP-amplitude in the mediolateral direction in the two-legged quiet stance trials with eyes open (EO) and eyes closed (EC) in the non-dancer (control) and dancer (dance) groups. (M) for men, (F) for women. *Statistically significant group effect ($p < 0.05$), #statistically significant vision effect ($p < 0.05$).

and the anteroposterior sway amplitude of CoP ($p < 0.001$, $\eta_p^2 = 0.339$) across the groups (**Figure 2**). However, there was no significant effect of vision on the mediolateral amplitude of CoP ($p = 0.439$, **Figure 2**). No significant vision-by-group interaction was also found for either path length ($p = 0.463$) or anteroposterior ($p = 0.237$) and mediolateral sway amplitudes ($p = 0.74$) (**Figure 2**). Furthermore, there was no statistically significant vision-by-sex interaction for the examined CoP balance parameters (path length, $p = 0.125$; anteroposterior amplitude, $p = 0.794$; mediolateral amplitude, $p = 0.129$) (**Figure 2**). Likewise, no significant vision-by-group-by-sex interaction was found for either path length ($p = 0.84$) or anteroposterior ($p = 0.499$) and mediolateral sway amplitudes ($p = 0.814$) (**Figure 2**).

DISCUSSION

In this study, we found a superior postural balance performance in older dancers during one- and two-legged quiet stance tasks compared with a control group of older non-dancers, supporting the first hypothesis. Furthermore, we found that visual restriction similarly deteriorated balance performance in both groups; therefore, the second hypothesis concerning the greater contribution of the visual channel for postural balance control in older dancers was rejected. Finally, the findings confirmed current reports that found that older men and women have similar balance performances in the investigating balancing tasks when gender differences in anthropometric parameters are accounted for.

During the two-legged quiet stance balance task, we found a decrease in balance performance (i.e., CoP trajectory) in the closed-eyes condition in an average of 52% compared with the open-eyes condition. These findings are in agreement with previous studies that reported an increase in postural sway area and total CoP displacement between 30 and 45% in older adults with vision restrictions (Paulus et al., 1984; Lord et al., 1991; Simoneau et al., 1992). Vision provides direct information on the position and orientation of the body with respect to the environment and, together with proprioception and the vestibular system, is an important mechanism of the neuromotor system in generating accurate motor commands for balance control (Horak et al., 1989). The removal or restriction of these visual inputs deteriorates balance performance during quiet stance (Sarabon et al., 2013; Wiesmeier et al., 2015) and, thus, is a key factor for the overall perception and representation of body positions and movements during postural control. Although there is evidence that the threshold of the visual system for the perception of body sway is lower compared with proprioception (Speers et al., 2002; Doumas et al., 2008) and that older adults use proprioceptive information from the lower limbs more rather than visual and vestibular sensory signals for balance control (Fitzpatrick et al., 1994; Lord and Menz, 2000), visual inputs have been found to detect low-frequency body motions (i.e., 0.01 to 1 Hz) most accurately (Horak et al., 1989), thus contributing to balance performance.

We found a similar increase in CoP trajectory in the closed-eyes condition in both dancers and non-dancers, indicating the negligible effects of exercise history on balance performance deterioration due to visual restrictions. Contemporary dancers exhibited greater balance ability when the visual channel was available (Golomer et al., 1999; Hugel et al., 1999; Pérez et al., 2014) as a result of their increased specialization in tasks where postural control is regulated *via* visual information. Accordingly, we expected a greater contribution of the visual system in older dancers because the execution of artistically oriented intersegmental movements, in combination with the needed movement synchronization and adjustment with other dance partners, challenges the neuromotor system and increases the need for vision in postural control during dancing. Therefore, we expected a greater deterioration of balance performance during the closed-eyes condition in this group. The findings (i.e., similar balance deterioration in both groups with closed-eyes), however, did not confirm a higher contribution of vision in the balance control of dancers; thus, the visual channel cannot explain the superior balance performance of the investigated older dancers. Proprioception due to the lower threshold for the perception of body sway compared with visual and vestibular systems and the key role in postural control (Speers et al., 2002; Doumas et al., 2008) might be a candidate to explain the higher balance performance of older dancers. The probable higher effectiveness of dancers in multisensory integration to control balance and orientation could be further examined in conditions of increasing balance difficulty and differentiating visual cues, as an earlier study showed that the effectiveness of these mechanisms depends on the ability of a participant to identify and exploit appropriate non-visual frames of reference (Isableu et al., 2010).

In recent years, growing research evidence has stressed the importance of dance as an advantageous exercise modality for improving stability performance (Sofianidis et al., 2009, 2017; Granacher et al., 2012; Rehfeld et al., 2017), which consequently reduces the risk of falls in older adults. The one-legged quiet stance is a challenging balance task because the small base of support requires the successful integration of sensory information from the visual, vestibular, and somatosensory systems by the central nervous system to produce appropriate postural responses (Nashner, 1981; Horak, 2006). Greek traditional dancing is characterized by coordinated, rhythmic, aesthetically oriented intersegmental movements with body rotations and the combinations of various steps under musical guidance, resulting in the continuous postural control of the body. In particular, through dancing, the lower extremities are engaged in a continuous alternation of the one- and two-legged stance phases, such as hopping, sideways steps, or crossing one foot over the other, thus favoring the strong sensorimotor control of body sway (Guzmán-García et al., 2011). As a result, dancers become very familiar with weight shifts that constantly challenge their postural control system to maintain equilibrium, as their centers of mass are voluntarily shifted to the limits of stability (Rehfeld et al., 2017). For instance, healthy older adults who have been participating in training interventions (8–12 weeks duration) under various styles of dance, either traditional/folk or modern, were able to achieve relevant improvements in one-legged stance balance performance compared with non-trained peers (Granacher et al., 2012; Rehfeld et al., 2017; Sofianidis et al., 2017). Thus, we can argue that the plethora of high-coordination movement repertoires in dancing, which challenges the human sensorimotor system, can trigger neuromuscular adaptational responses and consequently explain the observed superior balance performance of older dancers.

The findings also revealed no significant sex-related effect on one- and two-legged postural balance performance. Several studies have found that older men exhibit impaired balance performance (i.e., higher CoP sway movement) more than women (Era et al., 1997; Masui et al., 2005; Puszczalowska-Lizis et al., 2018), whereas other studies reported that older women have increased postural sway compared with men (Kim et al., 2010; Riva et al., 2013; Chen et al., 2019). It has also been reported that the smaller body height of women is an important explanatory factor for increased balance performance (Era et al., 1996; Bryant et al., 2005), with the difference between the sexes being largely removed when CoP trajectory was normalized to body height (Bryant et al., 2005). In our investigation, we used normalized body height and mass values to account for the observed greater mass and height in men; therefore, the findings showed that the postural balance performance of older men and women was similar when those parameters were accounted for.

This study has some limitations that should be addressed. The cross-sectional design of the study cannot suppress any selection effects on the balance performance on the investigated dancers (i.e., dancers inherently had greater balance ability and because of that self-selected to become dancers). The main purpose, however, was to examine the deterioration in balance performance due to visual restriction; therefore, the selection bias was reduced. To achieve this, the CoP outcomes were considered for 15 s in each condition. The protocol involved two visual conditions (eyes open and eyes closed) and two stance conditions (two-legged and one-legged); thus, with a longer duration per trial, fatigue would accumulate, most probably for the non-dancers, and could introduce bias into the comparisons between the two groups. Furthermore, the two groups were not exactly matched, as the dancers were heavier, taller, and younger compared with the non-dancers in the control group. The non-dancers were also not included in any systematic exercise programs, indicating generally lower physical activity levels compared with the dancers. Thus, we normalized the assessment parameters to body height and body mass excluding the bias due to body height and body mass; however, the bias of the different ages and physical activity levels remained in this study.

In conclusion, we found superior balance performance in the group of older dancers compared with the control non-dancer group without any systematic exercise experience. The restriction of visual information decreased balance performance similarly in both investigated groups, indicating that the visual channel was not responsible for the superior balance performance found in the dancers. The findings, however, indicate that dancing can be recommended as an advantageous exercise modality in order to improve postural stability as it has the potential to reduce the risk of falls in older adults. We argue that the advanced postural performance of dancers may be triggered by the aesthetically and highly coordinative movement repertoires in dancing that challenge sensorimotor integration.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the Ethics Committee of the School of Physical Education and Sport Science of the National and Kapodistrian University of Athens. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

M-EN and AA conceived the experiments, interpreted the data, and drafted the manuscript. M-EN performed the experiments and analyzed the data. AS and AA substantially contributed to the data analyses. VK and MK made important intellectual contributions during revision. All authors approved the final version of the manuscript and agreed to be accountable for the content of the article.

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Isolated Joint Block Progression Training Improves Leaping Performance in Dancers

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The purpose of this study was to investigate the effect of a 12-week ankle-specific block progression training program on *saut de chat* leaping performance [leap height, peak power (PP), joint kinetics and kinematics], maximal voluntary isometric plantar flexion (MVIP) strength, and Achilles tendon (AT) stiffness. Dancers (training group $n = 7$, control group $n = 7$) performed MVIP at plantarflexed (10°) and neutral ankle positions (0°) followed by ramping isometric contractions equipped with ultrasound to assess strength and AT stiffness, respectively. Dancers also performed *saut de chat* leaps surrounded by 3-D motion capture atop force platforms to determine center of mass and joint kinematics and kinetics. The training group then followed a 12-week ankle-focused program including isometric, dynamic constant external resistance, accentuated eccentric loading, and plyometric training modalities, while the control group continued dancing normally. We found that the training group's *saut de chat* ankle PP (59.8%), braking ankle stiffness (69.6%), center of mass PP (11.4%), and leap height (12.1%) significantly increased following training. We further found that the training group's MVIP significantly increased at 10° (17.0%) and 0° (12.2%) along with AT stiffness (29.6%), while aesthetic leaping measures were unchanged (peak split angle, mean trunk angle, trunk angle range). Ankle-specific block progression training appears to benefit *saut de chat* leaping performance, PP output, ankle-joint kinetics, maximal strength, and AT stiffness, while not affecting kinematic aesthetic measures. We speculate that the combined training blocks elicited physiological changes and enhanced neuromuscular synchronization for increased *saut de chat* leaping performance in this cohort of dancers.

Keywords: power, strength, joint kinetics, muscle, tendon, ankle

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INTRODUCTION

Progressive strength and conditioning programs that elicit increased power output are implemented in ballistic athletes to enhance stretch-shortening cycle (SSC) function (Cormie et al., 2010; Haff and Nimphius, 2012). As knowledge on the benefits of resistance exercise continually grows, additional training in ballistic athletes such as dancers, has become increasingly prevalent within dance companies, schools, and teams (Farmer and Brouner, 2021). Although the perception is shifting, optimal programming for dancers is still in the developmental stages (i.e., exercises, volume, intensity, and rest). For additional training to be effective, the following core questions must be addressed: (1) what is the overarching athletic goal, (2) what physical characteristics should be developed to achieve this goal, and (3) what are the known training modalities that will achieve these physical characteristics to achieve the overarching athletic goal? The pre-existing training volume dancers undergo from a young age must be considered when creating a strength and conditioning program. Furthermore, practitioners must be mindful of dancers' extreme range of motion and the need to preserve flexibility. While some researchers have accounted for training volume, range of motion, and flexibility, there is a lack of consideration for sport-specific training in dancers; that is, to model after dance-specific movements to further improve or expand physical attributes which enhance performance of dance-specific movement.

Saut de chat (split leap) performance is a fundamental skill in dance styles such as ballet, jazz, lyrical, and contemporary. Healthy dancers experience up to ∼3.4 × bodyweight of force during the unilateral take-off portion of a *saut de chat* (Jarvis and Kulig, 2016). To perform a *saut de chat,* dancers typically *chassé,* or take two steps, prior to a unilateral take-off. At take-off, dancers will *developpé*, or flex, the leading leg up to hip level into a full split aerial position (ideal would be 180°; see **Figure 3**). Throughout the entire leap, dancers are instructed to maintain posture wherein the hips are directly beneath the shoulders to achieve aesthetic appeal. The nature of this postural maintenance is recognized as an aesthetic constraint (described in further detail in a topical perspective Rice and Nimphius, 2020). That is, to leap as high (or as far) as possible while achieving a full split position, dancers must rely less on hip flexion and subsequent torque generation. Likely due to the aesthetic constraint, ankle power is reportedly higher than knee and hip power during the take-off phase of a *saut de chat* (Jarvis and Kulig, 2020). The importance of the ankle during leaping is further supported by recent findings where dancers with greater plantar flexion strength and whole body center of mass peak power output leap higher as well as *better* aesthetically (Rice et al., 2021). It was also found that medial gastrocnemius and Achilles tendon stiffness predicted leap height (Rice et al., 2021). Although limited data exists on isolated joint training, implementing additional training for dancers that emphasizes developing strength, power, and muscle-tendon properties, specifically about the ankle joint, may elicit improved dance-specific SSC performance like the *saut de chat*.

Isolated joint training, while seemingly too specific, is gaining traction among strength and conditioning practitioners to hone in on either weakness or strength of a joint's surrounding tissues (Baltich et al., 2014; Rajic et al., 2020; Rice and Nimphius, 2020). In the context of dancers and leaping, isolated ankle-joint training may serve a three-fold purpose: (1) to enhance dance-specific SSC performance (Rice et al., 2021), (2) to prevent injury (Moita et al., 2017), and (3) to maintain or increase aesthetic appeal (Brown et al., 2007). While it appears that multi-joint exercise interventions may still benefit dancers' athletic and qualitative dance-specific performance (Angioi et al., 2012; Dowse et al., 2020; Escobar Alvarez et al., 2020; Grigoletto et al., 2020), isolated ankle-joint exercises might elicit adaptations that directly translate to *saut de chat* leaping biomechanics while preserving aesthetics (Rice and Nimphius, 2020). Previous research demonstrates that increasing maximal strength, rate of force development, muscle cross-sectional area, tendon stiffness, joint stiffness, and joint power concomitantly result in improved SSC performance (Kyrolainen et al., 2005; Kubo et al., 2007; Lamas et al., 2012; Katsikari et al., 2020; Laurent et al., 2020). As is well-established in the strength and conditioning field, not one training regimen results in all the aforementioned training adaptations. Because of this, dancers might benefit from an isolated ankle-joint block progression training program that incorporates a variety of training modalities as opposed to a singular training style.

Block progression training is the organization of different training modalities in a successive fashion that assists in performance realization through phase potentiation (Suchomel et al., 2018). For example, the adaptations that occur from isometric training and plyometric training have shown to differ from one another (Kubo et al., 2017); however, both styles of training progress athletic development in different ways. When ordering the different training modalities, practitioners should consider the temporal aspect of adaptations elicited from each respective training modality. Most importantly, the final block of training should help to collectively realize physical adaptations for the overarching athletic goal to be accomplished. For dancers, the following sequence might best influence *saut de chat* leaping performance: isometric, dynamic constant external resistance (DCER), accentuated eccentric loading (AEL), and plyometric training modalities. It has been proposed that isometrics can be divided into two sub-categories: pushing/pulling and holding (Schaefer and Bittmann, 2017). Holding and *balancing* a pre-determined load, as opposed to exerting maximal force against a stationary resistance, likely requires greater specificity of neuromuscular control strategies (Schaefer and Bittmann, 2017). Thus, holding isometrics may tap into motor unit recruitment synchronization (feedback driven), whereas pushing/pulling isometrics might increase high threshold motor unit recruitment and maximum activation (central command driven) (Pucci et al., 2006; Jeon et al., 2020). Traditional strength training, referred to here as DCER (involving both eccentric and concentric muscle actions), has been studied from several aspects of neuroplasticity and strength development for improved athletic capabilities (Aagaard et al., 2002). For dancers, loading full range of motion exercises is crucial, particularly about the ankle-joint, to potentiate force-generating capabilities across a spectrum of joint angles (transferring to a *plié* prior to leaping) allowing for greater power output during SSC actions (Taber et al., 2016). Some known physiological effects of AEL are increased tendon stiffness, muscle fiber CSA, and number of sarcomeres in series (Vogt and Hoppeler, 2014). AEL additionally serves as a means of Achilles tendinopathy "*pre-habilitation*" (O'Neill et al., 2015), which is highly prevalent in dancers. Plyometric exercises, involving the intention to move as quickly as possible, overload the eccentric phase of a SSC and necessitate optimal muscle-tendon interaction during the transition (amortization) phase into the concentric phase (Hirayama et al., 2017). Due to several confounding factors, plyometric training effects vary, but have most consistently shown to increase SSC performance in some fashion (i.e., jump height, power output, joint stiffness, and joint moments) (Kyrolainen et al., 2005; Cormie et al., 2009; Hirayama et al., 2017).

Dancers tend to specialize much earlier than other team sport athletes and therefore generally have less experience with other types of exercise. Due to this, the prescribed stimulus is especially important for dancers; whose diverse choreography necessitates mixed training. The purpose of this study was to investigate the effect of an ankle-focused block progression training program (24 sessions) on *saut de chat* leaping performance, maximal plantar flexion strength, and Achilles tendon stiffness. Specifically, we sought to measure center of mass and joint kinetics and kinematics to identify biomechanical and aesthetic aspects of leaping performance. We hypothesized that the training group would significantly increase *saut de chat* leaping performance (leap height, center of mass peak power, ankle peak power, and braking ankle stiffness), maximal plantar flexion strength, and Achilles tendon stiffness after training compared to the control group.

MATERIALS AND METHODS

Experimental Design

To determine the effect of 12-weeks of ankle-specific block progression training, we assessed maximal strength, Achilles tendon stiffness, and *saut de chat* leaping performance in dancers. Dynamometry was used to assess maximal voluntary isometric plantar flexion strength at two ankle angles. Dynamometry as well as ultrasonic techniques were used to assess Achilles tendon stiffness. Reflective markers were placed on the dancers, and they completed six *saut de chat* leaps atop five force platforms surrounded by nine motion capture cameras. Relative center of mass peak power, leap height (center of mass displacement), ankle power, knee power, hip power, braking ankle stiffness, peak split angle, average trunk angle, and trunk angle variation were calculated from leaping trials. The training group then followed a 12-week ankle-specific block progression program including isometric, DCER, AEL, and plyometric training modalities while the control group continued dancing normally. Our contention was that dancers (whom already train 6+ h per week) may benefit from the previously described adaptations in each block progression for ankle-specific movements (Kanehisa et al., 2002; Haff and Nimphius, 2012; Suchomel et al., 2018). Block 1 (isometrics) was intended to enhance joint-specific strength at critical ranges of motion (Kanehisa et al., 2002; Kubo et al., 2006) and neuromuscular control. Block 2 (DCER) projected to continue strength enhancements with full range of motion dynamic exercises containing low-to-moderate repetition ranges. Block 3 (AEL) was anticipated to increase load tolerance related to power development (Haff and Nimphius, 2012). Lastly, block 4 (plyometrics) was programmed to maximize phase potentiation effects for SSC performance translation (Bohm et al., 2015). Thus, the residual training effects of the previous training blocks were intended to cooperatively prepare athletes for plyometric training that would subsequently result in improved *saut de chat* leaping performance. The overall structure of the blocks was intended to induce tendon remodeling, increase maximal strength, and improve ankle-specific SSC performance. Following the training,

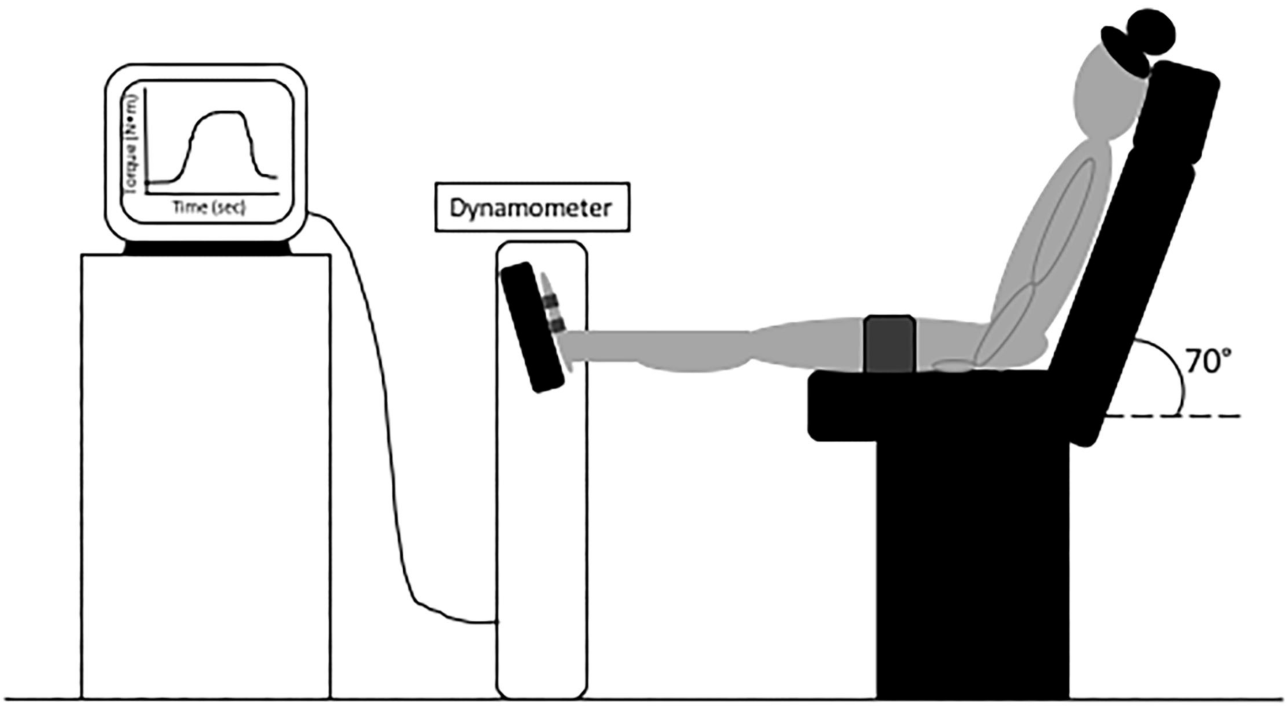

FIGURE 1 | Experimental set-up to perform maximal voluntary isometric plantar flexion and isometric ramping contractions with dynamometry.

all testing measures were repeated to determine whether there was a training effect on muscle-tendon properties and dance-specific SSC performance.

Experimental Procedures

A convenience sample size of dancers ($n = 14$) with a minimum of 10 years of ballet and jazz, modern, contemporary, or lyrical training that were dancing three or more times a week volunteered for the training group ($n = 7$; training age = 19.9 ± 5.6 year) or the control group ($n = 7$; training age = 19.4 ± 5.0 years). Due to some participants' schedules, the intervention was not feasible time-wise in addition to dance training, school, work, etc. The authors acknowledge that without random allocation, some level of bias may exist. The training group completed a 12-week ankle-specific block progression training program and dancing, while the control group continued dancing normally without additional training. Exclusion criteria required that participants reported no lower leg injuries within the previous 6 months, neuromuscular disease, or previous resistance training experience. Prior to data collection, the University Ethics Committee (#21229) approved all procedures. We sent an information letter to participants prior to arrival outlining all procedures.

Upon arrival, participants signed written informed consents and filled out a medical screening questionnaire. After we measured height and body mass, we performed whole body Dual-energy X-ray (DXA) scans (Hologic, Discovery A, Waltham, MA) to determine subject-specific lower limb segment masses for 3D motion capture kinematic and kinetic calculations. An operator positioned participants for the DXA scan while participants laid supine (Hart et al., 2015).

Participants then reported which leg was their preferred leaping leg (i.e., which leg would be leading) to determine the take-off leg that would be tested in maximal strength and tendon stiffness assessments with dynamometry (Biodex System 4, Biodex Medical Systems, Shirley, New York). We calibrated the Biodex prior to each testing session. Participants sat with the hips at an angle of 70° (slightly extended) in the Biodex (**Figure 1**). The preferred take-off foot was strapped securely in a pedal attachment with additional athletic tape. Furthermore, participants were moved to a position where the knee was slightly flexed at resting so that no heel lift occurred when the knee was fully extended. The testing leg was adjusted to be aligned directly in front of the hip both vertically and horizontally in the Biodex. To test maximal voluntary isometric plantar flexion (MVIP) at a slightly plantarflexed position, the ankle angle was moved to +10°. Participants then performed three MVIP trials, plantarflexing as fast and as hard as possible. We verbally encouraged participants and allowed 2 min of rest between each MVIP.

The ankle was then moved to a neutral position (0°), and an ultrasound probe (ProSound F75, Hitachi Healthcare Americas, Twinsburg, Ohio) with real-time imaging (5.0 MHz wave frequency with a 50-mm scanning length, 22 Hz) was

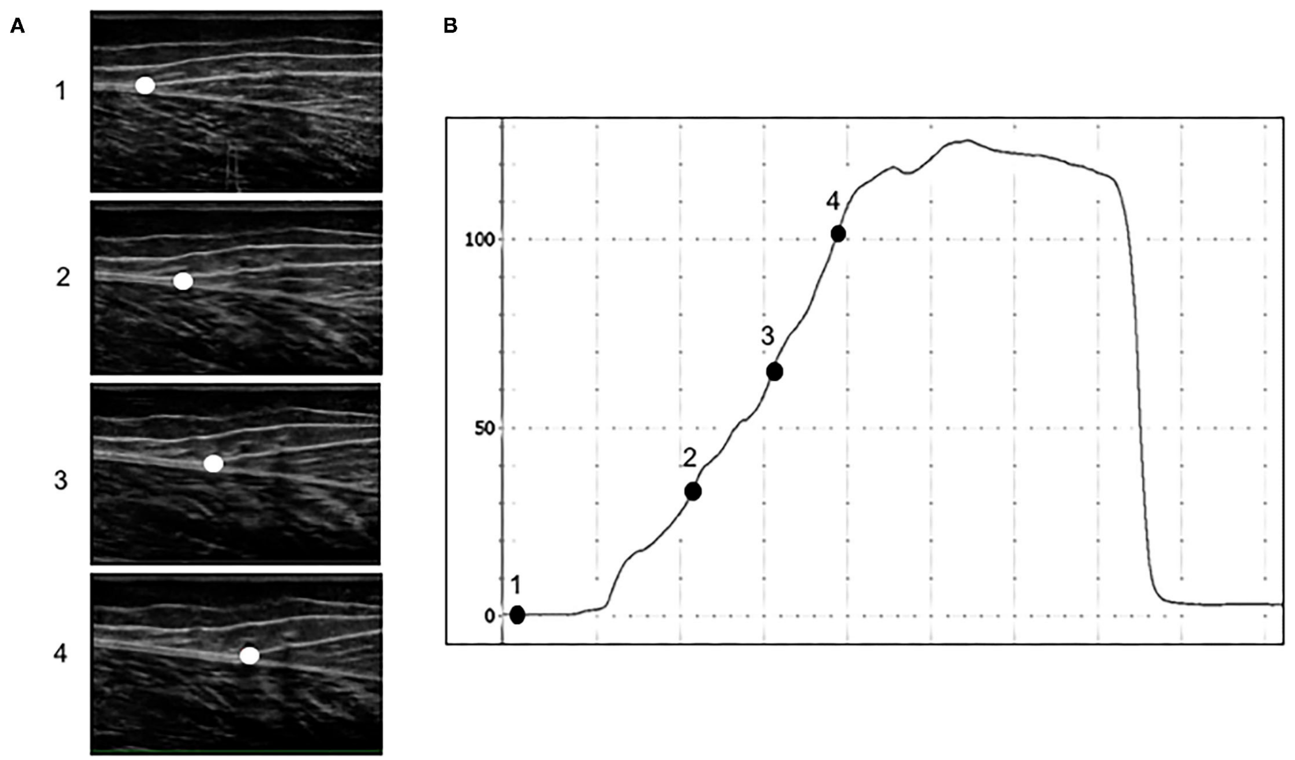

FIGURE 2 | An example of a participant's ramping isometric contraction at the medial gastrocnemius and Achilles tendon junction **(A)** ultrasound images and **(B)** torque-time curve during (1) resting, (2) 25%, (3) 50%, and (4) 80% of peak torque.

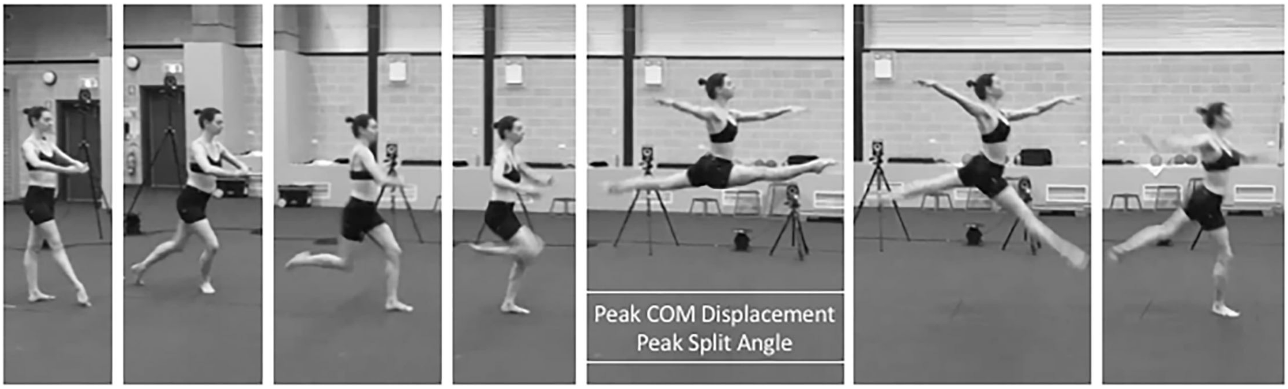

FIGURE 3 | An example of a participant equipped with reflective markers surrounded by 3D motion capture performing a *saut de chat* leap over in-ground force platforms.

secured to the skin at the gastrocnemius and Achilles tendon musculotendinous junction. Ultrasound data was synchronized with ankle torque output using LabChart software (version 8.1.5, ADInstruments, NSW, Australia) and a 16-bit analog to digital converter (PowerLab 16/35, ADInstruments, NSW, Australia). We again instructed participants to perform three trials of MVIP, each separated by 2 min of rest. After determining the highest peak torque from MVIP trials at a neutral ankle angle, we set visual feedback guidelines on a screen in front of participants ±10 N•m from the peak torque value. Participants performed three trials of ramping isometric contractions (**Figure 2**), wherein we asked that they isometrically plantarflexed for 3 s up to 100% of their target torque and steadily hold for 3 s before relaxing (McCrum et al., 2018). Each trial was separated by 90 s of rest. To obtain the triceps surae moment arm, we used a previously published digital photographic method (Pohl and Farr, 2010; Rice et al., 2017). Tendon force was calculated by dividing the peak torque at 25, 50and 80% of each ramping isometric contraction torque trial by the estimated triceps surae moment arm. We measured the corresponding tendon displacement at 25, 50, and 80% of peak torque from the initial resting position of each trial (**Figure 2**) with Tracker software (Version 5.1.4, https://physlets.org/tracker). Tendon forces and displacements were averaged across subjects and plotted against one another. Tendon stiffness for each participant was defined as the slope from 25 to 80%. The highest peak torque from the slightly plantarflexed (10°) and neutral (0°) MVIP trials was recorded for each dancer.

Participants then warmed up and stretched as they would normally to train, rehearse, or perform for the *saut de chat* leaps. We assessed *saut de chat* performance with three-dimensional motion capture to determine joint kinematics and kinetics with nine cameras (ViconMX F20, Vicon, Oxford, UK, 250 Hz sampling rate) surrounding five force platforms (900 × 600 mm, 9287CA and 9287BA, Kistler, Winterthur, Switzerland, 1,000 Hz sampling rate) beneath flooring (Mondo Sp.p.A., Alba, Italy). The motion capture system was calibrated prior to each data collection. We equipped dancers with 38 reflective markers (9.5 mm) in accordance with the UWA lower limb model (Chinnasee et al., 2018). Markers were adhered to the skin with double-sided tape on the trunk (C7, T10, clavicular notch, xiphoid process of sternum), hips (left and right anterior iliac spine, left and right posterior iliac spine), thighs (clusters of four), knees (right and left medial and lateral femoral condyles), lower legs (clusters of four), ankles (right and left medial and lateral malleoli), and right and left metatarsophalangeal and 5th metatarsal joints. For joint center determination and axes per the UWA kinematic and kinetic model, participants stood on a force platform surrounded by the cameras and performed a right leg "swinger" trial, a left leg "swinger" trial, five squats, and a static trial (Besier et al., 2003). Dancers then completed 2–3 familiarization leaps prior to data acquisition to a metronome of 106 beats per min (Jarvis and Kulig, 2016). We instructed dancers to take two steps, to leap as high as possible, and maintain arms in third position during the leap (one arm in the sagittal plane and one arm in the frontal plane, each at shoulder-height) (**Figure 3**; Wyon et al., 2013). Dancers performed a total of six *saut de chat* leaps.

We processed three-dimensional motion capture data with Nexus software (Version 2.11.0, Oxford, UK) and filtered marker trajectories with a 12 Hz cut-off frequency using a zero-lag low pass Butterworth 4th order filter. The three leaps with the highest whole body center of mass displacement (leap height), calculated from the UWA lower limb model output in Nexus, were selected for further analysis. The following leaping variables were analyzed with a custom-designed LabVIEW program (Version 19.0, National Instruments, Austin, TX): whole body center of mass peak power, joint powers, braking ankle stiffness, peak split angle, sagittal mean trunk angle and trunk angle range. Trials with insufficient marker data that were unable to calculate joint kinetics were removed from analysis. Marker data was deemed insufficient if gaps were unable to be filled as joint powers were unable to be computed. This resulted in a training group $n = 7$ and control group $n = 5$ for all pre- to post-testing joint kinetic measures. Forward dynamics were used to calculate velocity from force data. The initial center of mass position was corrected to maintain dynamic consistency since dancers were dynamic upon force platform contact (Rice et al., 2021). Relative center of mass peak power was calculated as the product of force and velocity. We obtained joint powers from the product of ankle-, knee-, and hip-joint moments and respective joint angular velocities during the take-off portion of the leap. Joint powers were normalized to body mass and relative concentric peak power was reported for each leaping trial. From here, relative joint power-time curves were re-sampled to 60 samples as was the average for all trials (the raw sample range was 49–75 samples). Pre-testing and post-testing average joint power-time curves were generated for training and control groups. Braking ankle stiffness was calculated from raw data as the slope of the ankle moment-angular velocity curve from force platform contact until the ankle angle began plantarflexing and contributing to propulsion. The braking ankle stiffness was then normalized to body mass. For aesthetic measures, peak split

TABLE 1 | Twelve weeks of isolated ankle-joint block progression training including sets, repetitions, inter-repetition rest, inter-set rest, intensity, and tempo of block exercises.

	Sets × reps	Inter-rep rest	Inter-set rest	Intensity	Tempo
Isometrics (Weeks 1–4)	3 × 3 s; “	3 s; “	60 s; “	85; 90; 95; 85%/ push/pull as hard as possible	0.3.0; “
Exercises	PF hold (L. muscle; Sh. muscle)	PF Push (L. muscle; Sh. muscle)	KB DF hold (neutral)	DF pull (L. muscle; Sh. muscle)	
DCER (Weeks 4–7)	4 × 6; “	—	90 s; “	85; 90; 95; 85%	1.1.1
Exercises	PF w/BB	Unilat. PF w/DB	Seated PF	DF w/KB and band	Ev/inversion w/band
AEL (Weeks 7–10)	3 × 6; “	—	90 s; “	130; 135; 140; 130%	3.1.1
Exercises (SM)	Standing PF	Seated PF	DF w/KB and band		
Plyometrics (Weeks 9–12)	4 × 6; “	—	1:10 work: rest; “	30; 35; 40; 30%	As fast as possible
Exercises	BB hops	Box drop hops	Unloaded band hops	SL side-side hops (med ball)	SL forward bounds (med ball)

PF, plantar flexion; DF, dorsiflexion; KB, kettlebell; L., long; Sh., short; BB, barbell; Unilat., unilateral; DB, dumbbell; Ev, eversion; SM, smith machine; SL, single-leg; “: the following weeks were the same as the initial week.

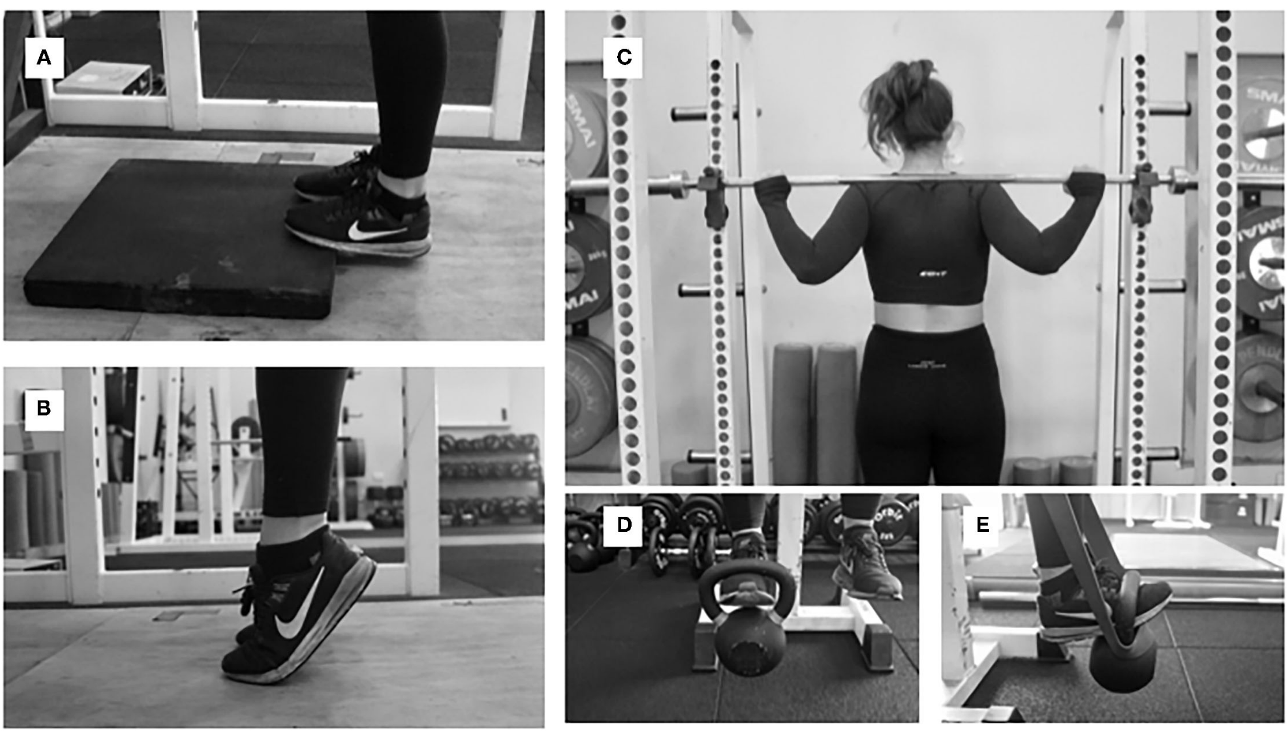

FIGURE 4 | (A) Long muscle ankle position, **(B)** short muscle ankle position, **(C)** bar setup for “pushing” isometrics, **(D)** ankle position for “holding” dorsiflexor isometrics with a kettlebell, and **(E)** band and kettlebell setup for DCER and AEL of dorsiflexor muscles.

angle, mean trunk angle and trunk angle range were determined. Peak split angle was determined from summated front leg global femur and tibia angles and summated back leg global femur and tibia angles calculated in Nexus. We identified the time at which peak split angle was the highest for both front and back leg angles by adding all four segment angles together and dividing by two. We recorded sagittal mean global trunk angle and trunk angle range from Nexus from when force decreased $<$10 N during take-off and at the onset of $>$10 N of force during landing. For statistical purposes, the absolute value of mean trunk angles were compared among groups and testing sessions, however, the true means and standard deviations were reported.

Training Intervention

After pre-testing, we instructed the control group to continue dancing normally. The training group attended two 30–45 min training sessions per week with the lead investigator (insert initials) for 12 weeks. **Table 1** lays out the ankle-specific multi-targeted block progression training program we implemented including four blocks: isometrics, dynamic constant external resistance, accentuated eccentric loading, and plyometrics (Rice and Nimphius, 2020). Prior to each workout, participants would complete a warm-up consisting of 10 bodyweight squats, 10 calf raises, 10 band dorsiflexions, and 10 band ankle eversions/inversions. To determine proper loading, dancers completed a one-repetition MVIP with the ankles at a neutral position standing atop a portable force platform with an immovable bar across the shoulders. The highest peak force from three trials was used to determine certain loads during the 12-weeks of training (see **Table 1**; **Figures 4–7**). We implemented a constant volume for each block but a progressive intensity for the first 3 weeks of each block followed by a downloading week that overlapped with the successive block. Once training concluded, participants rested for 1 week prior to commencing post-testing, which was identical to pre-testing.

Statistical Analysis

We performed a two-way factorial repeated measures analysis of variance (ANOVA) to identify within-subject effects and between-subject effects of training on group and time in the

FIGURE 5 | (A,B) Single-leg plantar flexion raises with performed with a dumbbell, and **(C,D)** single-leg seated plantar flexion with a Smith machine.

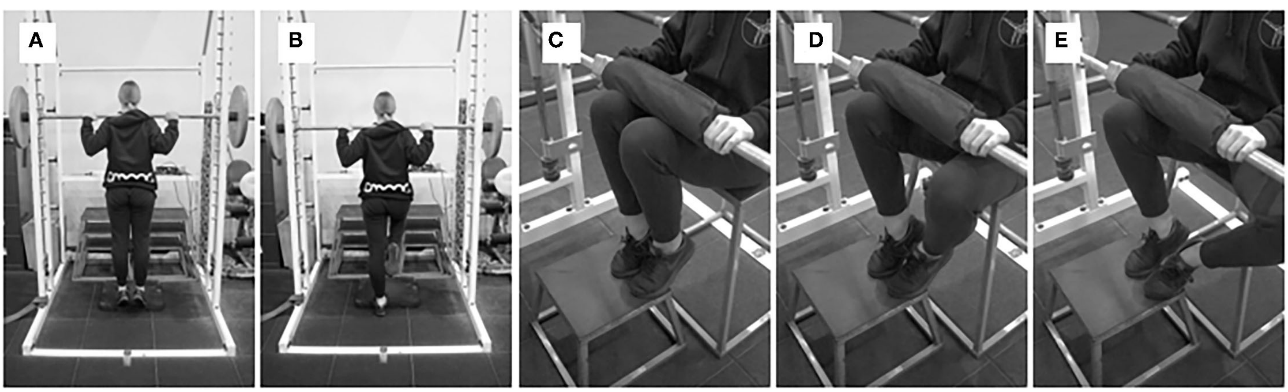

FIGURE 6 | (A,B) Standing AEL off plantarflexor muscles, and **(C–E)** seated AEL of plantarflexor muscles.

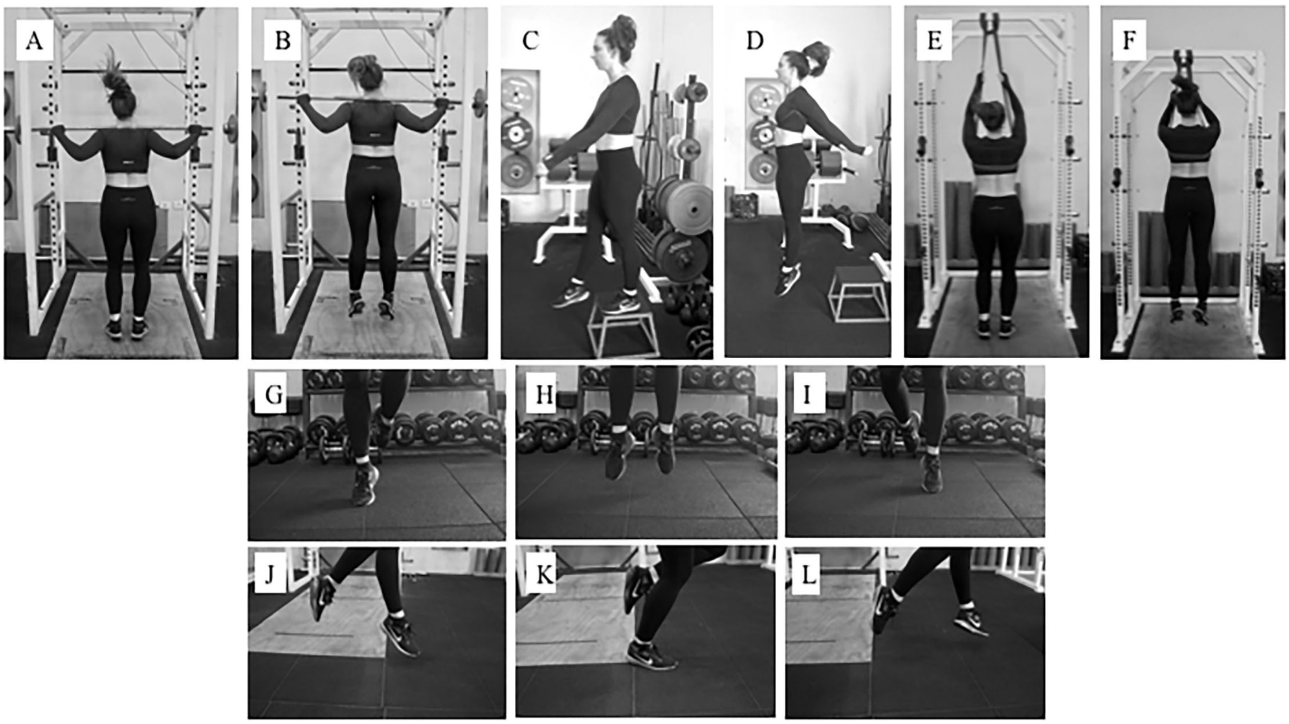

FIGURE 7 | (A,B) Barbell hopping, **(C,D)** Box drop hops, **(E,F)** Unloaded band drops, **(G–I)** Side-to-side hops, and **(J–L)** Single-leg forward bounds.

TABLE 2 | Anthropometric at pre- and post-testing sessions for the training and control groups.

Group	Age (yrs)		Height (m)		Body mass (kg)	
	Pre	Post	Pre	Post	Pre	Post
Training	24.86 ± 6.26	25.14 ± 5.98	1.62 ± 0.05	1.62 ± 0.05	58.38 ± 6.08	59.44 ± 7.33
Control	23.29 ± 4.39	23.71 ± 4.39	1.65 ± 0.04	1.65 ± 0.04	63.44± 10.10	63.40± 10.73

Values are shown as means ± SD.

following variables: Achilles tendon stiffness, MVIP peak torque at 0 and 10°, leap height, relative peak power during leaping, relative braking ankle stiffness, relative ankle peak power, relative knee peak power, relative hip peak power, sagittal mean trunk angle, and sagittal trunk angle range. Prior to performing each two-way repeated measure ANOVA, data was inspected for outliers and a Levene's Test of Equality of Variances to determine whether equal variance existed in the data between groups and testing sessions. For the average joint power-time curves, we performed an exploratory analysis using a multivariate ANOVA to determine whether group or time effects existed. We expressed results as mean ± SD. We additionally calculated mean differences (μd), 95% confidence intervals (CI), and Hedge's *g* effect sizes. Hedge's *g* effect sizes were interpreted as trivial (<0.25), small (0.25–0.50), moderate (0.5–1.0), and large (>1.0) (Rhea, 2004). Statistical analyses were performed using SPSS software (version 25.0, SPSS Inc., Chicago, IL, USA). We used Pillai's trace to test overall significance, which was set *a priori* at $P \leq 0.05$. We reported partial eta squared (η_p^2) for main and interaction effects, interpreted as small (0.01), medium (0.06), and large (0.14) (Cohen, 1992). Individual two-tailed Student's *T*-Tests were calculated if Pillai's trace indicated a significant effect of group, time, or group × time.

RESULTS

Anthropometry

For anthropometric measures between groups, no significant differences existed, and no significant changes occurred between pre-testing and post-testing (**Table 2**).

Maximal Voluntary Isometric Plantar Flexion Strength

A significant group effect existed between subjects ($P = 0.04$, $\eta_p^2 = 0.44$) and a significant time effect existed within subjects ($P = 0.005$, $\eta_p^2 = 0.62$) for 10 and 0° MVIP peak torque. Ensuing Univariate tests demonstrated that there was a significant time effect for 10° MVIP peak torque ($P = 0.007$, $\eta_p^2 = 0.47$) as well

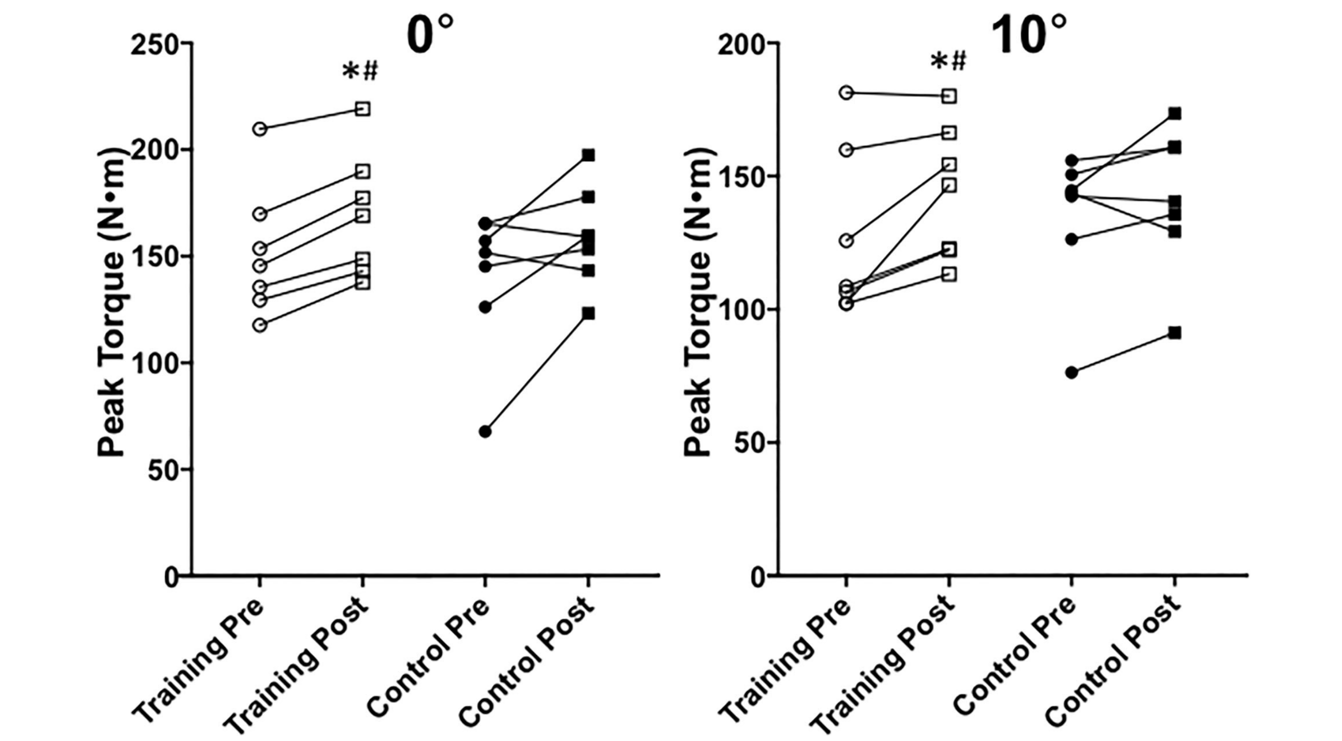

FIGURE 8 | Maximal voluntary isometric plantar flexion peak torque at 0° (neutral) and 10° (slightly plantarflexed) between training and control groups at pre- and post-testing sessions. *Indicates a significant ($P \leq 0.05$) training difference between pre- and post-testing sessions for the training group. #Indicates a significant ($P \leq 0.05$) difference between training and control groups at post-testing. Values are shown as individual data.

as 0° MVIP peak torque ($P = 0.001$, $\eta_p^2 = 0.61$). We found that maximal strength between the training and control groups was not significantly different at pre-testing (10°: $P = 0.22$; 0°: $P = 0.10$). At post-testing, the training group possessed significantly higher MVIP peak torque at 10° ($P = 0.05$, $\mu d = 37.65 \pm 41.63$ N; CI = 20.52 to 72.94; $g = 1.54$) and 0° ($P = 0.04$, $\mu d = 42.08 \pm 53.34$ N; CI = 9.51 to 89.16; $g = 1.60$) than the control group (**Figure 8**). **Figure 8** also demonstrates that the training group significantly increased MVIP peak torque at 10° ($P = 0.03$, $\mu d = 17.01 \pm 15.13$ N; CI = 3.01 to 31.00; $g = 0.56$) and 0° ($P < 0.0001$, $\mu d = 17.69 \pm 5.61$ N; CI = 12.50 to 22.88; $g = 0.55$) from pre- to post-testing. The control group's maximal strength did not significantly change from pre- to post-testing at either ankle angle (10°: $P = 0.20$; 0°: $P = 0.08$).

Saut de Chat Leap Height and Peak Power

A significant time x group effect was found for leap height ($P = 0.05$, $\eta_p^2 = 0.29$). The training group leaped significantly higher at post-testing than at pre-testing ($P = 0.02$, $\mu d = 0.036 \pm 0.030$ m; CI = 0.008 to 0.063; $g = 0.79$), and the control group remained unchanged ($P = 0.95$) pre- to post-testing. No significant differences existed between groups at pre-testing ($P = 0.85$) or post-testing ($P = 0.26$) for leap height. A significant group x time effect ($P = 0.02$, $\eta_p^2 = 0.40$) was found for relative peak power. The training group significantly increased relative peak power during leaping from pre-testing to post-testing (**Figure 9**; $P = 0.01$, $\mu d = 3.23 \pm 2.87$ W•kg^{-1}; CI = 0.58 to 5.89; $g = 0.51$). The control group's relative peak power did not significantly change from pre-testing to post-testing ($P = 0.09$) and was not significantly different from the training group at pre-testing ($P = 0.97$) or post-testing ($P = 0.98$).

Saut de Chat Leap Joint Kinetics

We found no significant main effects between groups at pre-testing for ankle power ($P = 0.13$, $\eta_p^2 = 0.997$). At post-testing, we found a significant group effect for ankle power between training and control groups ($P = 0.02$, $\eta_p^2 = 0.999$). Ankle power differed significantly between groups from 44 to 49% and 67 to 93% of the leap take-off (shown with shading in **Figure 10**). Ankle power remained unchanged from pre- to post-testing in the training group ($P = 0.50$, $\eta_p^2 = 0.96$) and the control group ($P = 0.28$, $\eta_p^2 = 0.98$). At pre-testing, groups did not significantly differ from one another in knee power ($P = 0.95$, $\eta_p^2 = 0.66$). Knee power did not significantly differ between training and control groups at post-testing either ($P = 0.51$, $\eta_p^2 = 0.95$). Knee power also remained unchanged from pre- to post-testing for the training group ($P = 0.77$, $\eta_p^2 = 0.88$) and the control group ($P = 0.99$, $\eta_p^2 = 0.44$). Hip power did not significantly differ between groups at pre-testing ($P = 0.36$, $\eta_p^2 = 0.98$) or post-testing ($P = 0.23$, $\eta_p^2 = 0.99$). From pre- to post-testing, hip power remained unchanged for the training

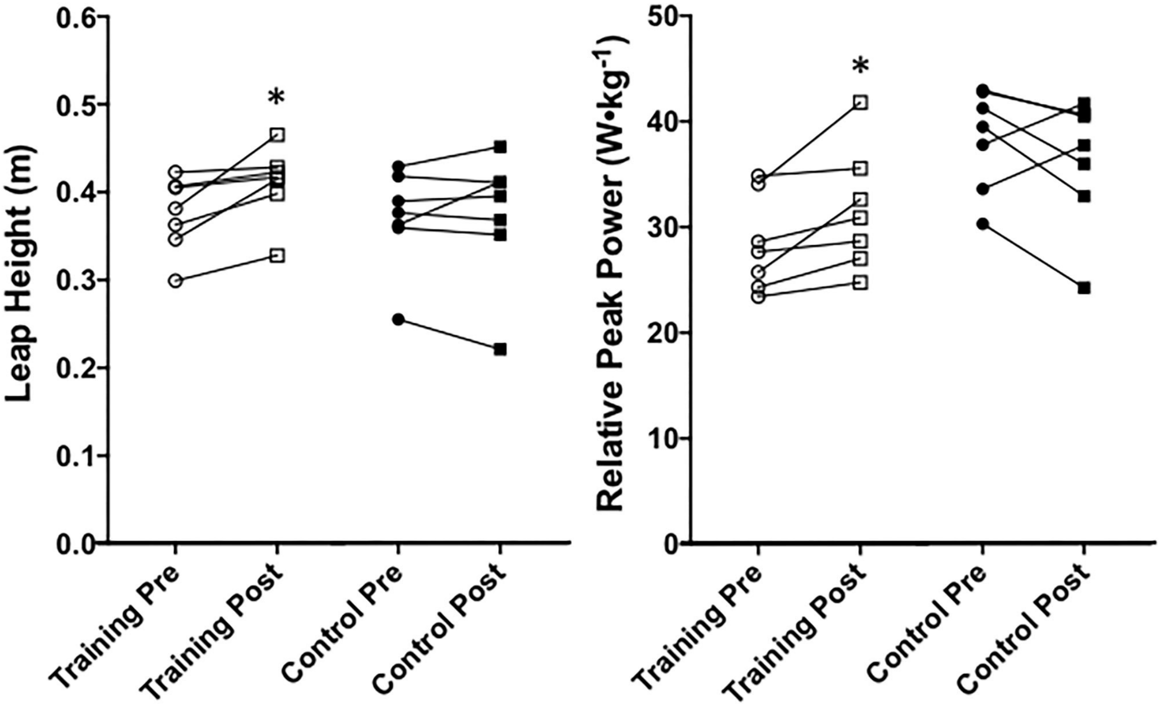

FIGURE 9 | Leap height and relative peak power of the center of mass during *saut de chat* leaping between training and control groups at pre- and post-testing sessions. *Indicates a significant ($P \leq 0.05$) training difference between pre- and post-testing sessions for the training group. Values are shown as individual data.

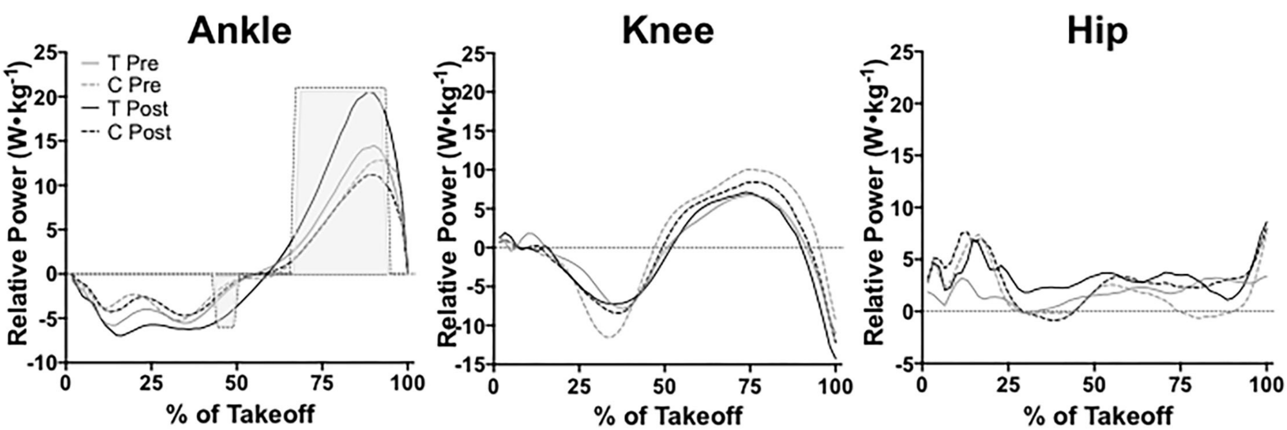

FIGURE 10 | Ankle, knee, and hip power during the entire take-off portion of leaping between training (T) and control (C) groups at pre- and post-testing sessions. Light gray shading indicates a significant ($P \leq 0.05$) difference between groups at post-testing. Values are shown as means.

group ($P = 0.76$, $\eta_p^2 = 0.89$). The control group's hip power also remained unchanged pre- to post-testing as well ($P = 0.67$, $\eta_p^2 = 0.88$).

We found a significant time x group effect for ankle peak power ($P = 0.04$, $\eta_p^2 = 0.35$), but not for knee peak power ($P = 0.26$, $\eta_p^2 = 0.13$) or hip peak power ($P = 0.53$, $\eta_p^2 = 0.04$). Ankle peak power did not significantly differ between the training group and control group at pre-testing ($P = 0.62$), however, the training group possessed significantly higher ankle peak power at post-testing than the control group ($P = 0.01$, $\mu d = 10.15 \pm 3.16$ W•kg^{-1}; CI = 2.91 to 17.39; $g = 1.77$). The training group also significantly increased ankle peak power from pre-testing to post-testing ($P = 0.04$, $\mu d = 6.51 \pm 6.35$ W•kg^{-1}; CI = 0.64 to 12.39; $g = 1.28$) shown in **Table 3**. The control group's ankle peak power did significantly change from pre- to post-testing ($P = 0.49$). We found a significant time x group effect for relative braking ankle stiffness ($P = 0.04$, $\eta_p^2 = 0.37$). Braking ankle stiffness significantly increased in the training group from pre- to post-testing ($P = 0.03$, $\mu d = 2.34 \pm 2.24$ W•kg^{-1}; CI = 0.27 to 4.40; $g = 1.25$), but not the control group ($P = 0.37$). No

TABLE 3 | Braking ankle stiffness, relative ankle, knee, and hip peak power between training and control groups at pre- and post-testing sessions.

Group	Ankle stiffness ($N \cdot rad^{-1} \cdot kg^{-1}$)		Ankle peak power ($W \cdot kg^{-1}$)		Knee peak power ($W \cdot kg^{-1}$)		Hip peak power ($W \cdot kg^{-1}$)	
	Pre	Post	Pre	Post	Pre	Post	Pre	Post
Training	3.80 ± 1.08	5.25 ± 1.44#	14.96 ± 4.64	20.56 ± 4.47*#	8.66 ± 3.24	8.31 ± 4.15	6.85 ± 2.89	6.50 ± 2.17
Control	5.87 ± 2.15	5.57 ± 2.32	13.98 ± 3.98	13.27 ± 5.93	10.42 ± 4.88	8.96 ± 2.67	6.09 ± 2.07	6.17 ± 1.57

*Indicates a significant ($P \leq 0.05$) training difference between pre- and post-testing sessions for the training group.
#Indicates a significant ($P \leq 0.05$) difference between training and control groups at post-testing.
Values are shown as means ± SD.

TABLE 4 | Aesthetic leaping variables (leap height, peak split angle, mean trunk angle, and trunk angle range) at pre- and post-testing sessions for training and control groups.

Group	Leap height (m)		Peak split angle (°)		Mean trunk angle (°)		Trunk angle range (°)	
	Pre	Post	Pre	Post	Pre	Post	Pre	Post
Training	37.48 ± 4.27	41.85 ± 4.10*	164.46 ± 6.35	164.55 ± 8.84	−5.17 ± 4.23	−4.76 ± 2.06	11.62 ± 1.72	12.34 ± 2.30
Control	37.79 ± 5.75	37.72 ± 7.59	163.45 ± 13.28	159.91 ± 15.25	−2.21 ± 4.02	−1.87 ± 5.26	7.71 ± 3.16	9.23 ± 3.72

*Indicates a significant ($P \leq 0.05$) training difference between pre- and post-testing sessions for the training group.
Values are shown as means ± SD.

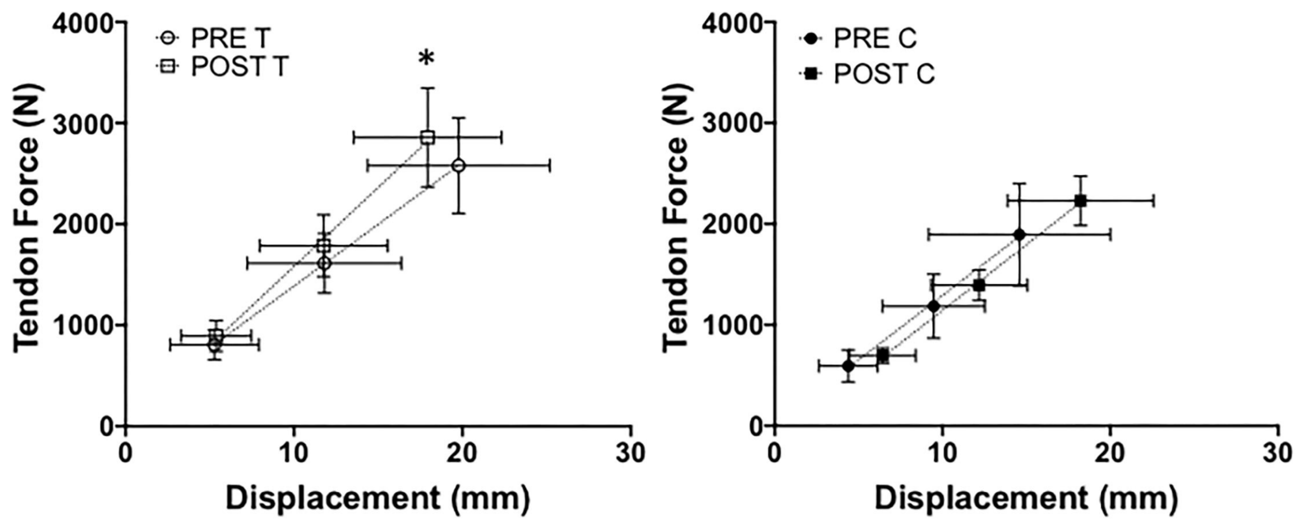

FIGURE 11 | Achilles tendon stiffness between training (T) and control (C) groups at pre- and post-testing sessions. *Indicates a significant ($P \leq 0.05$) training difference between pre- and post-testing sessions for the training group. Values are shown as means ± SD.

differences existed for braking ankle stiffness between groups at pre-testing ($P = 0.10$) or post-testing ($P = 0.23$).

Saut de Chat Leap Kinematics

We did not find a significant time x group effect for peak split angle during leaping ($P = 0.23$, $\eta_p^2 = 0.12$). Aesthetic trunk angle variables did not have a significant between time x group effect ($P = 0.48$, $\eta_p^2 = 0.12$) shown in **Table 4**.

Achilles Tendon Stiffness

Significant group x time interaction was found for Achilles tendon stiffness ($P < 0.0001$, $\eta_p^2 = 0.68$) shown in **Figure 11**. Tendon stiffness did not differ between groups ($P = 0.79$, $\eta_p^2 = 0.006$) at pre-testing ($P = 0.51$; PreT = 128.73 ± 45.49 N•mm^{-1}; PreC = 142.94 ± 41.60 N•mm^{-1}) or post-testing ($P = 0.14$; Post-T = 163.85 ± 52.44 N•mm^{-1}; Post-C = 136.90 ± 35.49 N•mm^{-1}). The training group significantly increased tendon stiffness from pre-testing to post-testing ($P < 0.0001$, $\mu d = 35.12 \pm 12.66$ N•mm^{-1}; CI = 23.41 to 46.83; $g = 0.67$) while the control group remained unchanged ($P = 0.40$).

DISCUSSION

This study assessed whether an isolated ankle-joint training program would influence muscle-tendon properties that contribute to dance-specific SSC function. The main findings of our study indicate that 12 weeks of ankle-specific block progression training for dancers significantly improves (1) *saut de chat* performance, (2) maximal plantar flexion strength, and (3) Achilles tendon stiffness, while not negatively affecting aesthetics (trunk and peak split angle variables). In particular, ankle peak power increased by an average of 59.8%, braking ankle stiffness increased by an average of 69.6%, *saut de chat* leap height increased by an average of 12.1%, and center of mass peak power increased by an average of 11.4%, after training. We speculate that our training progression (as was intended) improved maximal strength and tendinous tissue properties that translated into ankle kinetics during leaping, which likely contributed to enhanced overall *saut de chat* performance (Rice et al., 2021).

As hypothesized, *saut de chat* leaping mechanics significantly improved after 12 weeks of isolated ankle-joint training. We believe that this was due to greater availability and realization of muscular power surrounding the ankle joint, which also appeared in higher center of mass peak power. Our joint power-time curves were comparable to previously measured leaping joint kinetics, further supporting the hypothesis that the ankle contributes most to performance (Perry et al., 2019; Jarvis and Kulig, 2020). In order to further explore the roles of the ankle, knee, and hip during leaping, we computed correlations (for all participants) among joint peak power data and leap height prior to training ($n = 12$). We found a significant, strong relationship between ankle peak power and leap height ($P = 0.01$, $r = 0.69$), and insignificant relationships between knee peak power and leap height ($P = 0.26$, $r = 0.36$) and hip peak power and leap height ($P = 0.38$, $r = -0.28$). Jarvis and Kulig have similarly observed the ankle joint both absorbs and generates the greatest amount of mechanical energy during the take-off portion of a *saut de chat* (Jarvis and Kulig, 2020). The concomitant increases in leap height, braking ankle joint stiffness, and Achilles tendon stiffness further support the prominent role of the ankle during leaping. The authors acknowledge the low sample size and recognize that our results warrant further investigation with more participants. Nevertheless, unmistakable adaptations to leap performance and dynamic joint control occurred. We encourage dance practitioners to incorporate additional training for the ankle as a means of increasing ankle power and dynamic joint control during dance-specific SSC actions.

We demonstrated that isometric, DCER, AEL, and plyometric training significantly increased *saut de chat* leap height without affecting aesthetics (peak split angle, mean trunk angle, and trunk angle range). After strength, plyometric, and power training, previous research indicates that dancers significantly improve subjective dance performance (Brown et al., 2007; Girard et al., 2015; Dowse et al., 2020). In contrast, we elected to quantitatively report leap height, split angle, and trunk control to gauge the effect of our training program on aesthetic competency. We suspect the higher leap height was due to the multitude of targeted neuromuscular and mechanical musculotendinous adaptations (Suchomel et al., 2018), which manifested into greater neural drive and force-generating capabilities (Aagaard et al., 2002). To our surprise, training group participants provided feedback that one of the noticeable training effects (namely during isometrics), was improved balance. In support of this, Trajković et al. discovered that plantar flexion and dorsiflexion strength were predictors of postural balance in a large cohort of elite athletes (Trajkovic et al., 2021), suggesting that an increase in strength would simultaneously improve balance. At least one exercise required spinal loading with a barbell in each block of our program. While anecdotal, it may be that improved and maintained aesthetics occurred due to ankle-joint loading and the accessory requirement of postural maintenance during several exercises.

Maximal strength at 10 and 0° of plantar flexion increased on average by 17.0 and 12.2%, respectively (14.6% when averaged together), after block progression training. In dance styles such as ballet, jazz, and lyrical, both aesthetic appeal (subjective evaluation) and performance (greater impulse) are influenced by hyper-plantar flexion (Koutedakis and Jamurtas, 2004; Rice et al., 2018). Dancers have displayed that peak torque production at a slightly plantarflexed position better predicts *saut de chat* leaping peak power than at a neutral position (Rice et al., 2021). Thus, we believe that the increase we observed in dance-specific SSC peak power was a downstream effect of targeted full range of motion ankle strength and power development in our training program. Moss et al. found that a strong relationship existed between one-repetition maximum and maximal power with a 2.5 kg load, postulating that heavy resistance training concomitantly benefits lighter load performance (Moss et al., 1997), relevant to dancers. Other exercise intervention studies with dancers as participants have similarly found strength and power to improve with different training modalities, such as resistance and plyometric training (Brown et al., 2007; Angioi et al., 2012; Dowse et al., 2020), however, ankle strength and *saut de chat* peak power were not measured. Moreover, it has been suggested that lower strength levels may be associated with higher injury rates in dancers (Moita et al., 2017), which again incites these athletes to partake in some combination of additional resistance training.

We lastly found that Achilles tendon stiffness increased from training by 29.6% on average. Our increases are comparable with previous findings wherein the most robust increases in tendon stiffness resulted from isometric or eccentric loading regimens (see review Bohm et al., 2015). During both voluntary and involuntary muscle contractions, tendinous tissues deform in response to mechanical loading of the muscle. By manipulating frequency, intensity, and rate of strain, mechanotransduction cell signaling pathways can stimulate functional adaptation to occur in tendinous tissues (Lavagnino et al., 2003). Increasing tendon stiffness may amplify muscle power output based on greater resistance to deformation during SSC actions (Hirayama et al., 2017). Interestingly, dancers with varying tendinopathies have shown to exhibit altered joint kinetics during leaping, possibly due to poor technique subsequently affecting leap performance and injury pre-disposition (Fietzer et al., 2012; Shih et al., 2021). From a clinical perspective, AEL addresses some of the overuse and strength disparities observed in individuals suffering from tendinopathies (O'Neill et al., 2015). The authors would like to highlight that while AEL stimulates collagen fiber cross-linkage formation (Maffulli et al., 2012), nutrition status and sufficient rest to support tendon remodeling are equally critical for tendon health.

In conclusion, a 12-week block progression program including isometric, DCER, AEL, and plyometric training modalities effectively improves dancers' SSC performance *via* positive alterations in muscle-tendon properties. Our findings specifically indicate that isolated ankle-joint training for aesthetic athletes increases maximal plantar flexion strength and Achilles tendon stiffness that likely translate into increased *saut de chat* ankle peak power, braking ankle stiffness, center of mass peak power, and leap height. Similar to other effective training interventions (Aagaard et al., 2002; Brown et al., 2007; Cormie et al., 2009; Hirayama et al., 2017), we contend that joint-specific strength and power increases are advantageous to dancers from both a performance enhancement and an injury prevention lens (Rice and Nimphius, 2020). In agreement with preceding researchers' beliefs, dancers are a unique group of highly-specialized athletes that require additional training to supplement the volume and loads of training they experience (Koutedakis and Jamurtas, 2004). Ultimately, our ankle-specific block progression training program positively impacted sport-specific performance, which should be the primary focus of all strength and conditioning approaches.

Limitations

While our findings present interesting data on the effects of isolated joint training for dancers, there are some limitations. The small sample size ($n = 7$ in both groups) necessitates reproduction of the results with a larger sample size to generalize the outcomes for both dancers and other athletes. Specifically, two "visual outliers" existed for leap height and relative leap peak power: a training participant that improved quite a bit and a control participant that worsened some. Although equal variance existed in our data, these individuals could have affected statistics. Future researchers might also seek to compare isolated ankle-joint training with a generic resistance training program to verify that isolated joint training has added benefits. Manipulation of the block progression training (i.e., training modalities, block order, exercises, volume, intensity, rest, and program duration) might help to optimize training for dancers to improve *saut de chat* leap performance. Lastly, it would be interesting to observe empirically whether postural balance changes with isolated ankle-joint training in dancers by measuring center of pressure variables. Although more research is always needed, our training regimen did appear to induce neuromuscular and musculotendinous adaptations that benefited the overall athlete.

Practical Applications

Strength and conditioning practitioners working with highly specialized athletes should determine the types of appropriate movements and loads which will not only maintain already developed skills, but improve them as well. Aesthetic athletes require additional training, distinct from team sport athletes, to target joint-specific strength and power. We aimed to improve maximal plantar flexion strength, Achilles tendon stiffness, and *saut de chat* performance with a block progression program including isometric, DCER, AEL, and plyometric training modalities. By isolating the ankle-joint during additional training, dancers appear to successfully translate improved muscle-tendon properties and SSC function into dance-specific leaping performance. Given the near 20 years of dance training our participants possessed, enhancing their execution of existing movement strategy for improved performance is a difficult task to accomplish. We found that by employing additional training that targets ankle-specific stretch-shortening cycle, neuromechanical, and muscle-tendon unit development, dancers are capable of expanding on already well-engrained movement execution strategies. Strength and conditioning approaches specifically addressing sport-specific joint loads may benefit overall athletic prowess and should be further investigated. We hope that sport science and dance science practitioners alike will consider implementing strength and conditioning approaches that address sport-specific constraints, goals, and ultimately, individual athletes' needs to improve overall performance.

DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding author.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Edith Cowan University. The participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

PR contributed to the study design, data acquisition, data analysis, initial writing of the manuscript, and editing of the manuscript. KN and SN contributed to the study design and editing/final preparation of the manuscript. All authors contributed to the article and approved the submitted version.

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Walking on Mild Slopes and Altering Arm Swing Each Induce Specific Strategies in Healthy Young Adults

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Slopes are present in everyday environments and require specific postural strategies for successful navigation; different arm strategies may be used to manage external perturbations while walking. It has yet to be determined what impact arm swing has on postural strategies and gait stability during sloped walking. We investigated the potentially interacting effects of surface slope and arm motion on gait stability and postural strategies in healthy young adults. We tested 15 healthy adults, using the CAREN-Extended system to simulate a rolling-hills environment which imparted both incline (uphill) and decline (downhill) slopes (± 3°). This protocol was completed under three imposed arm swing conditions: held, normal, active. Spatiotemporal gait parameters, mediolateral margin of stability, and postural kinematics in anteroposterior (AP), mediolateral (ML), and vertical (VT) directions were assessed. Main effects of conditions and interactions were evaluated by 2-way repeated measures analysis of variance. Our results showed no interactions between arm swing and slope; however, we found main effects of arm swing and main effects of slope. As expected, uphill and downhill sections of the rolling-hills yielded opposite stepping and postural strategies compared to level walking, and active and held arm swings led to opposite postural strategies compared to normal arm swing. Arm swing effects were consistent across slope conditions. Walking with arms held decreased gait speed, indicating a level of caution, but maintained stability comparable to that of walking with normal arm swing. Active arm swing increased both step width variability and ML-MoS during downhill sections. Alternately, ML-MoS was larger with increased step width and double support time during uphill sections compared to level, which demonstrates that distinct base of support strategies are used to manage arm swing compared to slope. The variability of the rolling-hills also required proactive base of support changes despite the mild slopes to maintain balance.

Keywords: gait, stability, posture, arm swing, uphill, downhill

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INTRODUCTION

Everyday walking environments are complex as they vary in levelness and regularity (Allet et al., 2008). Challenging terrains require gait pattern modifications, through changes in spatiotemporal gait characteristics, kinematics, and kinetics, to accommodate the mechanical constraints. Responses to challenging terrain by the postural control system can be seen in adjustment of spatiotemporal gait characteristics. Compensatory changes such as increased double-support time or step width are a means of coping with uphill or downhill slopes, respectively (Kawamura and Tokuhiro, 1991; Sun et al., 1996; Gottschall and Nichols, 2011). The effectiveness of such changes may be determined by additionally quantifying stability. For example, taking wider steps has been linked to increased mediolateral margin of stability (ML-MoS) (McAndrew Young and Dingwell, 2012), indicating enhanced stability. Vieira et al. (2017) found downhill walking decreased ML-MoS and uphill walking increased ML-MoS compared to level walking, but not all concomitant gait strategies were explored.

The Americans with Disabilities Act mandates that sidewalks have a slope of <2.86°, and ramps be <4.76° (United States, 2010). Thus, everyday uneven terrain includes slight slopes ranging from 0 to 3°, yet most sloped walking studies examined larger and continuous slopes (3–20°) rather than smaller varying slopes (Sun et al., 1996; Leroux et al., 2002; Minetti et al., 2002; Prentice et al., 2004; Lay et al., 2006; Silverman et al., 2012; Kimel-Naor et al., 2017). Investigations of continuous 3° slope (Finley and Cody, 1970; Kawamura and Tokuhiro, 1991; Sun et al., 1996) found no differences between gait walking uphill compared to downhill. However, Prentice et al. (2004) found that stepping onto a 3° incline from level required modified swing limb kinematics, such as increased lower extremity joint flexion, and increased trunk forward inclination (Prentice et al., 2004). Recently, a rolling-hills (-3 to +3°) condition was used to simulate destabilizing terrain (Sinitski et al., 2015, 2019), but uphill and downhill steps were not examined separately despite the unique postural strategies required for each (Leroux et al., 2002).

During walking, the natural 1:1 contralateral arm-leg swing pattern reduces gait's metabolic cost by controlling angular momentum about the vertical axis of the center of mass (COM) (Meyns et al., 2013). This antiphase arm-leg swing pattern can be modulated by adjusting either arm motion or leg motion, which demonstrates the bidirectional nature of this relationship (Bondi et al., 2017). Different arm swing strategies have been shown to have unique impacts on gait stability. For example, walking with arms held may improve stability by increasing trunk inertia which limits CoM movement (Bruijn et al., 2010; Pijnappels et al., 2010). Conversely, some studies found decreased postural control and increased metabolic cost when walking without arm swing (Collins et al., 2009; Punt et al., 2015; Yang et al., 2015), or no difference in postural control between absent and normal arm swing (Bruijn et al., 2010; Hill and Nantel, 2019; Siragy et al., 2020). Alternatively, active arm swing may increase stability by more aptly counterbalancing torques that act on the COM's trajectory (Nakakubo et al., 2014; Punt et al., 2015; Yang et al., 2015; Wu et al., 2016). However, active arm swing's contribution to walking stability remains conflicting (Collins et al., 2009; Bruijn et al., 2010; Meyns et al., 2013; Siragy et al., 2020), especially when walking on challenging terrains.

The purpose of this study was to examine the effect of arm swing on spatiotemporal gait parameters, margin of stability, and postural strategies during uphill and downhill sections of a rolling-hills terrain. We expected that walking on slopes (uphill or downhill sections) with arms held would have compound increases in compensatory gait strategies that may increase stability, while the gait changes from active arm swing would conflict with the compensatory strategies used to navigate sloped walking.

METHODOLOGY

Fifteen healthy adults (8 male, 7 female; age 23.4 ± 2.8 years; height 170.2 ± 8.1 cm; weight 72.3 ± 13.5 kg) volunteered from the Ottawa area. An a priori power analysis revealed that 12 participants were adequate to achieve power at ß = 0.8. Participants had no neurological or orthopedic disorders affecting gait and no musculoskeletal injuries in the previous 6 months. The study was approved by the Institutional Review Board (University of Ottawa) and the Ottawa Hospital Research Ethics Board; all participants provided written informed consent.

Data Collection

Three-dimensional motion capture was completed using the Computer-Assisted Rehabilitation Environment (CAREN; CAREN-Extended, Motek Medical, Amsterdam, The Netherlands, **Figure 1**). This system combines a 6 degree-of-freedom platform with integrated split-belt instrumented treadmill (Bertek Corp., Columbus OH), 12-camera VICON motion capture system (Vicon 2.6, Oxford, UK), and 180° projection screen. Participants wore a torso harness attached to an overhead structure when on the treadmill. Platform motion was tracked by three markers, and full body kinematics collected using a 57-marker set (Wilken et al., 2012). Motion data were gathered at a rate of 100 Hz.

Experimental Protocol

For each trial, participants walked in a virtual park scenario which included a 20 m simulated rolling-hills terrain preceded and succeeded by 40 m of level walking. The rolling-hills terrain was produced by platform oscillations in the sagittal plane (pitch) based on a sum of four sines with frequencies of 0.16, 0.21, 0.24, and 0.49 Hz (Sinitski et al., 2015). Treadmill speed used the self-paced algorithm described by Sloot et al. (Sloot et al., 2014) (Methods 2c) which incorporated anterior–posterior pelvis position, velocity, and acceleration, referenced to the person's initial standing position (heels at the anterior-posterior midline of the treadmill). Visuals on the projection screen matched treadmill and platform conditions in speed and slope.

Trial order was randomized. Separate trials occurred for the three arm conditions: held, normal, and active. Instructions for the held condition were to volitionally hold arms in a still, relaxed position at the participant's sides. For the active condition, participants were instructed that the arms should be roughly horizontal at peak anterior arm swing.

Uphill sections included steps occurring when the average slope of the platform was between +1 and +3 degrees; downhill sections included steps occurring when the average slope of the platform was between −1 and −3 degrees (**Figure 2**). No uphill or downhill steps spanned a peak or trough in the rolling-hills terrain. Level walking included steps from the middle 20 m of the 40 m flat section preceding the rolling-hills terrain.

Data Analysis

Data were imported into Visual3D (C-Motion, Germantown, MD). Kinematic data were filtered at 10 Hz using a 4th order, zero-lag low-pass Butterworth filter, chosen using a residual analysis approach (Winter, 2009). Heel strike and toe-off gait events were calculated using a velocity-based algorithm as previously described (Zeni et al., 2008) and verified using ground reaction forces. Spatiotemporal parameters included speed, step length, step width, step time, percent double-support time (DST), and coefficients of variation (CoV) for step length, step width, step time, and percent double-support time. Speed was retrieved from D-Flow [Motek Medical, Amsterdam, The Netherlands; (Geijtenbeek et al., 2011)] which served as the control software for the CAREN system; we then averaged the speed over each step. Gait stability was quantified using mediolateral margin of stability (ML-MoS) and ML-MoS CoV using previously reported methods (Hof et al., 2005; Hak et al., 2013; Siragy and Nantel, 2020).

Step length was calculated for each step as the hypotenuse of the vertical and anteroposterior distance between the feet at heel strike of the leading leg. The MoS was calculated bilaterally at both heel strikes and defined as the distance of the Extrapolated Center of Mass (xCoM) to the right/left lateral heel marker:

$$\mathbf{MoS = Lateral\ heel\ marker - xCoM} \tag{1}$$

The formula for xCoM was:

$$\mathbf{xCoM = CoM_p} + \left(\frac{\mathbf{CoM_v}}{\omega\Theta}\right) \tag{2}$$

Where CoMp = CoM's position, CoMv = CoM's velocity. $\omega\Theta$ was calculated as:

$$\omega\Theta = \sqrt{\frac{\mathbf{g}}{\mathbf{l}}} \tag{3}$$

In this term, $g = 9.81\ \text{m/s}^2$ and l is the length of the inverted pendulum determined as the average distance of the right/left lateral heel marker to the CoM at heel-strikes. Visual 3D was used to calculate the CoM's position and velocity.

Kinematic measures included trunk angle (mid-point of the posterior superior iliac spine markers to C7 compared to global vertical, measured in the AP direction with a larger trunk angle indicating increased forward inclination) and trunk acceleration root-mean-square (RMS) in the ML, AP, and VT directions as a measure of upper body variability [with larger RMS values indicating greater variability (Menz et al., 2003; Marigold and Patla, 2008)]. All data reduction prior to statistical analyses were performed using the Julia programming language (Bezanson et al., 2017) and custom code (MacDonald et al., 2021).

FIGURE 1 | The CAREN-Extended virtual reality system used in this study.

Statistical Analyses

Separate 2-way repeated measures ANOVAs were used to examine significance between each slope (uphill, downhill) compared to level and across arm (held, normal, active) conditions, as well as potential interactions, for all variables using IBM SPSS Statistics 26 (IBM Analytics, Armonk, USA). Assumption of normality was confirmed using a Shapiro-Wilk test and Greenhouse-Geisser p was reported when Mauchly's Test of Sphericity was violated. Significance level was set at $p < 0.05$. A Bonferroni correction was used for *post-hoc* tests.

FIGURE 2 | Platform angles throughout the rolling-hills terrain and representative sample of heel-strike gait events and average step angles included in sloped conditions. Average step angles within the shaded region were not counted toward uphill or downhill steps. NB: Figure depicts the AP platform angle throughout the terrain condition, not the elevation of the virtual path.

TABLE 1 | Main effects for uphill vs. level walking.

Uphill vs. Level		Arms			Slope			Arms × slope		
Variable		***F*(2, 28)**	***p***	η_p^2	***F*(1, 14)**	***p***	η_p^2	***F*(2, 28)**	***p***	η_p^2
Speed		7.59	**0.007**	0.352	6.73	**0.021**	0.325	0.76	0.478	0.051
Step length	mean	49.32	**<0.001**	0.779	40.50	**<0.001**	0.743	0.85	0.439	0.057
	CoV	6.27	**0.006**	0.309	28.95	**<0.001**	0.674	5.23	**0.012**	0.272
Step width	mean	1.17	0.326	0.077	47.80	**<0.001**	0.773	0.02	0.983	0.001
	CoV	5.53	**0.009**	0.283	3.97	0.066	0.221	0.05	0.956	0.003
Step time	mean	15.02	**<0.001**	0.518	6.32	**0.025**	0.311	3.85	**0.033**	0.216
	CoV	1.08	0.355	0.071	23.90	**<0.001**	0.631	1.54	0.232	0.099
DST	mean	14.34	**<0.001**	0.506	30.91	**<0.001**	0.688	0.00	0.998	0.000
	CoV	0.76	0.475	0.052	3.90	0.068	0.218	0.73	0.490	0.050
ML MOS	mean	14.34	**<0.001**	0.506	27.64	**<0.001**	0.664	0.46	0.636	0.032
	CoV	1.22	0.310	0.080	6.45	**0.024**	0.315	0.01	0.986	0.001
Trunk angle		20.63	**<0.001**	0.596	28.02	**<0.001**	0.667	0.60	0.559	0.041
RMS	AP	56.33	**<0.001**	0.801	9.20	**0.009**	0.396	1.35	0.276	0.088
	ML	1.67	0.206	0.107	0.08	0.780	0.006	0.31	0.734	0.022
	VT	4.30	**0.040**	0.235	0.03	0.858	0.002	0.28	0.759	0.020

Boldfaced numbers indicate significant main effect.

TABLE 2 | Main effects for uphill vs. level walking.

Downhill vs. Level		Arms			Slope			Arms × slope		
Variable		***F*(2, 28)**	***p***	η_p^2	***F*(1, 14)**	***p***	η_p^2	***F*(2, 28)**	***p***	η_p^2
Speed		7.42	**0.008**	0.346	11.1	**0.005**	0.442	0.49	0.619	0.034
Step length	mean	51.96	**<0.001**	0.788	5.0	**0.042**	0.264	2.03	0.150	0.127
	CoV	2.65	0.088	0.159	15.2	**0.002**	0.521	1.80	0.183	0.114
Step width	mean	1.42	0.259	0.092	0.0	0.992	0.000	0.03	0.970	0.002
	CoV	5.65	**0.009**	0.287	2.5	0.140	0.149	0.26	0.777	0.018
Step time	mean	15.47	**<0.001**	0.525	5.4	**0.036**	0.279	2.05	0.148	0.128
	CoV	0.13	0.881	0.009	27.1	**<0.001**	0.660	0.52	0.598	0.036
DST	mean	13.76	**<0.001**	0.496	1.6	0.227	0.102	0.03	0.970	0.002
	CoV	0.94	0.404	0.063	13.3	**0.003**	0.488	0.79	0.462	0.054
ML MOS	mean	5.63	**0.009**	0.287	3.1	0.098	0.183	0.06	0.941	0.004
	CoV	0.52	0.602	0.036	4.4	0.055	0.239	0.57	0.575	0.039
Trunk angle		18.76	**<0.001**	0.573	7.5	**0.016**	0.349	0.28	0.755	0.020
RMS	AP	49.93	**<0.001**	0.781	4.2	0.060	0.230	2.12	0.139	0.131
	ML	0.04	0.892	0.003	1.4	0.252	0.092	0.69	0.510	0.047
	VT	3.78	0.055	0.213	0.1	0.780	0.006	0.03	0.975	0.002

Boldfaced numbers indicate significant main effect.

RESULTS

No significant interaction effects between arm swing and surface slope were found. Statistical information regarding main effects is included in **Tables 1**, **2**, with significant *post-hoc* findings presented in the following text. See **Tables 3**, **4** for spatiotemporal results and **Table 5** for postural kinematics. Tables including the number and average angle of steps analyzed are included in **Supplementary Materials 1**, **2**, with correlations analyses regarding margin of stability in **Supplementary Materials 3**, **4**.

Arm Swing During Uphill and Downhill Sections of the Rolling-Hills

In this section, corrected p-values for each result are presented in parentheses.

Walking with arms held decreased walking speed compared to normal ($p \leq 0.044$) and active arm swing ($p \leq 0.031$). Step length increased with increasing arm swing ($p \leq 0.01$) and, during uphill sections only, step length CoV was greater when walking with arms held compared to with normal arm swing ($p = 0.027$). Active arm swing increased step width CoV compared to normal

TABLE 3 | Comparison of speeds, spatiotemporal gait parameters, and coefficients of variation (CoV) in the three arm swing conditions (held, normal, active) during uphill, level, and downhill walking.

Slope	Arms	Speed (m/s)	Spatiotemporal				CoV (%)			
			Step length (cm)	Step width (cm)	Step time (s)	DST (% stride)	Step length	Step width	Step time	DST
Downhill	Held	**1.13 (0.19)**	**52.6 (7.47)**[a]	20.7 (4.34)	**0.50 (0.05)**[a]	**30.9 (3.84)**[a]	7.83 (3.93)	8.67 (4.12)	5.45 (1.48)	11.3 (2.39)
	Normal	**1.26 (0.21)**[b]	**59.0 (8.48)**[a,b]	19.9 (3.44)	**0.49 (0.03)**[a]	**28.8 (4.37)**[a]	5.92 (2.53)	**10.3 (5.37)**[a]	5.56 (1.82)	11.2 (3.41)
	Active	**1.29 (0.24)**[b]	**67.0 (9.41)**[b]	20.7 (4.56)	**0.55 (0.04)**	**27.5 (4.28)**	5.13 (2.23)	**13.8 (7.51)**	5.18 (1.30)	10.6 (3.44)
Downhill vs. Level		**0.005***	**0.042***	0.992	**0.036***	0.227	**0.002***	0.140	**<0.001***	**0.003***
Level	Held	1.23 (0.19)	56.1 (5.24)	20.8 (4.26)	0.51 (0.04)	31.1 (4.06)	5.26 (2.46)	11.0 (7.40)	3.94 (2.95)	8.84 (3.69)
	Normal	1.33 (0.17)	61.6 (6.27)	19.8 (4.22)	0.51 (0.04)	30.1 (3.40)	4.01 (2.21)	10.9 (4.42)	3.24 (1.61)	7.00 (2.47)
	Active	1.38 (0.18)	68.2 (5.50)	20.7 (4.06)	0.55 (0.04)	27.7 (3.15)	4.38 (3.41)	16.0 (7.25)	3.81 (2.71)	8.49 (3.54)
Uphill vs. Level		**0.021***	**< 0.001***	**< 0.001***	**0.025***	**< 0.001***	**< 0.001***	0.066	**< 0.001***	0.068
Uphill	Held	**1.13 (0.19)**	**48.8 (6.74)**[a]	23.2 (4.86)	**0.51 (0.06)**[a]	**33.6 (4.01)**[a]	**12.2 (6.77)**	8.21 (3.52)	7.07 (3.41)	9.49 (3.27)
	Normal	**1.28 (0.22)**[b]	**55.4 (6.60)** [a,b]	22.3 (4.33)	**0.52 (0.04)**[a]	**32.7 (3.95)**[a]	**7.91 (2.55)**[b]	**8.90 (4.42)**[a]	6.41 (2.73)	9.54 (1.99)
	Active	**1.30 (0.26)**[b]	**62.0 (8.52)**[b]	23.0 (5.80)	**0.58 (0.06)**	**30.2 (4.03)**	6.87 (2.56)	**13.7 (9.91)**	5.11 (2.16)	9.34 (4.29)

Data within each slope are represented as the mean values averaged for all 15 participants (8 male, 7 female), mean (standard deviation). Pairwise comparison p-values of slope conditions (Uphill vs. Level and Downhill vs. Level) from two-way repeated measures ANOVA are presented between surface conditions. Statistical significance set at $p < 0.05$ with Bonferroni correction.

Boldfaced numbers highlight significant differences with the following specifications.

[a]Different from Active.

[b]Different from Held.

**Different from Level.*

($p \leq 0.047$). Active arm swing increased step time compared to held ($p = 0.005$) and normal ($p = 0.001$). Active arm swing also decreased double support time compared to held ($p \leq 0.001$) and normal ($p \leq 0.014$). During downhill sections only, active arm swing increased ML-MoS compared to normal ($p = 0.014$).

Active arm swing decreased trunk angle compared to held ($p < 0.001$) and normal ($p \leq 0.006$). AP-RMS magnitude was larger with active arm swing compared to held and normal ($p < 0.001$) and smaller with arms held compared to normal ($p \leq 0.003$). During uphill sections only, main effects were found for VT-RMS but no *post-hoc* significance.

Uphill vs. Level

Walking on uphill sections decreased walking speed and step length and increased step width, step time, and double support time compared to level. Uphill walking also increased ML-MoS compared to level. Uphill walking increased step time CoV and decreased step length and ML-MoS CoV. Uphill walking increased trunk angle, and decreased AP-RMS magnitude compared to level.

Downhill vs. Level

Walking on downhill sections decreased walking speed, step length, and step time compared to level. Downhill walking decreased step length CoV and increased step time CoV and double support time CoV. Downhill walking decreased trunk angle compared to level.

TABLE 4 | Comparison of mediolateral margin of stability and coefficient of variability in the three arm swing conditions during uphill, level, and downhill walking.

Slope	Arms	ML-MoS (cm)	CoV ML-MoS (%)
Downhill	Held	10.9 (3.22)	31.2 (14.7)
	Normal	**10.6 (2.41)**[a]	36.1 (14.6)
	Active	**11.8 (2.77)**	33.4 (14.4)
Downhill vs. Level		0.098	0.055
Level	Held	10.1 (3.60)	36.6 (16.5)
	Normal	9.86 (4.15)	38.7 (23.3)
	Active	11.4 (3.60)	42.8 (19.4)
Uphill vs. Level		**< 0.001***	**0.024***
Uphill	Held	12.6 (2.70)	28.1 (7.82)
	Normal	12.6 (2.22)	29.7 (14.4)
	Active	13.0 (3.02)	35.2 (29.0)

Data within each slope are represented as the mean values averaged for all 15 participants (8 male, 7 female), mean (standard deviation). Pairwise comparison p-values of slope conditions (Uphill vs. Level and Downhill vs. Level) from two-way repeated measures ANOVA are presented between surface conditions. Statistical significance set at $p < 0.05$ with Bonferroni correction.

Boldfaced numbers highlight significant differences with the following specifications.

[a]Different from Active.

**Different from Level.*

DISCUSSION

This study investigated the effect of various arm swings on spatiotemporal parameters and postural strategies during uphill and downhill sections of a rolling-hills terrain compared to level walking. Regardless of slope, active arm swing increased step time and decreased double-support and trunk angle, while walking with arms held decreased walking speed and trunk angle. During both uphill and downhill sections, walking speed was consistently slower and caused postural and spatiotemporal changes from

TABLE 5 | Comparison of kinematic postural variables in the three arm swing conditions during uphill, level, and downhill walking in the anteroposterior (AP), vertical (VT), and mediolateral (ML) directions; Data within each slope are represented as the mean values averaged for all 15 participants (8 male, 7 female), mean (standard deviation).

Slope	Arms	AP		VT	ML
		Trunk angle (°)	RMS	RMS	RMS
Downhill	Held	**7.38 (4.01)[a]**	**1.18 (0.30)[a]**	2.28 (0.59)	1.21 (0.46)
	Normal	**6.65 (4.06)[a]**	**1.54 (0.40)[a,b]**	2.55 (0.61)	1.17 (0.44)
	Active	**4.31 (4.20)**	**2.03 (0.41)[b]**	2.66 (0.84)	1.18 (0.30)
Downhill vs. Level		**0.016***	0.060	0.780	0.252
Level	Held	8.00 (3.78)	1.16 (0.26)	2.29 (0.50)	1.08 (0.32)
	Normal	7.49 (3.90)	1.46 (0.34)	2.59 (0.62)	1.08 (0.44)
	Active	5.43 (3.53)	1.87 (0.35)	2.67 (0.69)	1.17 (0.31)
Uphill vs. Level		**< 0.001***	**0.009***	0.858	0.780
Uphill	Held	**9.54 (3.92)[a]**	**1.02 (0.16)[a]**	2.25 (0.68)	1.10 (0.24)
	Normal	**9.14 (3.93)[a]**	**1.31 (0.27)[a,b]**	2.62 (0.66)	1.13 (0.36)
	Active	**6.36 (4.30)**	**1.83 (0.36)[b]**	2.73 (0.99)	1.25 (0.51)

Pairwise comparison p-values of slope conditions (Uphill vs. Level and Downhill vs. Level) from two-way repeated measures ANOVA are presented between surface conditions. Statistical significance set at $p < 0.05$ with Bonferroni correction.
Boldfaced numbers highlight significant differences with the following specifications.
[a]Different from Active.
[b]Different from Held.
**Different from Level.*

level walking despite the slopes being mild. Within downhill sections, active arm swing corresponded to increased ML-MoS compared to normal. Compared to level walking, uphill sections increased step width and ML-MoS.

Variability of Rolling-Hills Condition Required Proactive Base of Support Changes

When walking on the rolling-hills terrain, the magnitude and timing of surface fluctuations was unpredictable (oscillating between $-3°$ and $+3°$) and required participants to navigate continuous changes in surface slope. For example, a posterior tilt in the surface shifting to an incline may interfere with a leg in late swing and precipitate unplanned foot contact, and an anterior tilt to a decline may induce a stepping response to catch balance. Prentice et al. (2004) investigated walking from a level surface onto a ramp and found that even the smallest incline (3°) required adaptations to the swing limb trajectory (Prentice et al., 2004). We believe that the increased step time CoV found in our study could be the result of a similar proactive strategy to optimize the base of support during the rolling-hills terrain. Using the rolling-hills terrain condition, Sinitski et al. (2019) similarly found that healthy adults increased step time variability as well as step length variability compared to level walking (Sinitski et al., 2019). They also reported that participants increased step width during the rolling-hills condition compared to level walking. While they only investigated the rolling-hills as a single walking condition, we found increased step width to be specific to the uphill sections. However, the steps counted within the uphill and downhill sections can each be considered a transition step which reflect characteristics of both the current state as well as the upcoming state (Gottschall and Nichols, 2011). Therefore, it remains uncertain whether the increased step width is attributable to the current uphill section or in preparation for the upcoming downhill section. In either case, participants proactively modified their base of support to stabilize the COM when navigating the rolling-hills terrain.

The increased step width and double support time during uphill sections coincided with increased ML-MoS and decreased ML-MoS CoV. Vieira et al. (2017) similarly found increased ML-MoS during uphill sections, which increased stability, but their results showed decreased ML-MoS during downhill sections which we did not find (Vieira et al., 2017). Our results are somewhat different from Kawamura and Tokuhiro (1991) who found no step width increase during uphill sections (Kawamura and Tokuhiro, 1991). However, Kawamura's study examined a relatively narrow ramp which may have affected participants' ability to increase step width. The decrease we found in ML-MoS CoV may also be linked to uphill steps being consistently wider compared to level walking. In healthy individuals, decreased step width variability is thought to reflect greater active attention toward foot placement (Maki, 1997; Siragy and Nantel, 2018; Siragy et al., 2020). Additionally, increases in ML-MoS during perturbations may indicate a compensation response to mitigate destabilizing effects of the terrain, particularly as this finding was unique to the present study compared to previous investigations of ML-MoS during both uphill and downhill walking (Vieira et al., 2017). This demonstrates that the healthy young adults did adjust to the incline, even though the slope was minor, and successfully maintained stability.

Mild Uphill and Downhill Slopes Required Spatiotemporal and Postural Modifications

Speed was slower for both uphill and downhill sections compared to level. This is somewhat similar to Kawamura and Tokuhiro (1991) which found a decrease in walking speed for both uphill and downhill conditions at 12°, but not at lower slopes (3, 6, 9°) (Kawamura and Tokuhiro, 1991). Our finding of decreased walking speed with slopes ranging from −3 to +3° may, therefore, be linked to the continuously varying nature of the rolling-hills terrain condition wherein a more cautious gait was employed for the duration of the terrain. Trunk posture was more backward during downhill sections and more forward during uphill sections, as hypothesized. Uphill walking is typically accompanied by a forward inclination of the trunk to aid in forward propulsion and stepping up (Leroux et al., 2002). Conversely, downhill walking is typically accompanied by a less forward trunk posture which assists in stepping down and the frictional demands on downhill slope (Leroux et al., 2002). The decreased walking speed and altered spatiotemporal and postural variables demonstrate that participants did make accommodations for the mild ($\leq$ 3°) slopes encountered. Therefore, participants navigated the rolling-hills primarily by decreasing walking speed, but even the mild slopes caused spatiotemporal and postural changes.

Active Arm Swing Required Proactive Strategies to Increase ML-MoS During Downhill Walking

We hypothesized that active arm swing may additionally perturb gait and require strategies that interact with those adopted for sloped walking. Instead, we found that the gait strategies used to manage active arm swing remained relatively consistent across slope conditions. However, the increase in ML-MoS seen with active arm swing compared to normal was only observed during downhill walking and corresponded to increased step width CoV. Hill and Nantel (2019) also found increased step width variability with active arm swing compared to normal during level walking (Hill and Nantel, 2019). They postulated that the more variable step width stemmed from the decreased coordination also found in the active arm swing condition and may have contributed to the concomitant increase in trunk local dynamic stability. The higher step width variability may demonstrate a proactive strategy to help stabilize the COM when walking with active arm swing, which was successful so far as to also increase ML-MoS in the downhill walking condition. This potentially shows that participants improved their mediolateral stability by varying their step width when managing the active arm swing.

Arm Swing Effects Were Consistent Across Uphill and Downhill Sections of Rolling-Hills

We hypothesized that walking with arms held would lead to compound compensatory strategies during both uphill and downhill sections of the rolling-hills to increase stability. In both uphill and downhill sections, walking with arms held decreased speed compared to normal and active, which may indicate an extra level of caution when walking without arm swing. However, this did not appear to alter any strategies adopted during sloped walking. In fact, spatiotemporal differences from arm swing primarily existed with active arm swing compared to held and normal, with no significant differences between held and normal. For example, compared to held and normal, active arm swing increased step time, seemingly to preserve the coupling of arm-to-leg swing when the arms had further to swing (Bondi et al., 2017). This is further evidenced by the concomitant increase in step length during the active arm swing condition. It may be the case that the speed adjustment made by participants during the held condition was adequate to approximate normal walking stability and limit further need for spatiotemporal adjustments. Conversely, walking speed during active arm swing was not significantly different from normal but led to significant spatiotemporal differences from normal arm swing. Compared to normal arm swing, both held and active conditions caused distinct postural differences. Adopting a larger trunk angle with arms held projects the CoM further anteriorly, potentially reflecting an attempt to facilitate forward progression (Leroux et al., 2002). In contrast, the more upright posture (smaller trunk angle) during active arm swing may be an attempt to compensate for the forward-shifted CoM from increased anterior arm swing. While held and active arm swing illicited different strategies, these strategies remained separate from those used to navigate the slopes.

Limitations

Both the “held” and “active” arm swing conditions could have led to increased attention compared to normal arm swing, which may approach the attentional requirements of some dual tasks. It is uncertain to what extent this affects the outcome parameters. The rolling-hills was a continuous slope condition wherein a range of angles were used rather than specific slope angles. While this is a more naturalistic terrain, it cannot provide insight to the strategies used to overcome specific surface angles or the extent of the spatiotemporal or postural strategies.

CONCLUSION

Our study demonstrates that arm swing caused equivalent changes in all surface conditions. ML-MoS and step width CoV both increased within downhill sections of the rolling-hills terrain with the use of active arm swing compared to normal. This indicates that young, healthy participants may have improved their mediolateral stability by varying their step width when managing the active arm swing. Alternately, the increase in ML-MoS during uphill sections compared to level was accompanied by wider steps and longer double support time. Because stability increased during active arm swing with ongoing base of support adjustments and during sloped walking with consistently wider steps and longer double support, this demonstrates that different stepping strategies were used to manage active arm swing compared to a mild incline. Participants successfully navigated the rolling-hills by decreasing walking speed, but even the mild slopes caused spatiotemporal and postural changes. Specifically, the variability of the rolling-hills required participants to proactively modify their base of support to stabilize the COM. As this study tested healthy young adults, the current findings can be used as a baseline comparison in future investigations of other populations. Future research should focus on sloped walking in populations at risks of or with gait impairments (i.e., older adults or those with gait disorders).

DATA AVAILABILITY STATEMENT

The software and dataset produced and analyzed during this work are openly available in Zenodo at: https://doi.org/10.5281/zenodo.5608535.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the Institutional Review Board of the University of Ottawa and the Ottawa Hospital Research Ethics Board. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

JN: conceptualized and organized the research project. M-EM: data analysis—main analysis. TS and AH: data analysis—secondary analysis. JN, TS, and AH: data analysis—review and critique. M-EM, TS, and JN: statistical analysis—design. M-EM: statistical analysis—execution. JN, TS, and AH: statistical analysis—review and critique. M-EM: manuscript—writing of the first draft. JN, TS, and AH: manuscript—review and critique. All authors contributed to the article and approved the submitted version.

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Effect of Torso and Breast Characteristics on the Perceived Fit of Body Armour Systems Among Female Soldiers: Implications for Body Armour Sizing and Design

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This study aimed to provide normative data characterising the torsos and breasts of female soldiers and to determine which torso and breast anthropometric measurements contributed to reports of poor body armour fit. Ninety-seven female Australian Army soldiers completed a questionnaire about their experience with current-issue body armour, including perceptions of fit. Participants also attended a single testing session where we took a three-dimensional scan of their breasts and torso and collected several anthropometric measurements to characterise their torso size and shape. Sixteen of the 22 breast and torso measurements collected were significantly related to the perceived fit of current-issue body armour systems. To improve perceptions of fit for female soldiers and, in turn, reduce movement interference, discomfort, and barriers to occupational performance, future body armour systems should cater to the wide range of female breast and torso shapes and sizes.

Keywords: anthropometry, equipment design, protective equipment, female soldier, body armour

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INTRODUCTION

Military body armour functions to protect the vital thoracoabdominal organs of soldiers from ballistic, fragmentation and stab threats (Choi et al., 2016; Laing and Jaffrey, 2019). Body armour, however, can introduce integration issues between the human user and the system, as well as present a mass burden (Knapik et al., 2004). These integration issues are amplified if the body armour system is ill-fitting. Soldiers wear body armour during operational deployment and training or field-based exercises for 5.2–6.4 ± 4.7 h per day (Coltman et al., 2020). It is therefore vital that a body armour system (comprising front and rear hard and soft ballistic plates encased in a carrier) interfaces appropriately with a soldier's torso (i.e., fits correctly) and integrates well with other elements of a soldier's combat ensemble (Furnell et al., 2017). Proper fit will maximise protection and maintain coverage requirements over this long duration of wear (Laing and Jaffrey, 2019). The fit of body armour also has important implications for job and combat performance. That is, poor-fitting body armour can impede a soldier's ability to run efficiently, to shoulder and fire a rifle, to manoeuvre in and out of a vehicle and to complete other mission essential tasks (Mitchell et al., 2010; Choi et al., 2016, 2018; Coltman et al., 2020, 2021a). Consequently, designing and sizing body armour to properly fit all soldiers is vitally important for military organisations to achieve correctly.

Previous research has revealed that most female soldiers are dissatisfied with the fit of their current body armour systems (Epstein et al., 2013; Toma et al., 2016; Coltman et al., 2020, 2021a; Davis et al., 2020). Specifically, questionnaire data related to the fit of body armour was completed by 147 female Australian Defence Force (ADF) soldiers in both combat and non-combat roles. Of the women surveyed, 68, 56, and 12%, respectively, found the body armour to be ill-fitting, too large, and too small (Coltman et al., 2020). Similarly, researchers investigating the fit and function of body armour in the United Kingdom found that female soldiers reported numerous instances of discomfort, particularly at the hip (Davis et al., 2020). In the same study 29–59% of participants experienced task interference during a range of basic job roles (Davis et al., 2020). Another group found that integration between a soldier's bra and body armour was a source of substantial discomfort (Malbon et al., 2020). The notion of sex-specific discomfort with body armour was supported by findings from an Australian study, which revealed that limited adjustability of the system, insufficient space for breasts and oversized length and width were all common problems reported by female soldiers (Coltman et al., 2021a). Female soldiers now comprise 10.4–17.5% of Defence Force populations in Australia, Canada, New Zealand, United Kingdom and United States of America (Defence People Group., 2017; Ministry of Defence., 2018; Mark, 2019; Service Women's Action Network., 2019; Government of Canada., 2020). Ensuring that body armour is appropriately sized and designed to enable female users to undertake their occupational roles is essential to fostering women's participation and inclusion in Defence.

A primary source of poor body armour fit for female soldiers is how it is sized and designed (Epstein et al., 2013; Toma et al., 2016; Coltman et al., 2020, 2021a; Malbon et al., 2020). Although several anthropometric surveys have been conducted on global soldier populations to inform the design and sizing of protective equipment, military body armour is commonly issued to soldiers in a limited, unisex sizing range (Todd, 2007). Moreover, the dimensions and specifications of body armour systems are traditionally based upon male anthropometric data (Todd, 2007; Epstein et al., 2013; Toma et al., 2016). This is likely a function of body armour primarily being designed for combat soldiers. Until 2011, women in the ADF were ineligible to enlist in combat coded roles (Wadham et al., 2018) and therefore only men served in combat. Furthermore, women comprise only a small percentage of the soldier population in anthropometric surveys and there is a lack of breast specific measurements included in such surveys to better inform body armour designs (Edwards et al., 2014). Given that women have different torso shapes and more developed breast tissue compared to men (Edwards et al., 2014; Gordon et al., 2014), designs that have not specifically accounted for female anthropometry are unlikely to properly fit female soldiers. Poor personal protective equipment (PPE) fit has also been observed among female workers in a range of historically male-dominated occupations such as firefighting and construction (Barker et al., 2012; Onyebeke et al., 2016). Evidence also highlights the sex disparity in PPE design affecting occupational task performance (Park and Hahn, 2014). No previous research, however, has specifically investigated which breast and torso characteristics contribute to poor fit of body armour systems for female soldiers. Such information is required to understand the association between anthropometric torso dimensions and equipment fit in body armour design and sizing. This exploratory study, therefore, aimed to: (i) provide normative data characterising the torso and breast size and shape of female soldiers to inform evidence-based design modifications to current-issue body armour, and (ii) determine which breast characteristics and torso anthropometric measurements contributed most to poor body armour fit reported by female soldiers. It was hypothesised that female soldiers would display a wide range of breast and torso characteristics and that these characteristics would differ between soldiers who reported that their body armour was too small, too large or a good fit.

TABLE 1 | Participant characteristics (n = 97).

Characteristic	Participant data
Age (mean ± SD)	25.6 ± 7.4 years
BMI (mean ± SD)	25.4 ± 3.3 kg/m^2
Underweight (<18.5 kg/m^2)	0% (n = 0)
Healthy weight (18.5–24.9 kg/m^2)	54% (n = 52)
Overweight (25–29.9 kg/m^2)	38% (n = 37)
Obese (>30 kg/m^2)	8% (n = 8)
Bra band size (mode; range)	10 (8–16)
Bra cup size (mode; range)	C (A–H)
Combat-arms employment categories	23% (n = 22)
Non-combat arms employment categories	77% (n = 75)
Years in Army (mean ± SD)	4.3 ± 4.8 years
TBAS V4.4 Tier 2 users	36% (n = 35)
TBAS V4.4 Tier 3 users	64% (n = 62)
Body armour wear duration during training (mean ± SD)	6.4 ± 4.7 h/day
Body armour wear duration during operations (mean ± SD)	5.2 ± 4.7 h/day

MATERIALS AND METHODS

Participants

Ninety-seven female soldiers (**Table 1**) from various units within the Australian Army volunteered to participate in the present study. Potential participants were excluded if they had epilepsy that might be induced by the flashing light of the three-dimensional scanner (described below). All participants had experience wearing ADF issue body armour [Tiered Body Armour System (TBAS) V4.4 Tier 2 or Tier 3; **Figure 1**] and provided written informed consent before being assigned to attend a single test session, scheduled either in May or June 2019. Ethical approval for the study was obtained from the Defence Science and Technology Group Low Risk Ethics Panel (LD 07-351) and cross-institutional approval was obtained from

FIGURE 1 | Current Australian body armour is the Tiered Body Armour System (TBAS) V4.4, issued to personnel in either Tier 2 **(A)** or Tier 3 **(B)**. The same size hard plate and soft armour inserts are used in both systems. Tier 3 affords slightly greater protection to the solider, as shown by increased soft armour coverage on the side wings.

the University of Canberra Human Research Ethics Committee (Project ID: 2013).

Torso Characteristics

To characterise each participant's torso dimensions, a series of anthropometric measurements (described in **Table 2**) were collected according to procedures outlined by the International Standards for Anthropometric Assessment (Marfell-Jones et al., 2012). Measurements were chosen based on their relevance to the design of military body armour. This included anthropometric dimensions specifically recommended by the Australian Warfighter Anthropometry Survey (AWAS) as being pertinent to body armour (Edwards et al., 2014) and several additional measurements from AWAS deemed potentially relevant by subject matter experts within the Australian Army. Data were collected while participants wore long pants and a bra only. Each measurement was repeated three times, with the mean of the three measurements recorded in centimetres (cm).

Breast Characteristics

A three-dimensional scanning protocol was used to characterise each participant's breasts. Adhesive markers (~1 cm in diameter) were placed on the participant's bra and breast to outline the boundaries of each breast (**Table 3A**). The participant was instructed to stand up straight and look forward with the heel of her hands resting on her hips. The breasts and torso of each participant were then scanned using a hand-held three-dimensional scanner (Artec™ Eva 3D Scanner, Artec Group, San Jose, USA) while she was wearing a standardised encapsulation-type bra (New Legend Underwire sports bra, Berlei, Wentworthville, NSW, Australia). The test bra was professionally fitted (McGhee and Steele, 2010) and chosen for its ability to separate the breasts and enable clear visualisation of a defined breast border. From this scan, a three-dimensional model of the breast was created (**Tables 3E,F**). Although three-dimensional breast characteristics are usually quantified while women are bare-breasted (Coltman et al., 2018, 2021b; McGhee and Steele, 2019), ethical limitations necessitated a modified protocol in the present study. As female soldiers wear a bra when using body armour systems, it was deemed appropriate to calculate breast characteristics while participants wore a standardised bra. The standardised bra was chosen for its ability to separate a participant's breasts without drastic compression so that all anatomical landmarks required to calculate the breast measurements were clearly visible. Importantly, the standardised bra was worn only during scanning and therefore did not have any effect on the participants' ratings of perceived body armour fit, which was answered in relation to whichever bra style and size they normally wore. This scanning protocol is also consistent with recommendations for breast measurements taken to inform the design of protective equipment worn external to a bra (Brisbine et al., 2020a) and previous research undertaken by the research team (Coltman et al., 2021b).

Eleven breast characteristics (including eight breast size and three breast position measurements; described in **Table 3**) were calculated from the scanned images using Geomagic Studio® software (Version 12; 3DSystems, South Carolina, USA). These measurements were selected because they were deemed relevant to the design of military body armour based on recommendations for other torso-borne female equipment items, including the design of sports bras (McGhee and Steele, 2011; Zhou et al., 2013; Coltman et al., 2015, 2017a) and breast protective equipment (Brisbine et al., 2020a). Linear anthropometric measurements (such as the Sternal Notch to Nipple Distance) and three-dimensional measurements (such as Breast Volume and Surface Area) have previously been found to be accurate and valid when derived from three-dimensional scans (Paquette et al., 2000; Losken et al., 2005; Han et al., 2010; Qi et al., 2011; Yip et al., 2012). All calculations were completed by one researcher [BRB], who had high reliability in deriving these measurements (all ICC > 0.982; $p < 0.001$). Details of how each measurement was calculated are provided in **Table 3**.

Perceived Body Armour Fit

Participants were asked to rank the overall fit of their current-issue body armour system (TBAS V4.4 Tier 2 or Tier 3) on a 5-point scale: 1 = *way too small/short/tight*; 2 = *too small/short/tight*; 3 = *good fit*; 4 = *too large/long/loose*; and 5 = *way too large/long/loose.* The responses *way too small/short/tight* and *too small/short/tight* were combined to form the category "too small" and responses *too large/long/loose* and *way too large/long/loose* were combined to form the category "too large" to create a categorical dependent variable with three groups. This question was from a larger 59-item questionnaire designed to explore the fit and function of current issue ADF body armour, which is described in more detail by Coltman et al. (2020). In brief, the 59-item questionnaire was developed following a focus group conducted with female soldiers ($n = 8$) and in consultation with subject matter experts within the Australian Army. The face validity and readability of the questionnaire were tested with personnel within the Defence Science and Technology Group and Australian Army. The questionnaire was published on a University of Canberra Qualtrics account (v0217; Provo, UT).

TABLE 2 | The anthropometric measurements collected during the current study, including a brief description of the protocol associated with each measurement.

Measurement	Description of how the measurement was taken
Stature (cm)*	The participant stood against a portable stadiometer (model: 213, Seca Corp., Maryland, USA) with her feet together and heels against the back of the stadiometer. With her head in the Frankfort plane, the participant was instructed to take a deep breath in while the researcher applied a gentle lift through the mastoid processes and then placed the headboard of the stadiometer firmly down on the participant's vertex.
Body Mass (kg)*	Body mass (recorded to the nearest 0.1 kg) was measured while each participant stood on calibrated body mass scales (model: RD 545, Tanita, Illinois, USA) without wearing shoes and socks.
Iliocristale Height (cm)*	The most lateral edge of the iliac crest on the participant's left ilium was palpated and marked. The participant was then instructed to stand up straight while the vertical distance between the floor (standing surface) and her iliocristale marking was measured using an anthropometer (Siber-Hegner, Zurich, Switzerland).
Waist Height (cm)*	The participant's waist was palpated and marked at the level of the narrowest point between her lower costal (10th rib) border and the iliac crest on the left side of her body. The participant was then instructed to stand up straight while the vertical distance between the floor (standing surface) and her waist marking was measured using an anthropometer.
Suprasternale Height (cm)*	The participant's suprasternal notch was palpated and marked. The participant was instructed to stand up straight and the vertical distance between the floor (standing surface) and her suprasternal marking was measured using an anthropometer.
Front Length (cm)*	The suprasternale height measurement was subtracted from the iliocristale height measurement to calculate front length.
Chest Depth (cm)*	With the participant in a standing position, a sliding caliper (Campbell 20, Rosscraft International, British Columbia, Canada) was placed at the level of her mesosternale (anteriorly) and on the spinous process of her vertebra (posteriorly) at the horizontal level of the mesosternale. The participant was instructed to breathe normally and the measurement was taken at the end of tidal expiration.
Chest Breadth (cm)*	The participant assumed a relaxed standing position with her arms abducted. The sliding caliper was positioned at the level of the mesosternale (anteriorly) and the distance was recorded between the most lateral aspect of the thorax at the end of tidal expiration.
Bi-acromial Breadth (cm)*	The participant assumed a standing position with her arms hanging by her sides while the distance between the most lateral points of her acromion processes was measured.
Neck Circumference (cm)	The participant assumed a relaxed standing position with her arms hanging by her sides and a tape measure (W606PM, Lufkin, United States) was applied around the base of her neck.
Waist Circumference (cm)*	The participant assumed a standing position with her arms folded across her thorax. The circumference was measured at the level of the narrowest point between the 10th rib and iliac crest at the end of normal expiration.
Hip Circumference (cm)	The participant assumed a relaxed standing position with her arms folded across her thorax. The circumference was taken at the greatest posterior protuberance of her buttocks.
Over-Bust Chest Circumference (OBCC) (cm)*	The participant assumed a standing position with her arms hanging by her sides and slightly abducted. The girth was taken at the level of the widest point of her bust at the end of a normal expiration.
Under-Bust Chest Circumference (UBCC) (cm)*	The participant assumed a standing position with her arms hanging by her sides and slightly abducted. The girth was taken at the level directly below her bust at the end of a normal expiration.

indicates measurements that were deemed relevant to the design of body armour in the Australian Warfighter Anthropometry Survey (Edwards et al., 2014).

Statistical Analyses

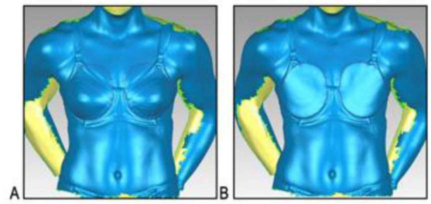
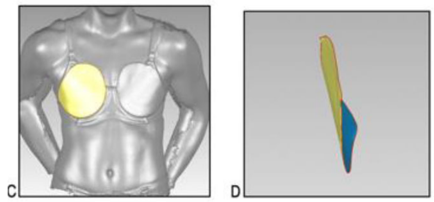
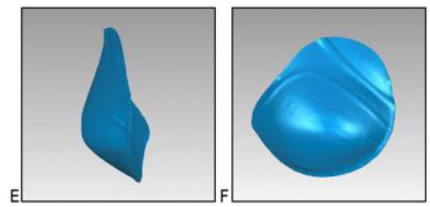
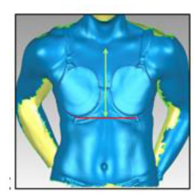
Descriptive statistics were calculated for all torso and breast characteristics listed in **Tables 2**, **3**, including mean, standard deviation, 95% confidence interval and range. The frequency of participant responses to the perceived overall body armour fit (i.e., too small, good fit, and too large) was also calculated. A series of one-way ANOVAs were then performed to compare the mean value of all normally distributed anthropometric measurements ($n = 12$) characterising the torso and breast among those participants who reported their body armour was "too small," a "good fit," or "too large." Games-Howell *post-hoc* analyses were then conducted to determine where any difference lay. For those measurements that did not meet the assumption of normality ($n = 10$), differences in mean rank were similarly assessed using a Kruskal-Wallis H-Test. Dwass-Steel-Critchlow-Fligner *post-hoc* analyses were then conducted to determine where any differences lay. Although multiple statistical tests were conducted, increasing the chance of incurring an error, no adjustment to the alpha level was deemed necessary given the exploratory nature of the study and the low cost associated with incurring a Type I error (Sinclair et al., 2013). Statistical analyses were performed in SPSS (Version 23, IBM Statistics, Chicago, USA) with an alpha level of $p < 0.05$.

RESULTS

Torso Characteristics and Perceived Fit

Normative data on the torso characteristics of the participants are presented in **Table 4**. Comparisons of the mean value or mean rank of the torso characteristics among participants who reported that their body armour was "too small" ($n = 14$; 14%), a "good fit" ($n = 30$; 31%) or "too large" ($n = 53$; 55%) are presented in

TABLE 3 | A description of how each breast measurement was digitally calculated, as well as a visual depiction of the measurement in Geomagic.

Measurement	Description of how the measurement was calculated in Geomagic	Visual depiction of the measurement
Breast Volume (mL)	From each participant's scan, a three-dimensional, isolated model of each breast was created by outlining the breast **(A)**, removing the breast from the torso **(B)** and attaching it to the corresponding anterior chest wall (whose curvature approximated the superficial surface of the pectoralis major muscle; **C,D**). These steps were performed to create a closed three-dimensional breast model **(E,F)**, from which breast volume (mL) was calculated.	
Breast Surface Area (cm^2)	Using the closed three-dimensional breast model **(E,F)**, the surface area (cm^2) of each participant's right and left breast was calculated.	
Anterior Breast Projection (mm)	Using the closed three-dimensional breast model **(E,F)**, the distance from the posterior breast wall to the most anterior point of the breast was calculated.	
Breast Length (mm)	The linear distance (mm) between the inferior and superior borders of each participant's right and left breast was measured at the longest vertical point.	
Breast Width (mm)	The linear distance (mm) between the medial and lateral borders of each participant's right and left breast was measured at the widest point.	
Sternal Notch to Nipple Distance (mm)	The linear distance (mm) from the sternal notch (where a marker had been placed before scanning) to the nipple of each participant's right and left breast was measured.	
Sternal Notch to Superior Breast Distance (mm)	The perpendicular distance (mm) between the sternal notch and a horizontal line drawn across the torso at the level of the superior border of each participant's right and left breast was measured.	
Sternal Notch to Inferior Breast Distance (mm)	The perpendicular distance (mm) between the sternal notch and a horizontal line drawn across the torso at the level of the superior border of each participant's right and left breast was measured.	

Unless otherwise stated, all measurements were taken on each participant's right and left breast and the mean value was recorded.

TABLE 4 | Torso and breast characteristic data (mean ± standard deviation, confidence interval, and range) of the study participants ($n = 97$) are shown.

Anthropometric characteristic	Mean ± SD AWAS study	Mean ± SD Current study	95% Confidence interval (lower bound, upper bound)	Range
Stature (cm)	165.4 ± 6.1	165.3 ± 6.3	164, 166.5	152.6–180.5
Body Mass (kg)	66.0 ± 9.7	69.6 ± 10.5	67.5, 71.6	49.7–98.7
Iliocristale Height (cm)	100.9 ± 4.8	100.9 ± 4.8	99.9, 101.8	90.8–116.5
Waist Height (cm)	105.6 ± 4.8[a]	106.0 ± 5.5	104.9, 107.1	96.1–138.4
Suprasternale Height (cm)	134.1± 5.4	134.5 ± 5.4	133.4, 135.5	123.2–148.7
Front Length (cm)	33.2 ± 2.2	33.7 ± 2.3	33.2, 34.1	27.8–41.1
Chest Depth (cm)	—[b]	38.4 ± 1.7	27.4, 28.1	34.9–44.8
Chest Breadth (cm)	27.4 ± 2.3	27.8 ± 1.8	38.0, 38.7	23.4–32.6
Bi-acromial Breadth (cm)	37.3 ± 1.7	37.6 ± 1.6	37.3, 38.0	34.3–42.6
Neck Circumference (cm)	32.9 ± 2.0	32.8 ± 1.8	32.4, 33.1	29.6–37.5
Waist Circumference (cm)	83.7 ± 9.0[c]	74.7 ± 6.8	73.3, 76.0	62.1–98.5
Hip Circumference (cm)	99.3 ± 6.5	102.8 ± 6.6	101.5, 104.1	91.3–122.5
Over-Bust Chest Circumference (cm)	90.7 ± 7.5	89.7 ± 8.6	88.0, 91.4	34.1–109.0
Under-Bust Chest Circumference (cm)	76.5 ± 5.9	76.9 ± 5.6	75.8, 78.1	67.4–95.2
Breast Volume (mL)	—	365.2 ± 167.7	331.8, 398.6	103.6–885.8
Breast Surface Area (cm^2)	—	443.9 ± 104.1	423.2, 464.6	250.5–706.6
Anterior Breast Projection (mm)	—	43.5 ± 10.3	41.5, 45.6	23.2–71.2
Breast Length (mm)	—	151.7 ± 14.8	148.8, 154.7	115.2–183.6
Breast Width (mm)	—	157.9 ± 17.3	154.5, 161.3	123.7–204.8
Sternal Notch to Nipple Distance (mm)	—	183.8 ± 19.5	179.9, 187.7	95.8–228.3
Sternal Notch to Superior Breast Distance (mm)	—	57.7 ± 10.8	55.6, 59.8	31.8–82.7
Sternal Notch to Inferior Breast Distance (mm)	—	209.4 ± 18.4	205.8, 213.1	164.3–266.3

Normative data of female soldiers from the Australian Warfighter Anthropometric Survey (AWAS) are shown in the first column for comparison purposes (Edwards et al., 2014).
[a]AWAS measurement of waist height was from the 10th rib, whereas the current study calculated Waist Height at the narrowest point between the 10th rib and the iliac crest.
[b]Although AWAS measurement of chest depth was collected, the measurement protocol was substantially different and comparative data have therefore been omitted.
[c]Waist Height was taken at different levels between studies as described in[a], affecting the Waist Circumference measures.
—indicates measurements that were not collected in AWAS.

Table 5. There was a significant difference in both Stature [$f_{(2)} = 3.715$, $p = 0.035$] and Suprasternale Height [$f_{(2)} = 3.899$, $p = 0.03$] between participants who rated the fit of their body armour as good compared to those who rated it as too large, whereby female soldiers who rated the fit as too large were, on average, shorter than female soldiers who rated the fit as good. Chest Breadth [$f_{(2)} = 11.547$, $p < 0.001$] and Waist Circumference [$\chi^2_{(2)} = 15.77$, $p < 0.001$] also differed significantly among those who rated the fit of their body armour as too small compared to too large, and too small compared to a good fit. Female soldiers who rated the fit as too small had, on average, larger torso breadths and circumferences than female soldiers who rated the fit as either good or too large. Similarly, Mass [$\chi^2_{(2)} = 12.85$, $p = 0.002$], Chest Depth [$f_{(2)} = 4.872$, $p = 0.014$], Neck Circumference [$\chi^2_{(2)} = 10.20$, $p = 0.006$] and chest circumference measures of OBCC [$\chi^2_{(2)} = 11.660$, $p = 0.003$] and UBCC [$\chi^2_{(2)} = 9.16$, $p = 0.01$] were also significantly larger, on average, in female soldiers who rated the fit as too small compared to those who rated the fit as too large. There was no significant difference in participants' ratings of fit for Waist Height [$\chi^2_{(2)} = 4.33$, $p = 0.115$], Hip Circumference [$\chi^2_{(2)} = 5.78$, $p = 0.056$], Front Length [$f_{(2)} = 2.290$, $p = 0.112$], Biacromial Breadth [$f_{(2)} = 0.125$, $p = 0.883$] or Iliocristale Height [$f_{(2)} = 1.760$, $p = 0.188$].

Breast Characteristics and Perceived Fit

Normative data on the breast characteristics of the participants are presented in **Table 4**. Comparisons of the mean value or mean rank of the breast characteristics among participants who reported their body armour was "too small" ($n = 14$), a "good fit" ($n = 30$) or "too large" ($n = 53$) are presented in **Table 6**. Breast Surface Area [$f_{(2)} = 4.596$, $p = 0.018$], Breast Width [$f_{(2)} = 6.150$, $p = 0.005$], and Sternal Notch to Nipple Distance [$\chi^2_{(2)} = 12.32$, $p = 0.002$] differed significantly among those who rated the fit of their body armour as too small compared to too large and too small compared to a good fit. Female soldiers who rated the fit as too small had, on average, larger breast sizes, shapes and positions than female soldiers who rated the fit as either good or too large. Similarly, Breast Volume [$\chi^2_{(2)} = 6.43$, $p = 0.04$],

TABLE 5 | Torso characteristic data compared between the three fit groups (too small, good fit, and too large) using One-Way ANOVA (normally distributed data; difference in group means) or Kruskal–Wallis (non-normally distributed data; difference in mean rank).

Torso characteristic	Fit	N	Mean	SD	SIG	vs. good fit	vs. too large
Stature (cm)	Too small	14	167.2	7.9595	0.035*	0.997	0.318
	Good fit	30	167.0	4.8955			0.028*
	Too large	53	163.8	6.23			
Body Mass (kg)	Too small	14	78.0	10.241	0.002*	0.092	0.001*
	Good fit	30	71.0	10.4861			0.214
	Too large	53	66.5	9.1836			
Iliocristale Height (cm)	Too small	14	102.1	5.93	0.188	—	—
	Good fit	30	101.7	3.61			—
	Too large	52	100.0	4.99			
Waist Height (cm)	Too small	14	107.3	5.5	0.115	—	—
	Good fit	30	106.6	3.52			—
	Too large	53	105.4	6.42			
Suprasternale Height (cm)	Too small	14	136.1	6.57	0.03*	1	0.307
	Good fit	30	136.0	4.07			0.023*
	Too large	53	133.2	5.46			
Front Length (cm)	Too small	14	33.9	1.15	0.112	—	—
	Good fit	30	34.3	2.17			—
	Too large	52	33.2	2.58			
Chest Depth (cm)	Too small	14	39.5	1.99	0.014*	0.418	0.037*
	Good fit	30	38.7	1.55			0.081
	Too large	53	37.9	1.58			
Chest Breadth (cm)	Too small	14	29.6	1.58	<0.001*	0.004*	<0.001*
	Good fit	30	27.7	1.84			0.557
	Too large	53	27.3	1.56			
Bi-acromial Breadth (cm)	Too small	14	37.8	1.64	0.883	—	—
	Good fit	30	37.7	1.73			—
	Too large	53	37.6	1.6			
Neck Circumference (cm)	Too small	14	33.9	1.54	0.006*	0.124	0.005*
	Good fit	30	32.9	1.84			0.349
	Too large	53	32.4	1.73			
Waist Circumference (cm)	Too small	14	80.9	5.6	<0.001*	0.026*	<0.001*
	Good fit	30	75.3	7.49			0.396
	Too large	53	72.7	5.66			
Hip Circumference (cm)	Too small	14	107	7.71	0.056	—	—
	Good fit	30	103	7.12			—
	Too large	53	101.5	5.45			
Over-Bust Chest Circumference (cm)	Too small	14	96.0	6.6443	0.003*	0.068	0.002*
	Good fit	30	90.5	7.1124			0.485
	Too large	53	87.6	9.1305			
Under-Bust Chest Circumference (cm)	Too small	14	80.3	4.4869	0.01*	0.111	0.006*
	Good fit	30	77.7	6.7732			0.592
	Too large	53	75.7	4.7941			

*The third last column provides the p-value for the main effects of fit group on each torso characteristic. The final 2 columns provide p-values of the pairwise comparisons between the three fit groups, as determined through post-hoc analysis. *represents significance at p < 0.05. For variables that were found to have no significant difference between fit groups, post-hoc tests were not conducted and the corresponding cells were marked with a long dash.*

Anterior Breast Projection [$\chi^2_{(2)} = 6.33, p = 0.042$], Breast Length [$f_{(2)} = 4.923$, $p = 0.013$] and Sternal Notch to Inferior Breast Distance [$f_{(2)} = 4.047$, $p = 0.027$] were significantly larger, on average, in female soldiers who rated the fit of their body armour as too small compared to those who rated the fit as too large. There was no significant difference in participants' rating of fit for Sternal Notch to Superior Breast Distance [$f_{(2)} = 0.859$, $p = 0.433$].

TABLE 6 | Breast characteristic data compared between the three fit groups (too small, good fit, and too large) using One-Way ANOVA (normally distributed data; difference in group means) or Kruskal–Wallis (non-normally distributed data; difference in mean rank).

Breast characteristic	Fit	*N*	Mean	SD	SIG	vs. good fit	vs. too large
Breast Volume (mL)	Too small	14	503.0	225.3185	0.04*	0.064	0.039*
	Good fit	30	344.5	152.4872			1
	Too large	53	340.5113	142.4827			
Breast Surface Area (cm^2)	Too small	14	528.8143	116.4542	0.018*	0.044*	0.018*
	Good fit	30	435.46	102.9021			0.913
	Too large	53	426.2811	91.6439			
Anterior Breast Projection (mm)	Too small	14	50.85	11.8747	0.042*	0.062	0.044*
	Good fit	30	42.2467	9.9361			0.977
	Too large	53	42.3509	9.3534			
Breast Length (mm)	Too small	14	162.3857	13.9594	0.013*	0.064	0.013*
	Good fit	30	151.4233	15.1327			0.773
	Too large	53	149.1226	13.8933			
Breast Width (mm)	Too small	14	173.3	18.1845	0.005*	0.02*	0.006*
	Good fit	30	156.4133	17.3884			0.89
	Too large	53	154.6679	14.9703			
Sternal Notch to Nipple Distance (mm)	Too small	10	203.01	14.9471	0.002*	0.006*	0.002*
	Good fit	25	181.724	16.6891			0.989
	Too large	49	180.9408	19.7643			
Sternal Notch to Superior Breast Distance (mm)	Too small	14	61.2214	12.9773	0.433	—	—
	Good fit	30	58.1867	9.0009			—
	Too large	53	56.4868	11.0649			
Sternal Notch to Inferior Breast Distance (mm)	Too small	14	223.6214	22.1102	0.027*	0.113	0.029*
	Good fit	30	209.61	16.6556			0.549
	Too large	53	205.6038	16.7754			

The third last column provides the p-value for the main effects of fit group on each breast characteristic. The final 2 columns provide p-values of the pairwise comparisons between the three fit groups, as determined through post-hoc analysis. ∗represents significance at $p < 0.05$. For variables that were found to have no significant difference between fit groups, post-hoc tests were not conducted and the corresponding cells were marked with a long dash.

DISCUSSION

This is the first published study to provide normative data on the torso and breast size and shape of female soldiers. These soldiers displayed a wide range of anthropometric characteristics that must be considered when sizing and designing body armour. Sixteen of the twenty-two measurements assessed were significantly associated with the soldiers' ratings of body armour fit, suggesting a link between breast and torso characteristics and the overall perceived fit of body armour systems by female soldiers. The implications of these findings in terms of the design and sizing of current-issue body armour systems are discussed below.

Given that body armour systems are torso borne, the size and shape of the torso is a key consideration when designing body armour systems to closely interface with the human body. Among study participants, torso characteristic data were similar to all comparable measurements previously reported by the AWAS for ADF female soldiers (Edwards et al., 2014). Large variations, however, were observed in individual measurements, evident in the large standard deviations (**Table 4**). This large variability highlights the range of shapes and sizes that body armour systems should be designed to accommodate, as well as sex-related differences in torso dimensions compared to male AWAS data (Edwards et al., 2014). For example, female soldiers in the present study had substantially narrower chests (27.8 vs. 30.5 cm AWAS; Chest Breadth) and smaller waists (74.7 vs. 92.4 cm AWAS; Waist Circumference) but larger hips (102.8 vs. 99.7 cm AWAS; Hip Circumference) when compared to male AWAS data (Edwards et al., 2014). Given the limited, unisex sizing range of current issue armour, it is unsurprising that many females have reported to be wearing ill-fitting body armour (Coltman et al., 2020).

Consistent with our hypothesis, several torso measurements were significantly associated with a poor fit. Participants who perceived their body armour to be too large ($n = 53$) were more likely to be shorter in Stature and Suprasternale Height (**Table 5**) compared to those who perceived their body armour to be a good fit ($n = 30$), suggesting the length of the in-service body armour system may be too long for many users. Body armour is designed to be positioned superiorly at the level of the suprasternal notch. As armour length increases relative to overall height and height of the suprasternal notch, it is increasingly likely to interfere with trunk mobility and task performance and, therefore, perceptions of armour fit (Molloy et al., 2020; Coltman et al., 2021a, 2022). Conversely, participants who reported their body armour to be too small ($n = 14$) had a significantly greater

Mass, chest circumference (OBCC and UBCC), Chest Depth, Chest Breadth, Waist Circumference, and Neck Circumference compared to those who perceived their body armour to be too large ($n = 54$). This either suggests that the body armour system is not wide enough to accommodate large torso circumferences or that the mechanism to tighten the system does not have sufficient adjustment capacity. Designers of body armour for women should pay particular attention to ensure adjustment points on the front and rear carriers, including the shoulder straps and cummerbund, can cater for the variability in chest curvature of women's torsos due to their additional breast tissue. Ultimately, military organisations should consider developing body armour systems that are better suited to the range of female soldier torso sizes, both large and small (Wen and Shih, 2020). Such systems are likely to improve perceptions of fit and have important implications for task performance and efficiency in the field (Mitchell et al., 2010; Choi et al., 2019; Coltman et al., 2021a). Issuance procedures (e.g., sizing and allocation of systems) and education (e.g., training on system use) should also be updated to accompany the introduction of any new equipment because previous research has identified that these factors contribute to reports of dissatisfaction among female body armour users (Coltman et al., 2021a).

Clearly, the breasts of female soldiers are substantially different to male soldiers, and this disparity can affect body armour fit. The mean Breast Volume reported in this study for female soldiers (365 ml) was consistent with mean Breast Volume for women from an athletic population (401.7 ml; Brisbine et al., 2020a), but smaller than that previously reported for the general female population (653 ml left and 647 ml right breast; Coltman et al., 2017a). This smaller average breast size of the female soldiers is likely to be a function of the younger age (mean: 25.6 years) and lower body mass index (mean: 25.4 kg/m^2) of the study participants compared to the general population (Edwards et al., 2014; Coltman et al., 2017a). There was, however, a wide range of breast sizes (see **Table 3**). This diversity in breast size has important implications for armour design. For example, female soldiers with large breasts are likely to benefit from body armour systems designed with space to accommodate additional breast tissue, although such space would not be needed for soldiers with small breasts. Given the similarity of average breast size between participants who rated their armour as a good fit (344.5 mL; $n = 14$; 14%) and too large (340.5 mL; $n = 30$; 31%) compared to those who rated their armour as too small (503.0 mL; $n = 53$; 55%; **Table 6**), it would appear that female soldiers with a breast size of medium or large (formally classed as 350 mL and above; Coltman et al., 2017b) are most affected by sizing of body armour systems around the chest. This represents a substantial portion of the female soldier population, thus highlighting the importance of using anthropometric data to inform future armour design.

In addition to breast size, breast shape characteristics, such as Breast Surface Area and Anterior Breast Projection, will influence the overall fit of a torso borne body armour system. Given the range of breast projection distances documented in the current study (23–71 mm), women with breasts that protrude further anteriorly from the chest wall are likely to experience greater difficulty achieving conformity between a body armour system and their torso (Coltman et al., 2021b). The resultant compression and deformation of the breast when wearing body armour, which is exaggerated in women with large breasts (McGhee and Steele, 2020), is likely to have negative functional and protection impacts (e.g., difficulty breathing, limited ROM, exposed lateral breast tissue, and compromised positioning of vital torso protection). Consequently, our findings corroborate previous reports that occupational body armour does not adequately accommodate the full range of female breast characteristics (Coltman et al., 2020, 2021a; Malbon et al., 2020; Niemczyk et al., 2020). It was therefore unsurprising that Breast Volume, Breast Surface Area and Anterior Breast Projection were each associated with the perception of poor body armour fit (**Table 6**). All three breast measures were significantly larger in female soldiers who perceived their body armour to be too small compared to those who perceived it to be too large, with Breast Surface Area additionally differing between those who perceived their body armour to be too small and a good fit. As previous research has demonstrated the capacity of body armour to cause mild restrictive ventilatory impairment in male soldiers (Armstrong and Gay, 2016; Armstrong et al., 2019), reports of difficulty breathing due to excessive compression of the breasts and chest in the present study are likely magnified as breast size increases. Further research, however, is required to confirm this notion. Similarly, experiences of body armour being too loose around the waist to accommodate additional tissue in the upper chest may be further associated with breast size and shape variations amongst female soldiers (Coltman et al., 2021a). In both instances, design modifications to body armour have the potential to reduce breathing restrictions and, in turn, minimise discomfort and impairments to performance (Armstrong and Gay, 2016). Body armour system weight and load carriage have previously been associated with expiratory flow limitations (Armstrong et al., 2019). Furthermore, excessive weight of body armour has been reported as the third-most disliked characteristic of current-issue body armour by female soldiers (Coltman et al., 2022). Therefore, a lighter body armour system, one with smaller sizing dimensions, is likely to afford similar benefits to respiratory function.

Female soldiers in the present study also displayed a wide variety of breast positions relative to their torso. This was illustrated by a wide range of Sternal Notch to Nipple Distances (95.8–228.3 mm) and breast position measures, whereby the superior border of the breast was found to sit just 3 cm below the sternal notch for some soldiers and more than 5 cm lower for others. Body armour should be positioned at the level of the sternal notch to ensure sufficient coverage of the vital thoracoabdominal organs (Laing and Jaffrey, 2019). Knowledge of breast position is therefore crucial when designing body armour for female soldiers. Importantly, female soldiers whose breasts were situated lower on their torso were more likely to perceive their body armour system fit as too small. Coupled with other breast size and shape measures, breast position data indicate where potential plate shape changes or additional adjustability features need to be incorporated into a body armour system to better accommodate female breasts. Police body armour manufacturers have explored thermal forming and darting to provide a female-specific shape to soft ballistic plates (Basich, 2007). There is similar support in the literature for a contoured plate based on three-dimensional torso scans or anthropometry of female wearers (Boussu and Bruniaux, 2012; Cichocka et al., 2014; Mahbub et al., 2014; Abtew et al., 2018). The concept of a formed plate also warrants consideration for military body armour applications. The benefits of contouring rigid military hard ballistic plates, however, are not fully understood, and it is unknown how mobility and ballistic performance may be influenced by any changes. Female soldiers have previously reported that body armour is either too tight over the breasts to fit correctly around the waist or too loose around the waist to fit correctly over the breasts (Coltman et al., 2021a). An average difference of 15 cm was reported between mean chest circumference (OBCC; 89.7 cm) and mean Waist Circumference (74.7 cm) in this study, suggesting that incorporating multiple adjustments points on body armour systems is warranted. Any form changes, however, must be evaluated against protection implications because protection remains the primary function of military body armour systems (Laing and Jaffrey, 2019).

Fit and function issues associated with body armour are further exacerbated by wearing a bra because a bra can change breast shape and limit breast deformability under a hard plate. Moreover, bras function to lift and support the breasts to reduce pain and discomfort associated with breast movement (Scurr et al., 2014; Brisbine et al., 2020b), whereas body armour places downwards pressure on the breasts, effectively acting against the support provided by a bra (Niemczyk et al., 2020). Therein lies an additional challenge in designing body armour suitable for female soldiers with large breasts. That is, large breasts require more upwards support from a bra (thus limiting deformability) yet are more susceptible to the downward compression from body armour that may act to counteract this support. Bras can also be a major source of discomfort for female wearers of body armour because of the way body armour affects the fit and form of otherwise well-fitting, supportive bras (Niemczyk et al., 2017, 2020; Burbage et al., 2021). Furthermore, body armour can limit the type of bra a female soldier can wear. Compression-style sports bras, which are commonly wire-free and function to reduce anterior protrusion of the breast, are the most commonly worn type of bra among Australian (Coltman et al., 2021b) and British female soldiers (Burbage et al., 2021). Compression-style bras, however, do not provide sufficient breast support for women with medium-large breast sizes (McGhee et al., 2008), who comprised 44% ($n = 43$) of the study population. Researchers examining police body armour have suggested that designing a supportive bra to specifically integrate with body armour may be an effective short-term solution (Malbon et al., 2020; Niemczyk et al., 2020). Such a bra should be a wire-free design that reduces anterior breast protrusion while distributing breast tissue to improve the fit of body armour over the breasts and alleviate discomfort associated with current bra-armour incompatibility (Malbon et al., 2020; Coltman et al., 2021b).

The findings of the current study must be considered in light of their limitations. Although this study was the first to provide normative data on female soldiers' torso and breast characteristics, no objective measures of body armour fit or performance were collected. Therefore, breast characteristics could only be associated with the fit of body armour reported by the female soldiers. Participant perceptions have been used in other research evaluating protective equipment (Park and Hahn, 2014) and therefore are deemed to be a valuable metric for assessing overall fit in the present study. Future research, however, is recommended to incorporate both subjective and objective measures related to body armour fit, including static, dynamic, occupation-specific and cognitive fit (Stirling et al., 2020). Additionally, the sizes of the body armour systems issued to the participants were not recorded; TBAS Tier 2 is only available in one size, but TBAS Tier 3 is available in four sizes (S-XL), which may account for some differences in participant ratings of subjective fit. Participants were not asked to bring their body armour system to the test session and most participants were unaware of the size of the body armour that they had been issued, preventing collection of these data. As this study focused exclusively on the anthropometry, design, and sizing requirements of female soldiers, it is also recommended that future research similarly profile the male torso to assess potential fit and form issues experienced by male soldiers and to better target design and sizing improvements to the entire soldier population. Fundamentally, any design changes must be considered against known protection requirements, including vital thoracoabdominal organ protection.

CONCLUSION

Normative data characterising the torso and breasts of 97 female soldiers highlight the variation in anthropometric dimensions that body armour systems must cater for, as well as the implications of the varied torso and breast sizes and shapes for perceived body armour fit. Future body armour systems should cater for female soldiers' physical diversity by developing an expanded sizing range and female-specific design features to improve perceptions of fit. Improved perceptions of fit will, in turn, reduce movement interference, discomfort, and barriers to performance in the field. Any modifications to body armour should be informed by anthropometric data representing female soldiers and aim to ensure that vital thoracoabdominal organ protection recommendations are maintained.

DATA AVAILABILITY STATEMENT

The datasets presented in this article are not readily available because participants of this study did not agree for their data to be shared publicly. Requests to access the datasets should be directed to Celeste.Coltman@Canberra.edu.au.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Defence Science and Technology Group Low Risk Ethics Panel, Department of Defence, Australia. The participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

CC was the primary investigator. She was responsible for the study design, recruiting the participants collecting, analysing and interpreting the data, co-writing the first full draft of the manuscript, and approving the final version of the manuscript. BB was responsible for analysing and interpreting the data, co-writing the first full draft of the manuscript with the primary investigator, and approving the final version of the manuscript. RM was responsible for helping to develop the study design, recruiting the participants, assisting to interpret the data, providing feedback on versions of the manuscript, and approving the final version of the manuscript. JS was responsible for helping to develop the study design, assisting to interpret the data, providing substantial feedback on versions of the manuscript and approving the final version of the manuscript. All authors contributed to the article and approved the submitted version.

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Effects of Occasional and Habitual Wearing of High-Heeled Shoes on Static Balance in Young Women

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The purpose of this study was to examine the effects of occasional and habitual wearing of high-heeled shoes on static balance in young women. Groups of habitual high-heel wearers and non-wearers ($n = 7$ in both groups) were asked to stand quietly on a force platform without shoes (WS condition) or with high heels (heel area 1 cm^2, heel height 7 cm) (HH condition). During the trials, the center-of-pressure (CoP) position in the anterior-posterior direction was measured, and its root mean square (as a measure of postural sway magnitude, CoP_{RMS}) and mean velocity (as a measure of regulatory activity, CoP_{MV}) were calculated. To further examine the effect of high-heel wearing on the temporal aspects of slow and fast processes in static balance, the CoP sway was decomposed into low- (below 0.5 Hz) and high- (above 0.5 Hz) frequency components, and then spectral analysis was performed. Results showed that the CoP_{RMS} was not significantly different between the groups or between the shoe conditions, indicating that wearing high heels with a heel height of 7 cm did not increase the magnitude of postural sway, irrespective of high-heel experience. The CoP_{MV} was significantly larger in the HH condition than in the WS condition, whereas it was not significantly different between the groups. This result indicates that wearing high heels increased the amount of regulatory activity in both habitual wearers and non-wearers. The spectral analysis further showed that habitual high-heel wearers showed significantly decreased rate of regulatory activity than non-wearers, both while standing with and without high heels. These results suggest that use-dependent changes in static balance control are evident in both high-heeled and without shoes conditions.

Keywords: high-heeled shoes, postural control, women, center-of-pressure (CoP), quiet standing

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INTRODUCTION

In the mid-nineteenth century, high-heeled shoes became popular among women in all socioeconomic classes in many countries. During this period, women wore high heels to prevent their dresses dragging on the ground (Danesi, 2018). Although high heels are sometimes uncomfortable and unsuitable for locomotion, millions of women currently wear high heels in their daily lives. An important perceived benefit of wearing high heels is increased attractiveness of the wearer. Lewis et al. (2017) demonstrated that, when participants stood still in high heels, their lumbar curvature increased, and they were perceived as more attractive. Regarding walking, Morris et al. (2013) reported that walking in high heels enhanced the perceived femininity of gait by reducing stride length and increasing rotation and tilt of the hips.

However, various negative outcomes have been reported to be associated with habitual wearing of high heels, including orthopedic problems such as hallux valgus (Menz and Morris, 2005), knee osteoarthritis (Kerrigan et al., 2005), and lower back pain (Lee et al., 2001). In addition, regular use of high heels can induce structural and functional changes in the calf muscle-tendon unit (MTU), with studies reporting reduced fascicle length of gastrocnemius medialis (Csapo et al., 2010; Cronin et al., 2012), increased cross-sectional area and stiffness of the Achilles' tendon (Csapo et al., 2010), and reduced active range of motion of the ankle (Csapo et al., 2010) among women who regularly wear high heels. Recent studies have reported that functional interactions between muscle fibers and tendinous tissues are relevant not only for storing/releasing energy during dynamic movements (Fukunaga et al., 2002) but also for controlling static balance (Loram et al., 2005a). Therefore, changes in structural and mechanical properties of the calf MTU induced by long-term use of high heels are expected to result in substantial changes in the control of static balance. Some previous studies have investigated the effects of high-heel experience on static balance by comparing center-of-pressure (CoP) sway between habitual high-heel wearers and non-wearers while wearing high heels with different heel heights (Hapsari and Xiong, 2016; Wan et al., 2019). Wan et al. (2019) reported that the habitual high-heel wearers exhibited significantly smaller mean velocity of CoP (CoP_{MV}) in the anterior-posterior (AP) direction for various heel heights (1, 5, 8, and 10 cm) compared with non-wearers. It should be noted that these previous studies compared static balance between habitual high-heel wearers and non-wearers while participants stood with heeled shoes. However, in the current study, we hypothesized that changes in static balance would become more prominent when standing barefoot or without shoes, because the reduced fascicle length of the calf muscle leads the resting positions of the ankle to be more plantarflexed (Csapo et al., 2010) and regular high-heel wearers often experience calf muscle pain when standing barefoot (Opila et al., 1988). Therefore, we examined this hypothesis by comparing CoP sway between habitual high-heel wearers and non-wearers while standing with and without high heels.

To quantify the CoP sway, two conventionally used statistical measures were calculated: the root mean square (CoP_{RMS}) and CoP_{MV}. CoP_{RMS} represents sway magnitude, or in other word, the effectiveness of postural control system, whereas CoP_{MV} represents the amount of regulatory activity (Hufschmidt et al., 1980; Prieto et al., 1996). Therefore, participants who exhibit slow and large drifts in the body position would have a large CoP_{RMS}. In contrast, those who make more fast ballistic corrections of balance would have a large CoP_{MV} (Kirshenbaum et al., 2001). It has also been shown that low- (CoP_{LF}) and high- (CoP_{HF}) frequency components of the CoP sway reflect different processes in the control of static balance (Zatsiorsky and Duarte, 2000; Loram et al., 2005b). The CoP_{LF} below 0.5 Hz corresponds to the horizontal projection of the center of body mass (CoM). On the other hand, the CoP_{HF} around 1 Hz corresponds to ballistic regulation of ankle joint torque by which the postural control system regulates the CoM movement. Therefore, to further examine the effects of occasional and habitual wearing of high-heeled shoes on the temporal aspects of two different processes in static balance, we decomposed the CoP sway into the CoP_{LF} and CoP_{HF}, and then performed spectral analysis.

MATERIALS AND METHODS

Participants

Because previous studies have reported significant effects of high-heel experience on postural control during quiet standing (Hapsari and Xiong, 2016; Wan et al., 2019), we expected substantially large interaction in regulatory activity between habitual high-heel wearers and non-wearers. When we conducted *a priori* power analysis using G*power (Faul et al., 2007), with the two-way mixed-design analysis of variance (ANOVA), 0.05 significance threshold, 0.8 detection power, and effect size of partial eta squared (η_p^2) = 0.26, the required total sample size was calculated to be ten. We recruited 14 young women aged 21–39 from our colleagues and students. The participants were divided into two equally sized groups: a habitual high-heel wearer group and a non-wearer group (n = 7 in both groups). Participants in the wearer group (mean ± standard deviation (SD): age 31.4 ± 6.7 years, height 161.1 ± 7.1 cm, weight 54.3 ± 14.5 kg) wore high heels with heel height > 5 cm at least 5 days per week over the previous 2 years or longer. In contrast, participants in the non-wearer group (age 28.6 ± 7.7 years, height 159.9 ± 5.0 cm, weight 52.1 ± 4.6 kg) wore high heels with heel height > 5 cm < 2 days per week over the previous 2 years or longer. Independent-samples t-tests revealed no significant differences in age, height, or weight between the two groups. All participants were healthy and had no history of traumatic injury or surgical operation. All participants gave written informed consent to participate in this study. The experimental procedures used were approved by the Ethics Committee on Human Experimentation at the Graduate School of Arts and Sciences, The University of Tokyo, in accordance with the Declaration of Helsinki (#13-56).

Protocol

To eliminate the effects of swelling of the feet and fatigue, all experiments were conducted in the morning or early afternoon. Custom-made high-heeled shoes (heel area 1 cm^2, heel height 7 cm) in a range of sizes were prepared for this study (**Figure 1**). These high-heeled shoes had two adjustable straps with Velcro fasteners around the ankle and the dorsum of the foot. Prior to the experiment, a shoe-fitting session was performed to choose the best-fitting shoe size for each participant. In the shoe-fitting session and experiment, the tension of the shoe straps was adjusted by a professional shoe fitter (AY-Y). During a few minutes of shoe fitting and measurement preparation, the participants became accustomed to the standardized shoes. Participants were asked to wear stockings and stand quietly on a force platform (Type 9281B, Kistler, Switzerland) without shoes (WS condition) or while wearing high heels (HH condition) with their eyes open for just over 60 s. Participants held their arms comfortably by their sides, with their feet parallel to each other. Previous study demonstrated that narrow stance width (intermalleolar distance of <8 cm) increases postural sway in the

FIGURE 1 | A side view of the custom-made high-heeled shoe (heel area 1 cm^2, heel height 7 cm). The shoe had two adjustable straps with Velcro fasteners around the ankle and the dorsum of the foot.

mediolateral (ML) direction (Day et al., 1993). Therefore, we set the intermalleolar distance to 10 cm in order to minimize CoP sway in the ML direction. Five successive trials of the WS condition were followed by five successive trials of the HH condition. To avoid fatigue during the experiment, a rest of several tens of seconds was provided between trials in each shoe condition, and a rest of 5 min was provided between WS and HH conditions.

Analysis

From the ground reaction forces measured by a force platform at a sampling frequency of 100 Hz, the CoP position was calculated. It has been reported that static balance in the AP direction is under the control of the ankle (plantar/dorsiflexor), whereas that in the ML direction is under the control of the hip (abductor/adductor) (Winter et al., 1996). Because we were interested in the effects of wearing of high heels on the ankle mechanism of static balance, CoP sway in the AP direction was measured and analyzed. The CoP time-series was digitally smoothed with a cut-off frequency of 3.0 Hz (Gage et al., 2004) using a built-in low-pass filter in commercial software (LabChart 7, AD Instruments, Australia). For the subsequent analysis, 60 s of data in the middle portion of the recorded data were used. From the CoP time-series, the CoP_{RMS} and CoP_{MV} were calculated, as follows:

$$CoP_{RMS} = \sqrt{\frac{\sum_{i=1}^{N}(x_i - \tilde{x})^2}{N}} \tag{1}$$

$$CoP_{MV} = \frac{\sum_{i=1}^{N-1}|x_{i+1} - x_i|}{T} \tag{2}$$

where x_i denotes the CoP position at the i-th instant (i runs from 1 to $N-1$ or N), $\tilde{x}$ denotes the mean CoP position, N denotes the number of data points used in the analysis ($N = 6{,}000$), and T denotes the duration of each trial ($T = 60$ s). The CoP_{RMS} and CoP_{MV} were normalized by participants' body height (BH) (Cattagni et al., 2014; Oba et al., 2015).

The CoP time-series was decomposed into the CoP_{LF} and CoP_{HF}. The CoP_{LF} time-series was calculated by low-pass filtering the CoP time-series with a cut off-frequency of 0.5 Hz (Caron et al., 1997; Loram and Lakie, 2002). By subtracting the CoP_{LF} time-series from that of the CoP, we obtained the CoP_{HF} time-series. We quantified the temporal aspects of the CoP_{LF} and CoP_{HF} time-series using the mean power frequency (MPF). In the spectral analysis, the CoP_{LF} and CoP_{HF} time-series for a single trial was detrended and then divided into five segments (20 s each) with 50% overlap. A fast-Fourier transform algorithm was applied to each segment to yield the power spectral density (PSD) after being passed through a Hamming window (*pwelch* function in MATLAB R2021a, MathWorks, USA). The PSDs of individual segments were ensemble-averaged into the function for a single trial. Note that our use of a 20 s data window resulted in the frequency resolution of 0.05 Hz. The MPF was calculated as follows:

$$MPF = \frac{\int f \cdot P}{\int P} \tag{3}$$

where f and P denote the frequency and PSD, respectively.

Results are presented as mean and 95% confidence interval (CI) in the text and as mean and individual values in the figures. In the statistical analysis, differences between the groups and test conditions were examined using a two-way mixed-design ANOVA with repeated measures on the shoe condition (IBM SPSS Statistics 21, IBM, USA). The significance level was set at $P < 0.05$, and Bonferroni correction was used when necessary. The effect size was reported as η_P^2. According to Bakeman (2005), we defined an η_P^2 of 0.02 as small, one of 0.13 as medium, and one of 0.26 as large.

RESULTS

Figure 2 shows representative recordings of the CoP sway for one habitual high-heel wearer (*left*) and one non-wearer (*right*) in the WS (*A*) and HH (*B*) conditions. It should be noted that these CoP recordings were presented with respect to their mean values. Several comparable features in these recordings should be noted. First, in the WS condition, the overall excursion of the CoP appeared to be no different between the wearer and non-wearer, whereas the fast ripples (∼1 s per cycle) were more prominent in the CoP recordings for the wearer. In contrast to the WS condition, these fast ripples were observed in the recordings for both the wearer and non-wearer in the HH condition, and no apparent difference was observed between the two recordings.

Figure 3 shows results for the CoP_{RMS} (*top*) and CoP_{MV} (*bottom*) for each group and each test condition. For the CoP_{RMS}, there was no significant difference between the wearer group (mean = 3.098×10^{-3}, 95% CI [2.434×10^{-3}, 3.762×10^{-3}]) and non-wearer group (mean = 2.797×10^{-3}, 95% CI [2.134×10^{-3}, 3.461×10^{-3}]) [$F_{(1,12)} = 0.487$, $P = 0.499$, $\eta_P^2 = 0.039$] or between the WS condition (mean = 2.942×10^{-3}, 95% CI [2.259×10^{-3}, 3.625×10^{-3}]) and HH condition (mean = 2.953×10^{-3}, 95% CI [2.501×10^{-3}, 3.406×10^{-3}]) [$F_{(1,12)} = 0.001$, $P = 0.973$, $\eta_P^2 < 0.001$]. The CoP_{MV} was significantly larger in the HH condition (mean = 3.611×10^{-3}, 95% CI [3.313×10^{-3}, 3.909×10^{-3}]) than in the WS condition (mean = 2.735×10^{-3}, 95% CI [2.393×10^{-3}, 3.076×10^{-3}]) [$F_{(1,12)} = 27.800$, $P < 0.001$, $\eta_P^2 = 0.698$], whereas there was no significant difference between the wearer group

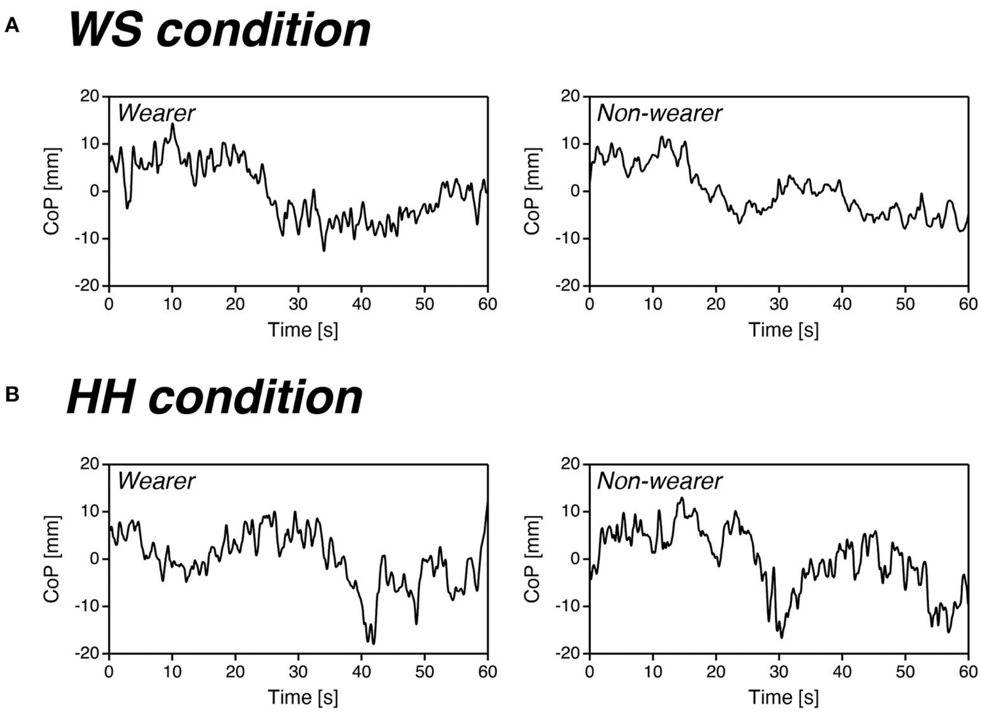

FIGURE 2 | Representative recordings of the CoP sway for one high-heel wearer (*left*) and one non-wearer (*right*) in the WS **(A)** and HH **(B)** conditions. **(A)** Overall excursion of the CoP showed no difference between the two participants, whereas fast ripples were more prominent in the CoP recordings for the wearer. **(B)** Fast ripples were observed in the recordings for both the wearer and non-wearer, and no apparent differences were observed between the two participants.

(mean $= 3.280\times10^{-3}$, 95% CI [2.906×10^{-3}, 3.654×10^{-3}]) and non-wearer group (mean $= 3.065\times10^{-3}$, 95% CI [2.691×10^{-3}, 3.439×10^{-3}]) [$F_{(1,12)} = 0.790$, $P = 0.392$, $\eta_P^2 = 0.062$]. There was no significant interaction between group and test condition [$F_{(1,12)} = 4.127$, $P = 0.065$, $\eta_P^2 = 0.256$].

Figure 4 shows representative examples of the CoP (*thin gray line*) and CoP_{LF} (*bold black line*) recordings (*left*) and corresponding CoP_{HF} recordings (*right*) for one non-wearer (the same participant and trials as shown in **Figure 2**) in the WS (*A*) and HH (*B*) conditions. Note that only 20 s of data from the 60-s recordings are presented in this figure to emphasize the details of the waveforms. From these two CoP_{HF} recordings, it can be seen that this participant exhibited increased regulatory activity while standing with high heels.

Figure 5 shows results for the MPF of CoP_{HF} (*top*) and CoP_{LF} (*bottom*) for each group and each test condition. The MPF of CoP_{HF} was significantly higher in the non-wearer group (mean = 1.008, 95% CI [0.968, 1.049]) than in the wearer group (mean = 0.939, 95% CI [0.898, 0.979]) [$F_{(1,12)} = 7.048$, $P = 0.021$, $\eta_P^2 = 0.370$], while there was no significant difference between the WS condition (mean = 0.959, 95% CI [0.916, 1.001]) and HH condition (mean = 0.989, 95% CI [0.958, 1.019]) [$F_{(1,12)} = 1.907$, $P = 0.192$, $\eta_P^2 = 0.137$]. There was no significant interaction between group and test condition in the MPF of CoP_{HF} [$F_{(1,12)} = 1.942$, $P = 0.189$, $\eta_P^2 = 0.139$]. For the MPF of CoP_{LF}, there was no significant difference between the wearer group (mean = 0.124, 95% CI [0.107, 0.141]) and non-wearer group (mean = 0.116, 95% CI [0.099, 0.133]) [$F_{(1,12)} = 0.565$, $P = 0.467$, $\eta_P^2 = 0.045$] or between the WS condition (mean = 0.113, 95% CI [0.098, 0.129]) and HH condition (mean = 0.127, 95% CI [0.108, 0.146]) [$F_{(1,12)} = 1.418$, $P = 0.257$, $\eta_P^2 = 0.106$].

DISCUSSION

Differences in the CoP_{RMS} and CoP_{MV}

To examine the effects of occasional and habitual wearing of high heels on static balance, we first compared CoP sway during quiet standing between habitual high-heel wearers and non-wearers. The results revealed no significant differences in CoP_{RMS} between groups, or between test conditions (**Figure 3**, *top*). Given that CoP_{RMS} represents the sway magnitude, this result indicates that wearing high heels with a heel height of 7 cm did not increase the magnitude of postural sway, irrespective of high-heel experience. In addition, the results revealed that CoP_{MV} was significantly greater in the HH condition compared with the WS condition (**Figure 3**, *bottom*), suggesting that participants exhibited more and/or strong regulatory changes in ankle joint torque while standing with high heels. It should be noted that the two-way ANOVA revealed a marginal significance for an interaction between the group and test condition in the CoP_{MV} ($P = 0.065$). Because of the small sample size of the present study ($n = 7$ each group), there may be a concern about making a Type II error by wrongly accepting a false null hypothesis. However, when we conducted *post-hoc* power analysis with $\eta_P^2 = 0.256$ and

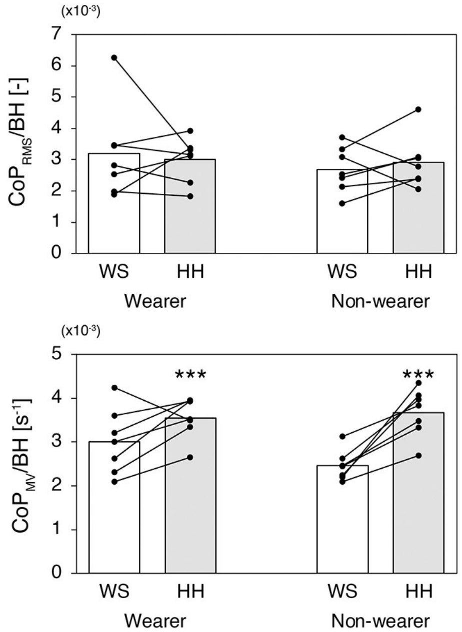

FIGURE 3 | The CoP_{RMS} (*top*) and CoP_{MV} (*bottom*) for each group and each test condition. The white bar indicates the WS condition and the gray bar indicates the HH condition. The small dots represent data for individual participants. ***$P < 0.001$ between the test conditions.

correlation among repeated measures = 0.265, the power was calculated to be 0.913. This result indicates that there was enough statistical power in the present study.

Increased CoP_{MV} in Habitual High-Heel Wearers

Although it failed to reach significance as stated above, the habitual wearers tended to exhibit marginally greater CoP_{MV} in the WS condition. Specifically, CoP_{MV} in the HH condition was almost the same for the wearers and non-wearers, whereas that in the WS condition is about 20% larger for the habitual wearers (**Figure 3**, *bottom*). Use-dependent changes in the structural and functional properties of the calf MTU may be responsible for the increased CoP_{MV} found in habitual high-heel wearers. First, because the reduced fascicle length of the calf muscle induced by the long-term use of high heels leads the resting positions of the ankle to be more plantarflexed (Csapo et al., 2010), greater passive plantarflexion torque was expected to be exhibited in these participants in the anatomical ankle position (i.e., while standing without shoes). Changes in passive plantarflexion torque per unit displacement of the ankle joint angle were also expected to be greater in high-heel wearers because of the non-linearity of the tendon stiffness (Kubo et al., 2001). Second, greater tendon stiffness strengthens the mechanical coupling between the contraction forces of the muscle fibers and the reaction force acting on the ground (Lakie et al., 2003). Finally, it has been reported that habitual high-heel wearers exhibit greater maximal voluntary isometric plantarflexion torque (∼10%) compared with non-wearers, at a wide range of ankle positions (−20° to +20°) (Csapo et al., 2010). All of these factors may contribute to amplification of the regulatory modulation of the ankle joint torque in habitual high-heel wearers, thereby increasing their CoP_{MV}.

Increased CoP_{MV} during quiet standing has commonly been observed in studies of people aged 60 years and above (Prieto et al., 1996; Kouzaki and Masani, 2012). Moreover, increased amplitude of the fast component of CoP sway has been reported to be associated with increased risk of falls in older people (Collins et al., 1995; Piirtola and Era, 2006). Although the underlying physiological mechanisms may differ between older people and young habitual high-heel wearers, the increased CoP_{MV} may represent an unfavorable change in balancing strategy, such as in terms of energy efficiency. To minimize the potential negative effects of long-term use of high heels, habitual wearers are recommended to perform stretching routines after wearing high heels to maintain flexibility of the calf MTU.

Differences in the MPF of CoP_{HF} and CoP_{LF}

Frequency domain analysis on the CoP_{HF} revealed that habitual high-heel wearers exhibited significantly lower MPF than non-wearers, regardless of the test conditions (**Figure 5**, *top*). This result indicates that habitual wearers showed decreased rate of regulatory activity than non-wearers, both while standing with and without high heels. Interestingly, occasional wearing of high heels by novice wearers resulted in an opposite effect on the MPF of CoP_{HF}. That is, although not statistically significant, the MPF of CoP_{HF} in novice wearers was slightly increased in the HH condition (mean ± SD: WS condition 0.98 ± 0.06 Hz, HH condition 1.04 ± 0.06 Hz). It is also interesting to note that, in habitual wearers, the MPF of CoP_{HF} remained almost unchanged in the WS and HH conditions (WS condition 0.94 ± 0.08 Hz, HH condition 0.94 ± 0.05 Hz), suggesting that long-term, habitual wearing of high-heeled shoes may change static balance control irrespective of shoe conditions. In contrast to the results for the MPF of CoP_{HF}, there were no significant differences in the MPF of CoP_{LF} between groups or shoe conditions (**Figure 5**, *bottom*), indicating that occasional or habitual wearing of high heels has no effect on the temporal aspects of slow bodily sway.

Biomechanical and Neurophysiological Effects of Wearing High Heels

Standing with high heels is similar to a condition in which participants stand over a declined surface (although some part of the forefoot is placed on a horizontal surface). Mezzarane and Kohn (2007) investigated the postural control on the inclined surfaces (14° toes-up, horizontal, and 14° toes-down) and found that declined surface significantly increased MPF of CoP and electromyography activity of soleus. Sasagawa et al. (2009b) also found increased tonic activities of soleus and gastrocnemius during toes-down standing and speculated that these increased tonic activities play a role in enhancing the ankle stiffness. With regard to the high-heel wearing, especially in non-wearers, it is likely that the posterior calf muscles increase tonic activity to eliminate the slack in the Achilles' tendon and to enhance ankle stiffness. Furthermore, high-heel wearing can affect the

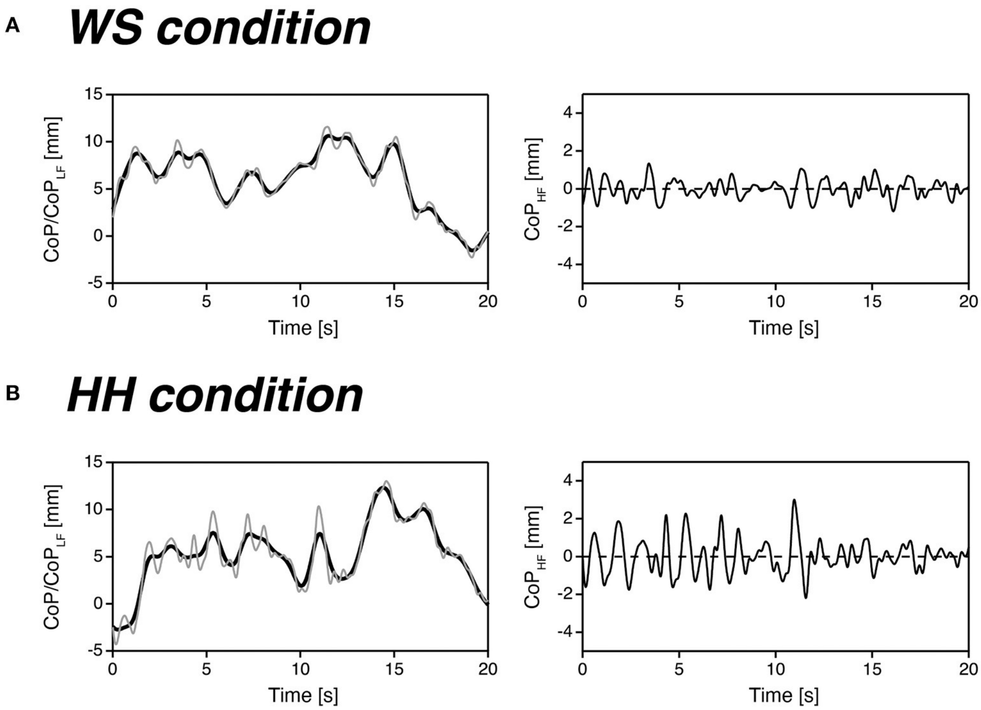

FIGURE 4 | Representative examples of the CoP (*thin gray line*) and CoP_{LF} (*bold black line*) recordings (*left*) and corresponding CoP_{HF} recordings (*right*) for one non-wearer in the WS **(A)** and HH **(B)** conditions.

neurophysiological properties of the anterior calf muscles. For example, because the ankle joint is plantarflexed during standing with high-heels, increased Ia discharges is expected from the stretched anterior calf muscles. Loram and colleagues (Di Giulio et al., 2009; Loram et al., 2009) have suggested that passive and unmodulated anterior calf muscle (e.g., tibialis anterior), uncomplicated by fluctuations in muscle activity, enables a better proprioception of small, joint rotation during quiet standing. Although it is uncertain whether the expected increase in Ia discharges from the anterior calf muscles interferes with proprioception as mere background noise or enhances it via a mechanism known as stochastic resonance (Cordo et al., 1996), high-heel wearing may have a certain influence on the proprioception of postural sway.

Comparison With Previous Studies

Hapsari and Xiong (2016) examined the effects of heel height and high-heel experience on static balance, reporting that wearing high heels with a heel height up to 7 cm did not impair postural stability, irrespective of high-heel experience. Similarly, Wan et al. (2019) reported no significant differences in CoP_{RMS} among four different heel conditions (1, 5, 8, and 10 cm) when habitual wearers and non-wearers were analyzed together. In addition, the authors reported no significant differences in CoP_{RMS} between high-heel wearers and non-wearers in 1 and 5 cm heel conditions (Wan et al., 2019). These CoP_{RMS} results are consistent with the current findings. However, there are several discrepancies in CoP_{MV} results between studies. For example, Wan et al. (2019) demonstrated that CoP_{MV} decreased as heel height increased from 1 to 8 cm. In addition, high-heel wearers were reported to exhibit significantly smaller CoP_{MV} in all heel conditions compared with non-wearers (Wan et al., 2019). The difference in the mean age of participants could have potentially caused the discrepancies found between studies. In particular, because the high-heel wearers in Wan et al.'s study were relatively young (mean $\pm$ SD: 24.6 $\pm$ 2.1 years), it is possible that these participants were not fully accustomed to wearing high heels (Wan et al., 2019). Therefore, the results reported by Wan et al. (2019) may reflect the transition process of becoming accustomed to wearing high heels. The high-heel wearers recruited in the present study were substantially older (31.4 $\pm$ 6.7 years) than those in Wan et al.'s study (Wan et al., 2019). Thus, it is likely that most of the high-heel wearers in the present study were fully accustomed to wearing high heels. Differences in the methods of measurement and analysis may have also contributed to the discrepancies between the studies. For example, Wan et al. (2019) measured the CoP position using an insole measuring system, whereas we measured it using a standard force platform system. Furthermore, the previous study did not perform any filtering before calculating the CoP_{MV}. This may have resulted in the large values of CoP_{MV} reported in that study (~20 mm/s). The CoP_{MV} of healthy young adults during quiet standing was reported to be around 5–6 mm/s (Prieto et al., 1996; Ushiyama and Masani, 2011), which is comparable to the values observed in the present study (note that unnormalized data are not reported in the present study).

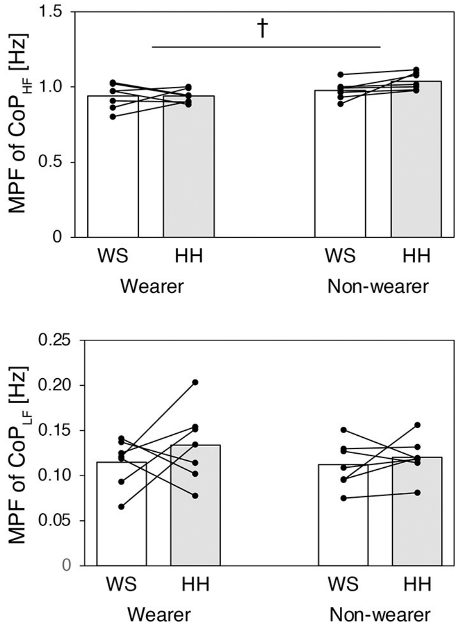

FIGURE 5 | The MPF of CoP_{HF} (*top*) and CoP_{LF} (*bottom*) for each group and each test condition. The white bar indicates the WS condition and the gray bar indicates the HH condition. The small dots represent data for individual participants. $^{\dagger}P < 0.05$ between the groups.

Limitations and Future Research

It should be noted that the CoP position in the AP direction is proportional to the ankle plantarflexion torque (Winter et al., 1998). Therefore, analyses of CoP sway only provide information regarding control around the ankle joint. Although the ankle joint plays a crucial role in controlling static balance (Winter et al., 1998; Masani et al., 2003), several recent studies have indicated significant contributions of the proximal joints (e.g., the hip and knee) even in an unperturbed stance (Sasagawa et al., 2009a, 2014; Yamamoto et al., 2015). In addition, Hapsari and Xiong (2016) reported that participants employed more hip strategy (less ankle strategy) to control static balance as the heel height of the shoes increased. Although the present study did not measure the CoP sway in the ML direction, this increased hip contribution may affect the postural sway in the ML direction. Therefore, to fully understand the changes in static balance control induced by occasional or habitual wearing of high heels, future studies should include three-dimensional, whole-body motion analysis and pay attention to multi-joint coordination.

CONCLUSIONS

In the present study, we examined the effects of occasional and habitual wearing of high heels on static balance by comparing the CoP sway between high-heel wearers and non-wearers while standing with and without high heels. We first found that wearing high heels with a heel height of 7 cm did not increase the magnitude of postural sway, irrespective of high-heel experience. We also found that wearing high heels with heel height of 7 cm increased the amount of regulatory activity in both high-heel wearers and non-wearers. We further found that habitual high-heel wearers showed decreased rate of regulation of the ankle joint torque than non-wearers, both while standing with and without high heels. These results suggest that use-dependent changes in static balance control are evident in both high-heeled and without shoes conditions.

DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding author.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethics Committee on Human Experimentation at the Graduate School of Arts and Sciences, the University of Tokyo. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

AY-Y and NI: conceptualization. AY-Y: investigation. AY-Y and SS: formal analysis and writing—original draft preparation. AY-Y, SS, KN, and NI: writing—review and editing. All authors contributed to the article and approved the submitted version.

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Skill-Related Adaptive Modifications of Gaze Stabilization in Elite and Non-Elite Athletes

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The vestibular ocular reflex (VOR) provides gaze stability during head movements by driving eye movements in a direction opposing head motion. Although vestibular-based rehabilitation strategies are available, it is still unclear whether VOR can be modulated by training. By examining adaptations in gaze stabilization mechanisms in a population with distinct visuomotor requirements for task success (i.e., gymnasts), this study was designed to determine whether experience level (as a proxy of training potential) was associated with gaze stabilization modifications during fixed target (VOR promoting) and fixed-to-head-movement target (VOR suppressing) tasks. Thirteen gymnasts of different skill levels participated in VOR and VOR suppression tasks. The gain between head and eye movements was calculated and compared between skill levels using an analysis of covariance. Across experience levels, there was a similar degradation in VOR gain away from −1 at higher movement speeds. However, during the suppression tasks, more experienced participants were able to maintain VOR gain closer to 0 across movement speeds, whereas novice participants showed greater variability in task execution regardless of movement speed. Changes in adaptive modifications to gaze stability associated with experience level suggest that the mechanisms impacting gaze stabilization can be manipulated through training.

Keywords: vestibular ocular reflex, gymnastics, cancellation, suppression, training, gaze, visuomotor

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INTRODUCTION

Experienced coaches who train athletes to perform aerial acrobatics skills (i.e., gymnastics, diving) have long observed the importance of visual cues (Hondzinski, 1998; Davlin et al., 2004). The regulation of flight time and an estimation of landing (or time of collision T_c) are the key factors. Full vision improves landing success, and a prospective control strategy uses not only take-off parameters in a feedforward manner, but also vision as feedback to determine landing time (Bardy and Laurent, 1998). Novice athletes sometimes ignore or are unaware of these cues and rely on timing to complete aerial acrobatics successfully. This is considered an open-loop control strategy and is often used in lower-level skill execution. However, when they start to do more advanced aerial activities, visualization of the landing area becomes increasingly important to confirm a safe and stable landing (Davlin et al., 2004). Flipping (lower level) skills are defined as aerial rotations around a medial–lateral axis. When twisting (higher level) is added to these aerial activities (rotation around a superior–inferior axis), visualizing the landing zone becomes increasingly complicated since head motion is now multi-axial. Vestibular information about the orientation of the head relative to the ground is lost during flight (Bardy and Laurent, 1998). Experience and training not only improve the development of gaze stabilization, but also promote skill progression and the successful execution of higher-level, multiple rotation, and multi-axis activities. This paper describes the relationship between training and the ability to manipulate gaze stabilizing eye movements that are counter to the gain of the vestibular ocular reflex (VOR).

The VOR facilitates a stable visual field and reduces retinal smear in tasks that necessitate head motion by driving eye movements in a direction opposite to those of the head (Baloh et al., 1984). However, when executing aerial acrobatics skills requiring multi-axis rotations that include both active and passive had motions, the action of the VOR may produce performance decrements since the landing surface may be outside the field-of-view (FOV) several times during skill performance. Specifically, when the landing surface is outside the FOV (behind or below the athlete), the VOR would delay the ability to see this surface since eye gaze would be driven in a direction opposite to the twist (away from the prospective landing surface). In this paper, the term VOR supression (VORs) is used to describe an eye movement that is counter to the normal VOR. VORs would bring that surface into the FOV sooner and facilitate a faster acquisition of the area of interest as it comes into view. This circumstance would promote gaze toward the landing site and may provide a landing time estimate to the participant. The ability to modify gaze stability has been found to occur in populations who are frequently exposed to twists and aerial acrobatics (e.g., dancers, ice-skaters, and pilots during specific tasks) (Lee et al., 2004; Alpini et al., 2009; Maheu et al., 2019). The presence of context-specific control of VOR gain provides evidence that VOR can be manipulated with exposure and training. How VORs is promoted with graded exposure to multi-axial twisting and flipping (i.e., gymnastics) as a function of skill level is unknown. Higher skill level athletes are typically exposed to activities that require higher angular velocities to complete successfully. The presence of context-specific control of VORs in this population may provide evidence that VOR gain can be manipulated with exposure and training.

In an effort to understand how exposure and training affect the adaptive modification of gaze stability, this exploratory study investigates how VOR gain changes during classic and suppression-based visuomotor tasks at 7 different head velocities in athletes trained to perform multi-axial airborne rotational skills. Specifically, we predict that there is a positive correlation between an athlete's skill level and VOR gain (i.e., as experience level decreases, associations between VOR gain and head velocity would deviate from the optimal values of −1 for classic visuomotor task and 0 for VOR suppression visuomotor task) as evidenced by either a rotation (**Figure 1**—shaded) or translation of the regression line of this relationship (**Figure 1**—arrows).

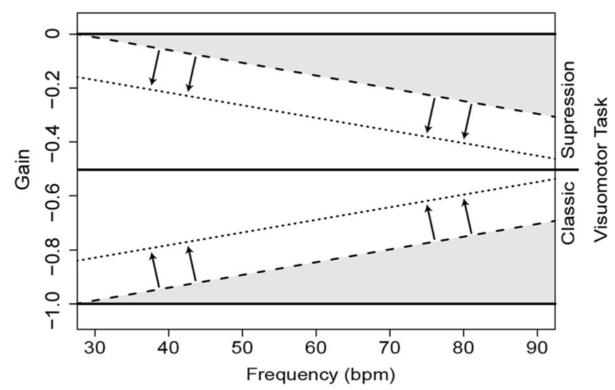

FIGURE 1 | Prediction schema of VOR gain relationship changes as a function of experience level and movement speed. Solid lines represent an optimal performance across tasks as movement speeds increase for the VOR suppression (top) and classic visuomotor task (see bottom). Decrements in performance due to experience could occur through rotation of the regression line (e.g., dashed lines and shaded portions), with steeper slopes indicating poorer performance, or through translation of the regression line (e.g., dotted lines, see arrows), representing a shift in baseline performance.

TABLE 1 | Participant demographics.

Subject	Age	Twist direction	Hand dominance	JO level
01	16	R	R	10
02	20	R	R	10
03	14	R	R	8
04	16	R	R	8
05	16	R	R	10
06	18	L	R	5
07	14	L	R	8
08	13	L	R	7
09	13	L	R	7
10	15	L	R	5
11	27	L	R	8
12	23	L	R	9
13	26	R	R	9

Level represents competition level based on USA Gymnastics Junior Olympic (JO) Program, with higher numbers indicating a more advanced gymnast.

METHODS

Participants

In this study, gymnasts with a skill level of 5 have limited training or exposure to aerial skills. They act as a control group comparison. At Level 6, uniaxial aerial skill is introduced. As gymnasts progress to Level 10, the complexity of the aerial skills increases and these athletes are required to complete 3 different kinds of flips and an acrobatic series with at least two connected flips.

Thirteen female gymnasts (21.0 ± 5.7 year) took part in this preliminary study (**Table 1**). Their ability level ranged from novice to elite with seven participants of the 13 still active in competition. They were consented on-site and told that they could exit the study at any time without recourse. The study was approved by the Internal Review Board of Virginia Commonwealth University (IRB Approval HM2M0008030). Age, preferred twist direction, hand dominance, and level of competition were acquired *via* self-report.

Experimental Setup

All data were collected using a kinematic system to monitor head and trunk position with 6 degrees-of-freedom and a binocular eye dark-pupil tracking system to monitor horizontal and vertical eye position. Data collection with these two systems was integrated providing a synchronous dataset. The participants were seated on a wooden stool at the end of a 0.76-m-high workbench. They were asked to wear a helmet fitted with a camera-based infrared eye-tracker (EyeLink II, SR Research Ltd. Mississauga, Ontario Canada). The eye-tracker had a tracking range of $\pm$ 30° in the horizontal direction and $\pm$ 20° in the vertical direction with resolution of 0.01°. These data were collected at 250 Hz and provided eye-in-head position. Head and trunk position were collected from two electromagnetic (EM) motion sensors (Motion Monitor™, Innovative Sports Training, Chicago, Illinois, USA). One EM sensor was on the helmet that also held the eye-tracking hardware and provided head-in-space position. The second was mounted on a strap worn around the participant's thorax providing a head-on-trunk position. Data were collected from the kinematic system at 100 Hz.

A transmitter was located behind the subject and provided orthogonally oriented electromagnetic fields. The theory of operation for this type of sensor is that it orients itself based on transmitter field strength. The field strength decreases as a square of the distance from the transmitter. The transmitter used in this application had a functional radius of 3 m. The participants (and sensors) were located well within this range. Care was taken to avoid the presence of metal within the transmitter field to minimize eddy current distortions. To further reduce any effect of metal, a mapping procedure was performed prior to data collection. This metal mapping procedure uses known sensor location data to construct a distortion map of the collection space. These data are then used to linearize any measurement error across the mapped space with a resolution of 0.5 mm linearly and 0.1° angularly. All collected data were stored in a coded file using subject initials, the date, and trial number or descriptor.

Participant Setup

After securing the helmet and sensors, the subject was asked to sit steady and face an LED display placed on the workbench at a distance of 1 m. The subject was positioned at eye level with the center target on this board. A world-based right-hand coordinate system was defined on the workbench top. The origin was located to the participant's right. Subject landmarks were defined in a prescribed manner using a stylus fixed to a free-floating sensor. These standardized locations allowed the reconstruction of a rigid body model of the subject for interpretation of position data. Left–right symmetry was assumed.

Visuomotor Task

Classic Condition: Static Target Fixation (VOR)

In the classic visuomotor condition, participants were seated at the end of a cloth-draped tunnel to remove the visual scene and visually fixated on a single LED target located centrally on the calibration screen and subtended to a visual angle of 0.03°. Participants were then instructed to actively rotate their head in the transverse plane (yaw or "no" motion) while continuing to fixate on the LED target for an approximate movement amplitude of $\pm$ 15° from neutral head position. Active head rotation was selected for this experiment since it mimicked the *in situ* performance environment of the gymnast. Movement speeds were dictated by a metronome set at seven different frequencies ranging from 72 to 196 beats/min with increments of 20 beats/min. This combination of speed and amplitude resulted in angular movement velocities between 35° and 100°/s. Participants were verbally encouraged to maintain the prescribed angular head range during the higher tempos to provide a full range of angular head velocities. Following each trial of 5 complete rotation cycles, head range of motion was reviewed, and the trial repeated if necessary. A typical example of these data is illustrated in **Figure 2A**. Angular velocities of the head (from the head-mounted sensor) and eye were calculated using a Savitzky–Golay differentiation filter. After removing any stationary head movement data first (<3° amplitude), VOR gain was calculated *via* a custom MATLAB script as the average of a point-by-point ratio between head and eye velocities (**Figure 2B**).

VOR Suppression Condition: Fixation of Head-Mounted Target

In the VOR suppression condition, the calibration display was replaced with a black trifold foam board. A laser pointer was fitted on the participant's helmet, so the red dot appeared at the center of the trifold foam. This configuration fixed the laser generated target to head motion. The target subtended a visual angle of ∼0.01°. Participants then repeated the same movement amplitude and speed conditions of the classic condition described above, while maintaining stable gaze at the target created by the head-fixed laser pointer. A typical example of these data is illustrated in **Figure 3** for position (**Figure 3A**) and angular velocity (**Figure 3B**) for the head and eye.

Data Analysis

Horizontal eye and head angular data were used to compute the VOR gain. Any data offset was removed by subtracting the mean of 500 ms of pretrial non-movement initial data from the entire trial. Left and right head movement data were separated for analysis. Trials missing data were removed from the analysis, 3 trials for the VOR condition within one participant (13) due to failing to maintaining the angular velocity at the higher metronome speeds, and 6 VOR suppression trials for 2 participants (13 and 01) due to difficulty maintaining metronome speed (13), or eye tracking errors (01). In total, a remainder of 88 VOR and 85 VOR suppression trials were analyzed. Task initiation and termination were defined as the first and last instance the absolute head angle exceeded 3° to remove static eye

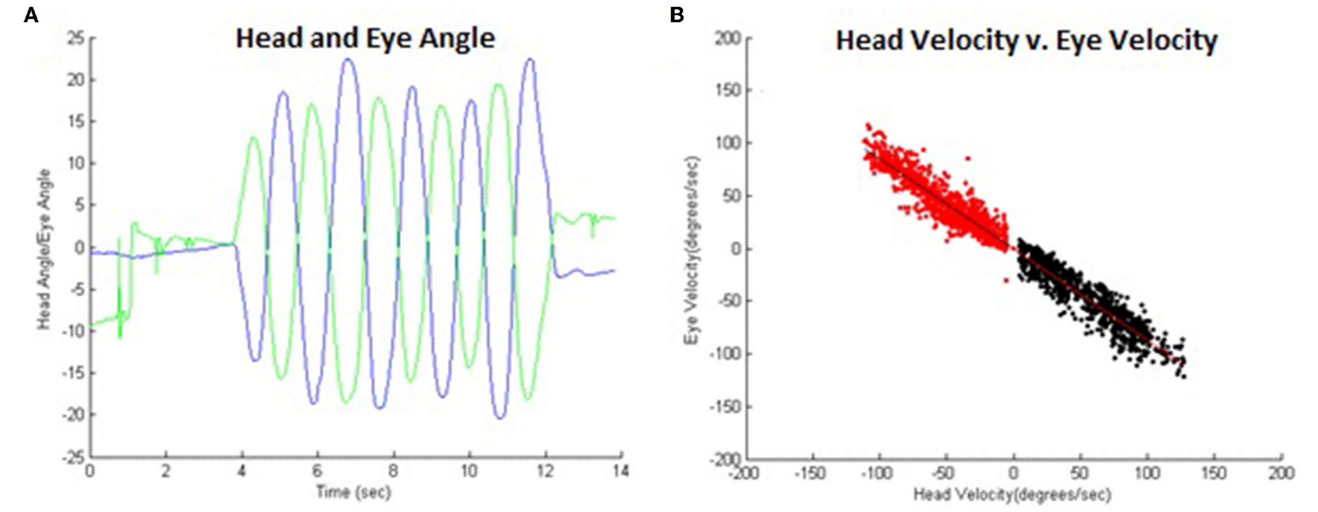

FIGURE 2 | (A) Typical example of VOR temporal head (blue) and eye (green) angle data. **(B)** These same data comparing head vs. eye angular velocity illustrating VOR. Leftward (black) and rightward head movements (red) are separated for analysis with head velocities below 5°/s excluded since the VOR response is negligible at low head velocities. The VOR gain was ∼-0.8 in this example.

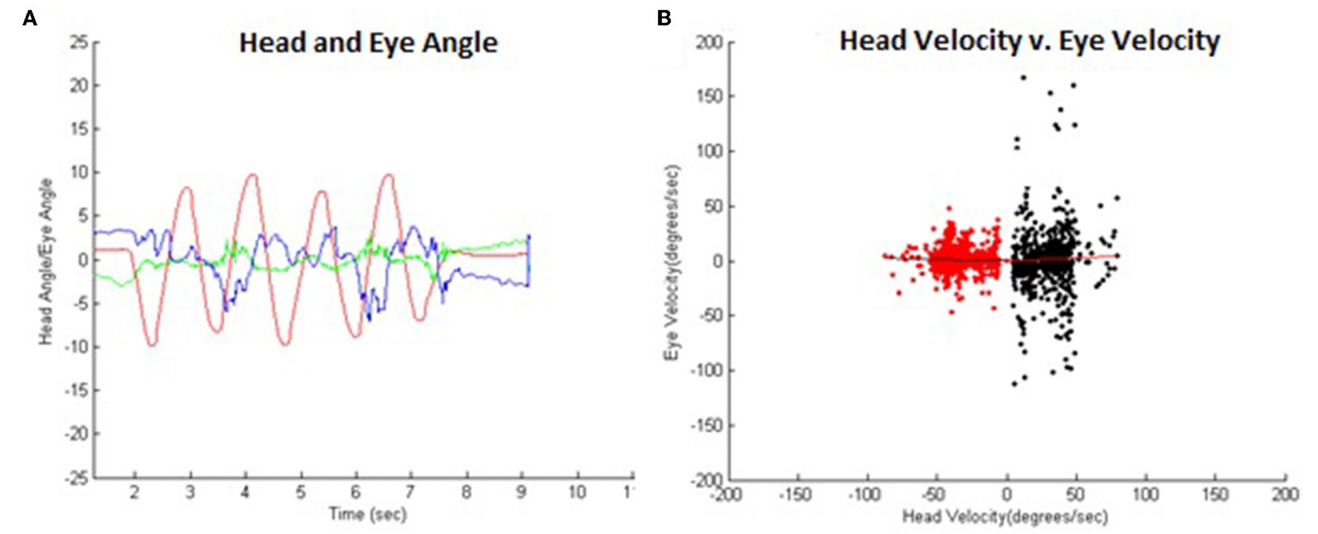

FIGURE 3 | (A) Typical example of VOR suppression temporal head (red), right eye (blue), and left eye (green) angle data. **(B)** These same data comparing head vs. eye angular velocity illustrating VOR suppression. Leftward (black) and rightward head movements (red) are separated for analysis with head velocities below 5°/s excluded since the VOR response is negligible at low head velocities. The VOR gain was approximately zero in this example.

and head data from the trial (represented in **Figures 2B**, **3B** as the band around reflecting turning points of head motion). The range of head movement was calculated as the average amplitude of the head oscillation (VOR 29.1 ± 8.4°, VOR suppression 29.9 ± 7.3°). The frequency of the movements was calculated as the number of oscillations made in the time between task initiation and termination. Angular velocity was calculated by determining the time rate of change of the head and eye angular position data. As mentioned previously, VOR gain was calculated *via* a custom MATLAB script as the average of a point-by-point ratio between head and eye velocities.

Statistical Analysis

Correlations for repeated measures were undertaken using the rmcorr package in R (version 0.4.1, Bakdash and Marusich, 2017). Briefly, this method accounts for non-independence among observations using analysis of covariance (ANCOVA) to statistically adjust for interindividual variability. Correlation coefficients accounting for repeated measures (r_{rm}) are presented, with 95% confidence intervals estimated *via* bootstrapping (number of iterations = 1,000).

To test whether the relationship between VOR gain and movement frequency changed as a function of experience level, significance testing was undertaken on regression coefficients (i.e., slope). Coefficients were first normalized using Fisher's z-transformation, with the difference between z-transformed coefficients compared with a critical z-score (Suzuki et al., 2008; Weaver and Wuensch, 2013). To account for multiple comparisons, Bonferroni adjustments were applied (such that critical $Z = 2.58$, $p < 0.005$).

RESULTS

Angular Head Velocity

Classic Condition: Stationary Visual Target

Associations between VOR gain and experience levels across all movement speeds showed few changes in task accuracy as experience level increased (**Figure 4A**, top panel) despite clear changes in the proportion of trials closer to optimal for the most elite and novice groups (i.e., Level 10 vs. Level 5). When examined on a participant basis (**Figure 4A**, bottom panel), VOR gain and movement speed showed a strong positive relationship ($p < 0.001$) regardless of experience level with intercept values close to −1 (−1.12 to −0.88) with a positive slope (0.0013–0.0038), which indicates relative deterioration in task execution as movement speeds increased.

VOR Suppression: Head-Fixed Target

In contrast to the classic visuomotor task, VOR gain showed a significant negative relationship to frequency ($p < 0.001$), with, greater changes as a function of experience level in the VOR suppression task (**Figure 4B**, top panel). This could be largely attributed to the greater variability and poorer execution shown by novice participants (e.g., Level 5 values deviating from 0). When compared to movement speed (**Figure 4B**, bottom panel), higher experience levels (i.e., Levels 9 and 10) showed a greater maintenance of successful VOR suppression even with increasing movement speed, with regression coefficients (i.e., slope) and y-intercepts nearing zero (e.g., Level 10, regression coefficient = −0.0008, y-intercept = −0.09). This contrasted with novice participants, who showed a general inability to successfully suppress VOR across the range of movement speeds (as indicated by a shift in the regression line and deviation of y-intercept away from 0, i.e., Level 5 = −0.26). Despite the magnitude in changes between regression coefficient values between experience levels, statistical analysis showed that there was insufficient evidence to confirm such differences. In contrast, significant differences found between experience level in VOR suppression intercepts and slopes were seen between Level 5 vs. Level 7 ($p < 0.001$) and Level 8 vs. Level 9 ($p < 0.001$).

DISCUSSION

This study investigated whether gymnasts of differing experience levels exhibited adaptations in VOR gain that reflected task requirements. VOR gain showed characteristic changes during the VOR suppression visuomotor task that tended to align with our expected predictions (refer to **Figure 1**), such that expert participants were able to maintain VOR gain values closer to zero compared to their novice counterparts, even at higher movement speeds. VOR suppression is a necessity for the aerial acrobatics part of the competition at these higher experience levels (i.e., Levels 9 and 10). VOR suppression, however, is likely to decline at higher speeds or in lower skill level gymnasts as exposure and demand of training are likely to have been less demanding of the skill to suppress VOR. Interestingly, VOR gain during the classic visuomotor did not show the same changes across experience level, which suggests that certain visuomotor task may have greater utility for training purposes.

These data showed that all gymnasts were able to achieve VOR gains near −1 at the low head velocities in the fixed visual target task; however, this relationship deteriorated at higher angular velocities. These results are similar to the general public and were expected, as VOR gains below an absolute 0.68 have been proposed as abnormally low (MacDougall et al., 2009). No statistical difference in VOR performance was found between gymnastics skill level. VOR suppression (or VORs), however, did show some statistical differences as a function of skill level. The higher skill level athletes had more stable VOR suppression gains ($r = 0.019$) and were able to maintain this VOR suppression (0.33 ± 0.48) throughout the range of imposed head angular velocities, as shown in the smaller slope. The lowest level athletes were more variable ($r = 0.208$) and showed a limited ability for VOR suppression (mean ± SD 0.17 ± 0.38) across head velocities. The intermediate skill levels showed a graded response in VOR suppression (0.22 ± 0.42) when compared to the expert and recreational athletes. This could be the reason why expert gymnasts are more reliant on visual information during salto (backward aerial summersaults) than novice gymnasts as described by Bardy and Laurent (1998).

Besides skill level, the ability to suppress VOR could be attributed to age as skill level and exposure to multi-axial twists increase with age in our population. However, evidence from student aviation pilots showed greater VOR gains (i.e., closer to −1 or 1) after training than students before training (Lee et al., 2004). Besides these results supporting the plasticity of the VOR, Lee et al. did not find any relationship between age and VOR. The development of VOR has shown to occur rapidly until the age of 6 years, when it then progresses at a slower rate to adult values by 16 years (Wiener-Vacher and Wiener, 2017). Although gymnast younger than 16 years of age were included in the experiment, VOR gains ranges for the gymnasts with ages below 16 were not different than the group older than 16 (−0.80 to 80 versus −0.71 to 0.87 respectively). Additionally, the younger gymnasts show to reach a VOR gain of −80 during the VOR suppression task, further indicating that age is not the driving factor in the ability to suppress VOR.

Because these athletes have a preferred twist direction, it was thought that the ability to suppress VOR would trend in that direction as well (i.e., VOR gain during leftward head motion would be closer to zero than during rightward head movement for the gymnasts that preferred leftward twists and *vice versa*). A secondary analysis showed that this was not supported by the data. No significant differences were found between left and right VOR gains for left and right head motion for either left or right preferred twisting gymnasts.

The correlation of VOR suppression to training dose suggests that VOR gain control can be enhanced through training. The training stimulus in this case is extreme and not directly suited

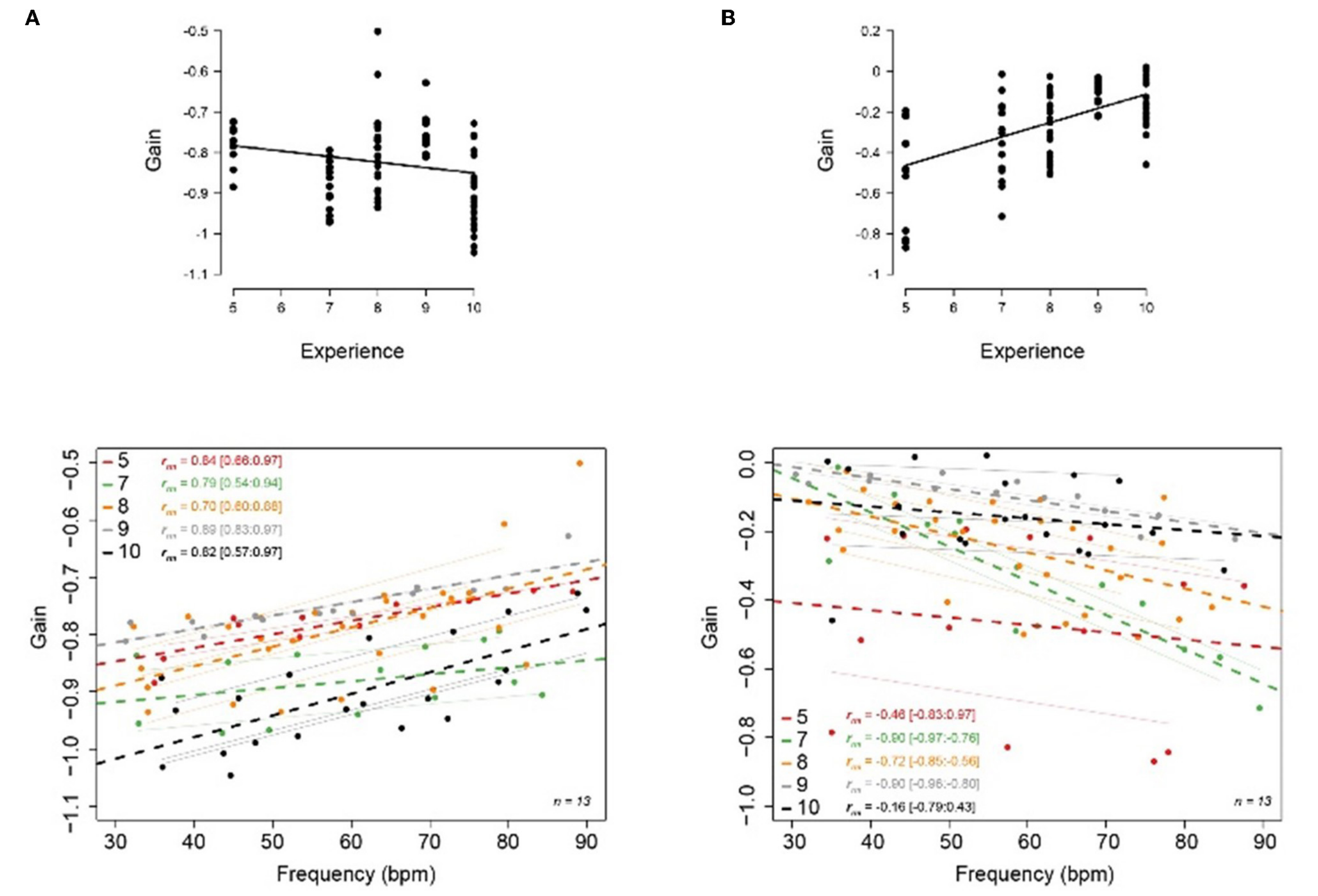

FIGURE 4 | Regression lines comparing head frequency with VOR gain during head rotation for the different skill levels (colored) and participants (light gray) for the **(A)** classic condition and **(B)** VOR suppression condition. The top panels represent group average, over all participants and the different velocities, the bottom panels represent gain regression lines for the groups (in color) and per gymnast (gray).

for the treatment of patients suffering from concussion or traumatic brain injury (TBI). However, from a neurophysiologic standpoint, it may be important to alter VOR gain in certain circumstances to enhance visual input. Understanding this process in trained athletes has implications in the treatment of conditions that affect gaze stabilization ability. The gain of head and eye movements is disturbed resulting in retinal smear, which often causes symptoms of nausea and imbalance during movement (Crawford and Vilis, 1991; Paige, 1994; Gurley et al., 2013), for example, the cause of gait and balance problems following TBI and concussion (Parker et al., 2005, 2006; Kaufman et al., 2006; Pickett et al., 2007; Slobounov et al., 2008; Wares et al., 2015). The results of this study support the application of exposure and training to adapt the VOR to reduce these symptoms. However, the existing rehabilitation tools for training VOR gain are based on the qualitative estimates of head movement and gaze maintenance during treatment. In addition, patient reports of dizziness and nausea, often used to determine progression, can be problematic due to the variability of this metric relative to VOR gain. The lack of precision and standardization in evaluation and treatment of impairments in VOR gain has most likely contributed variability in treatment outcomes (Slobounov et al., 2008). Investigating how elite and novice athletes adapt their VOR may help us to understand how the integration of the sensory systems can be adapted in populations with poor balance. Therefore, this study will focus on VOR gain and VOR suppression in athletes trained to perform multi-axial airborne rotational skills.

LIMITATIONS

Real plasticity and adaptability of the VOR can only be assessed in a longitudinal study, following VOR suppression ability in athletes such as gymnast's through their development and progress from recreational to elite status or follow a patient population with VOR deficits during the rehabilitation period. However, this study indicates gymnasts increase in the ability to suppress the VOR due to more elaborate exposure to airborne activities.

This is an exploratory study, and only 10 gymnasts were included in the recent study, which allows for 2 or 3 gymnasts per skill group. This is probably the cause of the limited significant results in this study. However, the trends and significant results present in the current work align with our initial prediction (see **Figure 1**) and reinforce the notion that mechanisms impacting gaze stabilization may be manipulated through training.

CONCLUSION

The ability to suppress or cancel VOR increased with the level of experience in gymnastics and airborne skills. Therefore, it is likely that VOR can be trained or manipulated by exercises. However, further investigation is needed in the longitudinal development of VOR and VOR suppression in gymnasts and the effect of training populations with VOR deficits.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Internal Review Board of Virginia Commonwealth University. Written informed consent to participate in this study was provided by the participants' legal guardian/next of kin.

AUTHOR CONTRIBUTIONS

SV is responsible for the initial data analysis and manuscript preparation. AS is responsible for statistical and manuscript proof reading. JT and PP are responsible as laboratory managers and PI's on the project who proof read the manuscript.

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The Associations Among Self-Compassion, Self-Esteem, Self-Criticism, and Concern Over Mistakes in Response to Biomechanical Feedback in Athletes

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Athletes regularly face the possibility of failing to meet expectations in training and competition, and it is essential that they are equipped with strategies to facilitate coping after receiving performance feedback. Self-compassion is a potential resource to help athletes manage the various setbacks that arise in sport over and above other psychological resources. The primary purpose of this research was to explore how athletes respond to objective biomechanical feedback given after a performance. Specifically, we investigated if levels of self-compassion, self-esteem, self-criticism, and concern over mistakes were related to one another before and after a series of sprint tests interspersed with biomechanical feedback, and whether self-compassionate athletes achieved a better sprint performance after receiving and implementing biomechanical feedback. Forty-eight athletes (20 female: $M_{age} = 19.8$ years, $SD = 3.1$; 28 male: $M_{age} = 23.6$ years, $SD = 7.8$) completed online measures of self-compassion, self-esteem, self-criticism and concern over mistakes before performing four sets of 40-m sprints. Participants received personalized biomechanical feedback after each sprint that compared their performance to gold standard results. Following all sprints, they then completed measures of self-criticism, and reported emotions, thoughts, and reactions. Self-compassion was positively correlated with self-esteem ($r = 0.57$, $p < 0.01$) and negatively related to both self-criticism ($r = -0.52$, $p < 0.01$) and concern over mistakes ($r = -0.69$, $p < 0.01$). We also found that athletes with higher levels of self-compassion prior to sprint performance experienced less self-critical thoughts following biomechanical feedback and subsequent sprint trials ($r = -0.38$, $p < 0.01$). Although the results of this study provide some support for the effectiveness of self-compassion in promoting healthy emotions, thoughts, and reactions in response to sprint performance-based biomechanical feedback, a moderated regression analysis between the first and fourth sprint time variables revealed that self-compassion was not a moderator for change in sprint performance ($R^2 = 0.64$, $\Delta R^2 = 0.10$, $p > 0.05$). These findings suggest that there are likely longer-term benefits of athletes using self-compassion to cope with biomechanical feedback, but that any benefits might be limited in a short series of sprint trials.

Keywords: sport psychology, self-compassion, self-criticism, self-esteem, concern over mistakes, biomechanical feedback, sprinting, athletes

INTRODUCTION

The interplay between an athlete's body and mind can have a significant impact on overall sports performance, particularly when athletes are expected to efficiently absorb and implement feedback within the competitive environment. Previous evidence suggests that feedback can significantly improve sports performance (Baudry et al., 2006; Mauger et al., 2009). While feedback can create desirable outcomes in sports performance, coping with feedback and effectively executing biomechanical improvement during competition and training can be challenging for some athletes (Mononen et al., 2003). As athletes regularly face the possibility of failing to meet expectations in training and competition (Gustafsson et al., 2017), it is essential that they are equipped with the dispositions and strategies that facilitate coping after receiving performance feedback. Although researchers have extensively examined the significant role certain psychological characteristics (e.g., resilience, self-belief, optimism, etc.) play in the development of sports expertise (Gould et al., 2002; MacNamara et al., 2010; MacNamara and Collins, 2015), little is known about how these psychological characteristics might facilitate an athlete's response to feedback given about their sport performance.

Biomechanical feedback provides objective technical information with the purpose of enhancing performance (Harfield et al., 2014), and it can be given about a variety of performance outcomes using several types of modalities. The feedback that is specifically related to performance results is known as *outcome feedback* (Salmoni et al., 1984). Outcome feedback can either be intrinsic or augmented—*intrinsic feedback* is the information provided by an athlete's own sensory and perceptual systems (McGill, 2001), while *augmented feedback* provides information that athletes do not receive from their sensory systems and is provided by an external source (Utley and Astill, 2008). Augmented feedback is often provided to athletes by coaches or trainers (Schmidt and Wrisberg, 2008). For example, coaches can give verbal information to athletes based on their observations in practice or comment on a sprinter's running technique from objective measurements (e.g., timing gates, video analysis, movement sensors, etc.). Augmented feedback can be provided using a variety of modalities—visual (e.g., screen), auditory (e.g., speaker), haptic (e.g., vibrotactile actuator), or a combination of the above (Akamatsu et al., 1995).

Augmented feedback can be further classified into *knowledge of result* (KR) and *knowledge of performance* (KP). KR is feedback about goal achievement (e.g., the time it takes to run a certain distance), while KP feedback is directed toward movement characteristics that influence performance outcomes (e.g., a runner's step length or frequency; Mononen et al., 2003). KR feedback is usually provided for tasks that require scaling of a single degree of freedom movement or a single dimension response (Salmoni et al., 1984). KP feedback usually provides information about the kinematics or kinetics of a movement (Newell and Carlton, 1987). The type of feedback an individual receives can shift the attentional focus of athletes. For example, receiving feedback about movement effects or outcomes (i.e., KR feedback) could result in an external focus, while feedback provided on body movements and movement coordination (i.e., KP feedback) might direct athletes' attention internally (Wulf, 2007). Due to the complexity of certain sports skills, KR feedback might not be the most effective type of feedback to provide to athletes, as it may prevent individuals from using intrinsic feedback processing and error detection (Salmoni et al., 1984). Consequently, athletes who are given KP feedback that addresses the kinematics of sports skills are more likely to experience successful sport outcomes (Schmidt and Lee, 1999).

One challenge faced by sport researchers, coaches, and trainers alike is that athletes may receive and implement feedback differently. More specifically, some athletes might perceive feedback negatively and experience a setback. The ability to overcome such a setback is essential for athlete success. Therefore, it is likely of great interest to various stakeholders (e.g., coaches, athletic directors, sport organizations, etc.) that athletes are equipped with the dispositions, skills, and resources to persevere when setbacks do occur, which would ultimately enhance athlete sporting experience, performance, and overall well-being.

Self-esteem has been acknowledged as a resource for athletes experiencing setbacks (Neff and Vonk, 2009; Mosewich et al., 2011). Self-esteem is "an evaluation of our worthiness as individuals, a judgment that we are good, valuable people" (Neff, 2011: p.1), and has been established as a pathway to fostering positive sport experiences (Boyer, 2007) and overcome negative challenges (Neff and Vonk, 2009). Despite the benefits of self-esteem in promoting positive self-evaluations (Boyer, 2007), happiness (Baumeister and Vohs, 2018), and self-confidence (Tilindiene et al., 2014) in athletes, relying solely on self-esteem might not be ideal (Mosewich et al., 2011; Neff, 2011). High levels of self-esteem are predicated on the "better-than-average effect" in that unrealistic positive self-evaluations are created through the process of putting down others to boost yourself up (Neff, 2011). Furthermore, enhancing self-esteem is difficult as it is resistant to change (Swann, 1996), and attempts to increase self-esteem are often unsuccessful (Neff, 2003a).

Self-compassion is understood as being kind and understanding toward oneself when faced with personal shortcomings and weaknesses (Neff, 2003a), and has been suggested as a potential alternative to self-esteem in helping athletes cope with some of the difficult challenges they might endure in sport (Mosewich et al., 2013; Ferguson et al., 2014; Sutherland et al., 2014). Though self-compassion and self-esteem are complementary concepts that are significantly and positively correlated with one another (e.g., $rs = 0.56–0.59$: Neff, 2003a; Leary et al., 2007), self-compassion has been shown to predict unique variance in well-being beyond self-esteem among athletes (Mosewich et al., 2011). Furthermore, self-compassion is positively associated with autonomy, environmental mastery, self-acceptance (Ferguson et al., 2021), and well-being (Ferguson et al., 2014), and has been significantly correlated with increased perceptions of athletic performance (Killham et al., 2018). The relationship between athletic performance and self-compassion might be due to highly self-compassionate athletes' decreased levels of self-criticism and fear of failure (Mosewich et al., 2011, 2013).

Self-compassion may be a valuable resource to attenuate setback experiences, such as receiving feedback that may be perceived as performance mistakes. For example, evidence suggests that self-compassion interventions led to decreased self-critical thoughts and concern over mistakes (Jopling, 2000; Gilbert and Procter, 2006). Leary et al. (2007) measured undergraduate students' responses to emotionally difficult situations encountered in their daily lives and concluded that self-compassion explained unique variance beyond self-esteem when predicting an individual's adaptive emotions, thoughts, and reactions to negative or emotionally difficult scenarios. Self-compassion can also play an important role in emotional distress regulation relative to a sports failure (Ceccarelli et al., 2019). Reis et al. (2015) asked 101 women athletes to respond to hypothetical and recalled sport events to examine the effect self-compassion might have during emotionally difficult sport situations. Results indicated that the emotions, thoughts, and reactions of highly self-compassionate women athletes were significantly more adaptive than their less self-compassionate peers. Mosewich et al. (2013) also investigated the effects of a self-compassion intervention on self-criticism and concern over mistakes in highly self-critical women athletes. Findings indicated that a 7-day psychoeducational and applied practice intervention effectively decreased self-criticism, rumination, and concern over mistakes. Moreover, the self-compassion levels of women athletes had significantly increased when measured post-test and during a one-month follow-up ($\eta^2 = 0.43$). With these results in mind, it seems reasonable to speculate that self-compassion might be a relevant resource to help athletes deal with biomechanical feedback, particularly when the feedback presents an emotional challenge.

Study Purposes and Objectives

While both adaptive psychological characteristics and the delivery of feedback can meaningfully contribute to athletic performance, it is important to examine how specific psychological characteristics might influence an athlete's response to feedback during sport-specific skills performance. The current study attempts to bridge sports biomechanics and sports psychology. More specifically, this study aims to explore how athletes emotionally respond to objective biomechanical feedback, and how self-compassion, self-esteem, self-criticism, and concern over mistakes are associated with those responses. A secondary purpose is to explore whether self-compassion can promote adaptive emotions, thoughts, and reactions in response to biomechanical feedback in athletes. As our intention was not to complete an exhaustive evaluation of all the psychological characteristics that can contribute to changes in athletes' responses to biomechanical feedback, we do not claim that the psychological characteristics we chose fully encompass an athlete's experience. However, self-compassion, self-esteem, self-criticism, and concern over mistakes have each been shown to be related to self-evaluation, and thus were best suited to evaluate our study hypotheses.

This study consisted of three groups of hypotheses based on time point and analysis type. Our first group of hypotheses were focused on the relationships between studied psychological variables at baseline. We hypothesized that (1a) pre-trials self-compassion would be negatively associated with baseline self-criticism and concern over mistakes in athletes, and that (1b) pre-trials self-compassion would be positively associated with self-esteem. Hypothesis two addressed the relationships between studied psychological variables before and after sprint testing. We hypothesized that (2a) pre-trials self-compassion and self-esteem would be negatively associated with post-trials self-criticism, and that (2b) pre-trials self-compassion and self-esteem would be positively associated with adaptive post-trials emotions, thoughts, and reactions. We also hypothesized that (2c) pre-trials self-criticism and concern over mistakes would be positively associated with post-trials self-criticism, and that (2d) pre-trials self-criticism and concern over mistakes would be negatively associated with adaptive post-trials emotions, thoughts, and reactions. Our third and final hypothesis was that (3) athletes with higher levels of self-compassion would experience improved sprint performances over baseline after receiving and implementing biomechanical feedback.

METHODS

Methods Overview

This research was an interventional study with a within-between design where data was collected at multiple time-points. Approval to conduct the study was obtained from the institutional Human Research Ethics Board. Participants were recruited via various techniques including sending recruitment letters and posters to sport associations, coaches, and university athletes. Participation was voluntary, and all participants provided written, informed consent indicating that they fully understood the process and purpose of the study. All participants were treated in accordance with the ethical guidelines for human research set forth by the American Psychological Association. The study consisted of two phases. In Phase I, the biomechanical data of nine competitive University-level sprinters (five males, four females) were collected using timing gates and inertial sensors during three video-taped 40-m sprint tests. Sprinting was determined to be the ideal sport skill to examine our research objectives, as running is a fundamental form of human movement and is often used to evaluate athletic performance in elite and non-elite athletes (Lorenz et al., 2013). The most common measurements used to analyze sprinting performance are total sprint time and spatiotemporal stride characteristics (Bezodis et al., 2019), as stride length and stride frequency both play a key role in achieving maximum velocity in sprinting (Paruzel-Dyja et al., 2006). In addition to spatiotemporal parameters, trunk movement can also contribute to sprint performance (Kugler and Janshen, 2010). Phase I participant biomechanical information was used to establish gold standards for time, step length, step frequency, side-to-side (lateral sway), and front-to-back (anteroposterior sway) trunk movements. These gold standards were used as a reference for the biomechanical feedback delivered to participants in Phase II (instead of their own reference standard) with the goal to simulate prescriptive feedback (i.e., how to improve) and replicate learning environments where a coach gives feedback to help an early learner improve. While elite athletes can usually detect and correct their own errors in their primary sports, they might benefit from prescriptive feedback when learning new skills (Magill and Anderson, 2021).

Phase II involved collecting data from the main study participants. It consisted of one 60-min data collection session for each participant. At the start of the session, participant self-compassion, self-esteem, self-criticism, and concern over mistakes was measured via online questionnaires completed at the test site. Participants then performed four sets of sprint tests with timing gates and inertial sensors. The number of sets was chosen to minimize fatigue effects. Participants received biomechanical feedback after each sprint. The feedback contained a visual representation of their trial data in comparison to the gold standards established in Phase I. Participants were asked to improve their next sprinting performance using the biomechanical feedback that was provided (**Figure 1**). After performing all four sprint tests, participants completed a second online self-criticism questionnaire, as well as scales to rate how they reacted, felt, and thought in response to the biomechanical feedback (i.e., Performed Scenario Scale).

Phase I (Gold Standard) Methods

Phase I Participants

To establish gold standards for the Phase II participant feedback, nine competitive collegiate-level sprinters (4 females: $M_{age} = 22.0$ years, $SD = 3.3$; and 5 males: $M_{age} = 20.4$ years, $SD = 2.1$) participated in this study. Separate gold standards were created for males and females.

Phase I Procedures

Data were collected on an outdoor varsity-standard running track. After a warm-up length of their preference, each sprinter wore three inertial sensors on their trunk and ankles (see Phase II Instruments section) and performed three sets of 40-m sprints on a straight path at maximum speed. Participants were given three minutes of rest time between trials. The collegiate-level sprinters did not receive any biomechanical feedback about the trials until after all data had been collected. The data from each participant's fastest trial were used to calculate the gold standard data for Phase II. Separate means for males and females were each calculated for sprint time, stride length, stride frequency, side-to-side sway, and front-to-back sway.

FIGURE 1 | Sample biomechanical feedback presented to Phase II participants. Green bars indicate the sex-matched gold standard data. Red bars indicate the participant's performance for that trial. The rows display participant run time, step length, and cadence (i.e., step frequency). Step length is normalized to the participant's height. For trunk sway, the green arcs show the side-to-side and front-to-back gold standard trunk sways. The red arcs show the participant's sway.

Phase I Results

See **Table 1** for Phase I participant gold standard results.

Phase II (Main Participants) Methods

Phase II Participants

Fifty athletes (22 females: $M_{age} = 19.8$ years, $SD = 3.1$; and 28 males: $M_{age} = 23.6$ years, $SD = 7.8$) participated in this study as main participants. Participants were excluded if they fell outside the age criteria (i.e., 16–35 years old). Participants were also excluded if they had not trained at least once a week within the past year for an organized sport. Two female participants were excluded from the study—one did not meet the age inclusion criteria and the other had incomplete sprint time data. Included athletes were involved with a variety of team sports including basketball ($n = 9$), football ($n = 2$), hockey ($n = 2$), soccer ($n = 25$), and volleyball ($n = 10$). Their current level of competition ranged from recreational (i.e., competing in intramurals or in a recreational league; $n = 15$), local (i.e., competing against athletes from your city/town; $n = 13$), provincial (i.e., competing against athletes from around your province; $n = 8$), regional (i.e., competing against athletes from other provinces; $n = 8$), national (i.e., competing at national championships; $n = 2$), and elite (i.e., competing at an international level, either against athletes of the same age group or for your country; $n = 2$). The

TABLE 1 | Descriptive statistics of biomechanical variables for Phase I and II participants.

Variable	*M*	SD
Time (seconds)		
1st sprint	6.67	0.73
2nd sprint	6.40	0.54
3rd sprint	6.36	0.53
4th sprint	6.32	0.54
4th−1st sprint	0.40	0.40
Gold standard male	4.99	0.09
Gold standard female	5.57	0.28
Step length (participant height)		
1st sprint	0.90	0.06
2nd sprint	0.95	0.05
3rd sprint	0.95	0.04
4th sprint	0.95	0.06
Gold standard male	1.11	0.14
Gold standard female	1.03	0.05
Step frequency (steps/minute)		
1st sprint	120.3	9.4
2nd sprint	118.5	8.4
3rd sprint	118.7	8.1
4th sprint	119.9	8.3
Gold standard male	135.0	5.1
Gold standard female	129.2	3.3
Front-to-back sway (degrees)		
1st sprint	20.3	5.3
2nd sprint	21.7	5.0
3rd sprint	21.9	5.1
4th sprint	21.6	4.6
Gold standard male	40.9	6.5
Gold standard female	48.5	6.9
Side-to-side sway (degrees)		
1st sprint	9.3	3.0
2nd sprint	10.3	2.8
3rd sprint	10.9	3.4
4th sprint	11.2	3.4
Gold standard male	25.0	1.9
Gold standard female	27.1	3.7

main participants were primarily of White ethnicity (58.3%). One quarter (25.0%) of the participants were West Asian, and the remaining participants were either Arab, Black, Indigenous, or South East Asian.

Phase II Procedures

Upon arrival to the testing area, all participants were asked to complete several questionnaires (see below). Following a warm-up (i.e., jogging and dynamic stretches), all participants were asked to watch a short video that explained the feedback they would receive after each sprint. Kinematic data were collected using inertial sensors worn on the participant's trunk, and right and left ankles (see Phase II Instruments section). After preparation, each participant performed four sets of 40 meter sprint tests on a straight path in an indoor varsity-standard running track. Immediately after each sprint, participants received a combination of KR (total time) and KP (stride length, stride frequency, side-to-side sway, and front-to-back sway) biomechanical feedback that compared their individual performance to the gold standard data. The time of the first sprint test was used to establish participant baseline sprint performance. The time of the fourth sprint test established the final performance record for each participant. Three minutes of rest were provided between trials to avoid fatigue. Participants were permitted more rest if requested, but none asked for extra time.

Phase II Instruments

Demographics

Participants reported their age, sex, ethnicity, medical history, as well as their sports participation and training history within the past 12 months (Daniels and Leaper, 2006; Mosewich, 2008).

Self-Compassion Scale

An athlete-specific version of the Self-Compassion Scale (SCS-AV: Killham et al., 2018) was used to measure participants' self-compassion in sport. The original 26-item Self-Compassion Scale (SCS) developed by Neff (Neff, 2003b) is characterized by six subscales: Self-Kindness (five items; e.g., "I'm kind to myself when I'm experiencing suffering"), Self-Judgment (five items; e.g., "When times are really difficult, I tend to be tough on myself"), Common Humanity (four items; e.g., "When things are going badly for me, I see the difficulties as part of life that everyone goes through"), Isolation (four items; e.g., "When I fail at something that's important to me I tend to feel alone in my failure"), Mindfulness (four items; e.g., "When something painful happens I try to take a balanced view of the situation"), and Over-Identification (four items; e.g., "When I'm feeling down I tend to obsess and fixate on everything that's wrong"). The SCS-AV has been tailored for athletes by incorporating language specific to a sport context (e.g., substituting "athletes" for "people"), rather than measuring self-compassion at a domain-general level. Items from the self-judgment, isolation, and over-identification subscales are all phrased negatively, and were reverse scored before computing the total scale mean. Participants responded to items on a 5-point scale (1 = almost never, 5 = almost always), with higher composite scores reflecting higher levels of sport-specific self-compassion. Validity and reliability evidence supporting the use of the SCS-AV has been reported in the sport literature ($\alpha = 0.85$–0.88: Killham et al., 2018).

Self-Esteem Scale

The Rosenberg Self-Esteem Scale (RSES: Neff, 2003) was used to assess athlete self-esteem ($\alpha = 0.82$–0.87: Rosenberg, 1965). The RSES is a unidimensional questionnaire made up of 10 items (e.g., "I feel that I am a person of worth, at least on an equal plane with others"). Some items are reverse coded. Respondents were asked to rate items on a four-point scale (0 = strongly disagree, 3 = strongly agree), with higher total scores indicating higher levels of sport self-esteem.

Concern Over Mistakes Scale

The Concern Over Mistakes subscale of the Sports Multidimensional Perfectionism Scale-2 (Sport-MPS-2) was used to assess athletes' sport-specific concern over mistakes (α = 0.79: Li et al., 2018). The subscale consists of eight items (e.g., "The fewer mistakes I make in competition, the more people will like me"). Participants responded to items on a 5-point scale (1 = strongly disagree, 5 = strongly agree), with higher total scores reflecting higher levels of sport-specific concern over mistakes.

Self-Criticism Scale

To measure participants' levels of self-criticism, the state self-criticism measure by Gilbert and Procter (2006) was used (α = 0.84: Gotwals and Dunn, 2009). This seven-item scale asks participants about the frequency, power, intrusiveness, length, and the difficulty of distraction from their self-critical thoughts (e.g., "How long did your self-critical thoughts last?"). In this study, athletes were asked to complete a self-criticism questionnaire both pre- and post-sprint trials. To better suit the study purpose, the wording of the questionnaires was altered. Pre-sprint trials, participants were asked to think about the past year and rate their critical thoughts. Post-sprint trials, participants were asked to consider the biomechanical feedback given about their sprinting performance and rate the severity of their self-critical thoughts related to receiving and implementing this feedback. Participants responded on a ten-point scale (1 = not at all, 10 = a lot of the time), with higher total scores indicating higher levels of self-criticism.

Performed Scenario Scale

To assess their overall experiences after completing the sprint task and receiving biomechanical feedback, athletes were asked to rate their emotions, thoughts, and reactions using scales adapted from Leary et al. (2007) (see **Supplementary Data** for further information). To capture athlete emotions, four unique dimensions were assessed: Sadness (i.e., sad, dejected, down, and depressed), Anxiety (i.e., nervous, tense, worried, and anxious), Anger (i.e., angry, irritated, mad, and hostile), and Embarrassment (i.e., embarrassed, humiliated, disgraced, and ashamed). Each dimension was made up of four items, and participants rated each item based on a six-point scale (1 = not at all, 6 = extremely). A total scale score was calculated, with higher scores indicating higher levels of total negative affect.

To assess athlete thoughts after receiving biomechanical feedback, four unique dimensions were assessed: Catastrophizing (one item; e.g., "This is awful!"), Personalizing (three items; e.g., "I am such a loser"), Equanimity (two items; e.g., "In the long run, this doesn't really matter"), and Humor (one item; e.g., "This is sort of funny"). Participants rated each item based on a five-point scale (1 = I did not think this thought at all, 5 = I kept thinking this thought). Total scores were calculated for each dimension, with higher scores indicating higher levels of that type of thought. A second set of questions was asked to further assess athlete emotions. There were six items in total: (a) I seem to have bigger problems than most people do; (b) I'm a loser; (c) This isn't any worse than what lots of other people go through; (d) Why do these things always happen to me?; (e) In comparison to other people, my life is really screwed up; and (f) Everyone has a bad day now and then. This second set of questions was rated on a six-point scale (1 = not at all, 6 = extremely), and the score of each individual item was used in further analysis.

To evaluate athlete reactions, participants were asked to complete two sets of questions. First, they were asked to rate the degree to which they displayed the following reactions after receiving biomechanical feedback on their sprinting performance: (a) Remained relatively calm and unflustered; (b) Overreacted; (c) Experienced strong emotions but did not get carried away with them; (d) Had no emotional reaction whatsoever; (e) Took the feedback in stride; (f) Set aside the feedback quickly in order to deal with my emotions; and (g) Replayed the feedback in my mind for a long time afterward. Participants responded on a six-point scale (1 = not at all, 6 = extremely). Some items were reverse scored (i.e., [b] and [g]). Higher total scores reflected higher levels of behavioral equanimity. The second set of questions asked about the degree to which each athlete reacted after receiving biomechanical feedback: (a) I tried to be kind to myself; (b) I tried to make myself feel better; (c) I was really hard on myself; (d) I kept the feedback in perspective; (e) I tried to do things to take my mind off of the feedback; (f) I expressed my emotions to let off steam; (g) I took steps to fix the problem or made plans to do so; (h) I sought out the company of others; and (i) I gave myself time to come to terms with it. Participants responded on a six-point scale (1 = not at all, 6 = extremely) and the score of each individual item was used in further analysis.

Timing and Inertial Sensors

The first author led all data collection. A set of electronic timing gates (Freelap Pro Coach BLE, 0.02 second accuracy, Freelap, USA) were used to time the sprinters' performance over the 40 meters distance. The same timing gates were used for both the gold standard athletes and the main participants. These timing gates used a chip that attached to the athlete's waistband. The time data transferred from the chip to a transmitter placed at the end of the track. As the athlete passed the transmitter, the time data was sent to a smart phone via Bluetooth connection and then manually recorded. The remaining biomechanical feedback data were collected by small lightweight wearable wireless inertial sensors (Opal, APDM Inc., Portland, OR; 43.7 mm × 39.7 mm × 13.7 mm; < 25 g). The inertial sensors contain three-dimensional accelerometers, gyroscopes, and magnetometers that provide kinematic information about sensor orientation, acceleration, and angular velocity. Data were collected at a sample rate of 128 Hz and transmitted wirelessly to a PC-based laptop. The participants wore three inertial sensors: one each on the trunk, right ankle (RA), and left ankle (LA). The sensors on the ankles were securely fixed by loop and hook fabric straps on the anterior surface of the shin near the ankle to prevent motion artifact noise. These straps did not limit the range of motion of the ankle joint. The trunk sensor was fixed over the sternum using custom light-weight straps.

The angular velocity of the RA and LA extracted from sensors was analyzed using a custom algorithm written in

MATLAB (R2019a, Mathworks, Natick, MA) to calculate the total number of steps for each trial. The sagittal angular velocity of the RA and LA sensors were first filtered using a fourth order low pass Butterworth filter with a cut off frequency of 5 Hz. Distinct angular velocity peaks associated with each step were then identified using a thresholding technique, which gave a count of the number of steps in the data. The step indices were used in conjunction with the trunk resultant acceleration to automatically identify the first step. The externally measured sprint time was then used to determine the number of steps that were part of the 40-meter sprint. Step time was calculated by dividing the step count by the sprint time. The step frequency was calculated as the inverse of step time and multiplied by 60 to express it in units of steps per minute (i.e., $\textit{Step frequency}\ [\frac{step}{minute}] = \frac{1}{\textit{Step time (seconds)}} \times 60$). The average step length was calculated by dividing the 40-meter distance by the number of steps. To normalize the step length, values were divided by each participant's height. Data from the trunk sensor was used to calculate the side-to-side and front-to-back sway of the trunk using another custom MATLAB routine. Raw trunk angular velocity data in each movement axis was partitioned using the step timing information obtained from the ankle sensors. The angular velocity data for each step was then numerically integrated to obtain the angular displacement (i.e., sway) for each step. The range of motion of both side-to-side sway and front-to-back sway were calculated.

Feedback was given on a laptop screen using a custom MATLAB routine. After each sprint trial, athletes received their step length and frequency data in the form of a two-row bar graph. The top bar represented sex-matched gold standard data and the lower bar was the participant's performance for that trial (**Figure 1**). Numerical data for these variables were also presented beside the bar graphs. In addition, explicit text was provided for the participant to read about how their data differed from the gold standard. This feedback was either positive, meaning that the participant had exceeded the gold standard metric, or negative, meaning that the participant's results were worse than the gold standard metric. For example, participants would be told that their sprint time was SLOWER or FASTER than the gold standard. For feedback about side-to-side and front-to-back sway of the trunk, visual feedback in the form of two arcs showing the range of angular movement (with the gold standard data above the participant data) was given. To examine if there was any significant change in sprint performance after receiving and implementing feedback, all the data for each trial was saved for further analysis.

Phase II Data Analysis

Statistical analysis was computed using SPSS v22 (IBM Corp., Armonk, NY, USA) with the alpha level set at 0.05. Data were screened for missing responses and normality. No missing data were identified across any of the scales and subscales; thus, data from 48 participants were used for hypothesis testing[1]. Self-compassion, self-esteem, concern over mistakes, and pre- and post-trials self-criticism scales were normally distributed, whereas most of the items in the post-trials emotions, thoughts, and reactions subscales violated the normality assumption and were positively skewed. Based on the protocol used by other self-compassion studies in sport (Mosewich et al., 2011; Reis et al., 2015), which described that their substantive conclusions did not change after transforming similar data, we used the non-transformed scales values in all analyses.

A one-tailed Pearson correlation analysis was performed to test if the relationships among self-compassion, self-esteem, concern over mistakes, and pre-trials self-criticism were in the same directions as predicted in Hypothesis 1. To test Hypothesis 2, a one-tailed Pearson correlation analysis was used to determine the correlations between self-compassion, self-esteem, concern over mistakes, and pre-trials self-criticism with post-trials self-criticism, emotions, thoughts, and reactions. Hypothesis 3 was tested using a within-between 4 (sprint trial) × 2 (sex) repeated measures ANOVA to examine if any significant changes occurred in participant biomechanical variables after receiving and implementing biomechanical feedback. Afterwards, a moderated regression analysis was conducted using self-compassion as a moderator between the first and fourth sprint times. The fourth sprint time was entered as a dependent variable. The first sprint time was entered into the model in step 1. Self-compassion scores were introduced in step 2. Finally, to measure whether self-compassion was a moderator for change in sprint performance, the interaction between first sprint time and self-compassion (i.e., first sprint time × self-compassion) was introduced in step 3.

RESULTS

Descriptive statistics and the internal reliabilities for all psychological scales and subscales are presented in **Table 2**.

Hypotheses Testing

Hypothesis 1: Relationships Between Psychological Variables at Baseline

Hypothesis 1 was predicated on the relationships between psychological scales prior to the sprint tests and was supported by our results (see **Table 3**). We found negative correlations between self-compassion, and concern over mistakes and pre-trials self-criticism. Furthermore, we found a positive correlation between self-compassion and self-esteem.

Hypothesis 2: Relationships Between Psychological Variables Before and After Sprint Testing

The correlations among all variables relevant to Hypothesis 2 are shown in **Table 4**. Supporting Hypothesis 2a, there was a

[1]It was our intention to combine the psychological data from male and female subjects into a single sample, as assessing sex-based differences was not the primary purpose of this study. Preliminary independent t-tests were conducted to determine if sex-based differences existed across the four major psychological variables measured (female = 1, male = 2). No significant differences were found between males and females in each of the four psychological variables, including self-compassion [$t_{(1,46)} = 1.08$, $p = 0.29$], self-esteem [$t_{(1,46)} = -1.42$, $p = 0.16$], concern over mistakes [$t_{(1,46)} = 0.34$, $p = 0.74$], pre-trial self-criticism [$t_{(1,46)} = 1.22$, $p = 0.23$] and post-trial self-criticism [$t_{(1,46)} = 1.22$, $p = 0.23$]. However, since the present study included a relatively small sample size, results related to the testing of sex differences should be interpreted with caution.

TABLE 2 | Descriptive statistics and internal reliabilities of psychological scales and subscales for Phase II participants.

Variable	*M*	SD	α
Self-compassion	3.47	0.55	0.86
Self-esteem	3.20	0.40	0.79
Concern over mistakes	2.58	0.87	0.87
Pre-trials self-criticism	4.35	1.42	0.76
Post-trials self-criticism	3.62	1.25	0.71
Emotions (total negative affect)	20.73	7.78	0.86
Thoughts (set one)			
Catastrophizing	2.58	1.23	0.55
Personalizing	2.79	1.15	–
Equanimity	1.56	1.03	0.30
Humorous	1.73	1.16	–
Thoughts (set two)			
I seem to have bigger problems than most people do	1.27	0.54	–
I'm a loser	1.04	0.20	–
This isn't any worse than what lots of other people go through	1.87	1.23	–
Why do these things always happen to me?	1.02	0.14	–
In comparison to other people, my life is really screwed up	1.15	0.41	–
Everyone has a bad day now and then	1.73	1.16	–
Reactions (set one)			
Behavioral equanimity	28.73	3.51	0.86
Reactions (set two)			
I tried to be kind to myself	3.75	1.30	–
I tried to make myself feel better	3.45	1.54	–
I was really hard on myself	2.23	1.30	–
I kept the feedback in perspective	4.52	1.13	–
I tried to do things to take my mind off of the feedback	1.50	0.92	–
I expressed my emotions to let off steam	1.46	0.99	–
I took steps to fix the problem or made plans to do so	4.06	1.69	–
I sought out the company of others	1.73	1.16	–
I gave myself time to come to terms with it	2.44	1.44	–

α refers to Cronbach's alpha measure of internal consistency. For single items (i.e., personalizing and humorous thoughts, thoughts [set two] and reactions [set two]), internal consistency values could not be calculated. The ranges of the measures included in the above table are as follows: self-compassion (range = 1–5); self-esteem and concern over mistakes (range = 1–4); pre- and post-trials self-criticism (range = 1–10); total negative affect (range = 16–96); catastrophizing and personalizing thoughts (range = 2–10); equanimity and humorous thoughts (range = 1–5); set two thoughts (range = 1–6); behavioural equanimity (range = 7–42); set two reactions (range = 1–6).

negative correlation between post-trials self-criticism and both self-compassion and self-esteem. Hypothesis 2b was partially supported, as there was a negative correlation between both self-compassion and self-esteem with maladaptive emotions (i.e., total negative affect) and catastrophizing thoughts, and a positive

TABLE 3 | One-tailed Pearson correlations before receiving feedback for Phase II participants.

Variable	Self-esteem	Concern over mistakes	Pre-trials self-criticism
Self-compassion	0.57**	−0.69**	−0.52**
Self-esteem	–	−0.46*	−0.50**
Concern over mistakes		–	0.50**

$^{*}p < 0.05$; $^{**}p < 0.01$.

correlation with some positive reactions (i.e., "I tried to be kind to myself," and "I tried to make myself feel better"). However, Hypothesis 2b was not fully supported as no relationship was found between both self-compassion and self-esteem and the remaining thoughts and reactions. Hypothesis 2c was fully supported as there was a positive correlation between post-trials self-criticism and both concern over mistakes and pre-trials self-criticism. Lastly, partial support was found for Hypothesis 2d as both concern over mistakes and post-trials self-criticism were positively correlated to maladaptive emotions (i.e., total negative affect), but were unrelated to any other thoughts or reactions.

Hypothesis 3: Changes Between Biomechanical Variables Before and After Sprint Testing

We predicted that athletes with higher levels of self-compassion would have significantly better sprint performances after receiving and implementing biomechanical feedback. Descriptive statistics for the biomechanical variables are represented in **Table 5**. The results of a within-between repeated measures ANOVA revealed that there was a change in sprint time [$F_{(1.67,76.78)} = 21.61$, $p < 0.001$], step length [$F_{(2.43,111.79)} = 22.72$, $p < 0.001$], side-to-side sway [$F_{(2.40,110.56)} = 18.95$, $p < 0.001$], and front-to-back sway [$F_{(2.73,125.64)} = 8.10$, $p < 0.001$] over sprint trials. The pairwise comparison of the means with Bonferroni adjustment ($\alpha = 0.05/6$) showed that the changes in all four variables occurred between the first sprint set and the second, third, and fourth sprint sets. There were no interactions between these biomechanical variables and sex, but there were sex effects on sprint time [$F_{(1,46)} = 22.94$, $p < 0.001$], and front-to-back sway [$F_{(1,46)} = 13.362$, $p < 0.05$]. To account for this, sprint time and sway were expressed as percentages of the respective sex-based gold standards and the analysis was re-run. All other results remained the same, but the sex effects were no longer significant (see **Figure 2**).

As the changes in sprint time, step length, side-to-side sway, and front-to-back sway were similar after receiving biomechanical feedback, we chose to use sprint time as the main indicator of sprint performance to test Hypothesis 3 as it is likely the most tangible and familiar sprint-related concept to the reader and layman alike. A one-tailed Pearson correlation was used to assess the relationships between sprint time and psychological factors measured pre- and post-sprint trials. There were no significant relationships between sprint time and self-compassion, self-esteem, concern over mistakes, and post-trials self-criticism. The correlations between sprint time

TABLE 4 | One-tailed Pearson correlations of psychological scales and subscales for Phase II participants.

Variable	Self-compassion	Self-esteem	Concern over mistakes	Pre-trials self-criticism
Post-trials self-criticism	−0.38**	−0.36**	0.25*	0.59**
Total negative affect	−0.38**	−0.55**	0.32*	0.31*
Thoughts (set one)				
Catastrophizing	−0.37**	−0.35**	0.27*	0.02
Personalizing	−0.34**	−0.18	0.13	0.17
Equanimity	−0.24*	−0.11	0.08	0.06
Humorous	−0.03	0.02	−0.02	−0.11
Thoughts (set two)				
I seem to have bigger problems than most people do	−0.11	−0.24	0.18	0.20
I'm a loser	−0.19	−0.34**	0.04	0.06
This isn't any worse than what lots of other people go through	0.03	0.07	−0.10	−0.18
Why do these things always happen to me?	−0.04	−0.00	−0.10	0.10
In comparison to other people, my life is really screwed up	−0.06	−0.31*	0.10	−0.00
Everyone has a bad day now and then	−0.11	−0.21	0.10	−0.10
Reactions (set one)				
Behavioral equanimity	0.22	0.32*	−0.08	−0.15
Reactions (set two)				
I tried to be kind to myself	0.38**	0.31*	−0.33*	−0.22
I tried to make myself feel better	0.33*	0.32*	−0.23	−0.18
I was really hard on myself	−0.18	−0.24*	0.21	0.23
I kept the feedback in perspective	0.27*	−0.06	−0.15	−0.01
I tried to do things to take my mind off of the feedback	0.10	−0.04	−0.09	−0.02
I expressed my emotions to let off steam	−0.02	0.21	−0.00	−0.07
I took steps to fix the problem or made plans to do so	0.23	0.32*	0.10	0.05
I sought out the company of others	0.05	−0.02	0.03	0.10
I gave myself time to come to terms with it	0.23	0.19	−0.09	0.02

*$p < 0.05$; **$p < 0.01$.*

TABLE 5 | Pairwise means comparison of biomechanical variables between sprint trials of Phase II participants.

Variable	Mean difference	SE	P^a
Sprint time			
1st−2nd	0.26*	0.05	0.000
1st−3rd	0.31*	0.06	0.000
1st−4th	0.34*	0.07	0.000
Step length			
1st−2nd	−0.05*	0.01	0.000
1st−3rd	−0.06*	0.01	0.000
1st−4th	−0.50*	0.01	0.000
Step frequency			
1st−2nd	1.90	1.03	0.428
1st−3rd	1.49	0.88	0.575
1st−4th	0.22	1.12	1.000
Front-to-back sway			
1st−2nd	−1.37*	0.40	0.008
1st−3rd	−1.52*	0.39	0.002
1st−4th	−1.29*	0.38	0.008
Side-to-side sway			
1st−2nd	−1.02*	0.23	0.000
1st−3rd	−1.66*	0.31	0.000
1st−4th	−1.89*	0.32	0.000

The mean difference column represents the differences between the means of each biomechanical variable at each given sprint trial. SE, standard error; P^a, Significance level with Bonferroni adjustment.
$*p < 0.05/6$.

and post-trials self-criticism, emotions, thoughts, and reactions are presented in **Table 6**.

Further analyses were completed to evaluate the relationship between self-compassion and the sprint performances of athletes after receiving and implementing biomechanical feedback. A moderated regression analysis was run with the interaction between first sprint time and self-compassion entered as a moderator between the first and fourth sprint times. This interaction did not predict any unique variance in sprint performance, thus self-compassion was not a moderator for change in sprint performance ($R^2 = 0.642$, $\Delta R^2 = 0.10$, $p > 0.05$). In sum, the results of these analyses refuted Hypothesis 3. While sprint performance was improved after receiving biomechanical feedback, there was no relationship between the improvement of sprint performance and athlete self-compassion.

DISCUSSION

The purpose of this research was to explore whether self-compassion, self-esteem, concern over mistakes, and self-criticism could predict athletes' responses to biomechanical feedback after a series of sprint performances, and whether self-compassion could buffer against any maladaptive emotions, thoughts, and reactions experienced after receiving and implementing this feedback. We found that athletes with higher levels of self-compassion demonstrated higher levels of self-esteem, less concern over mistakes, and lower levels

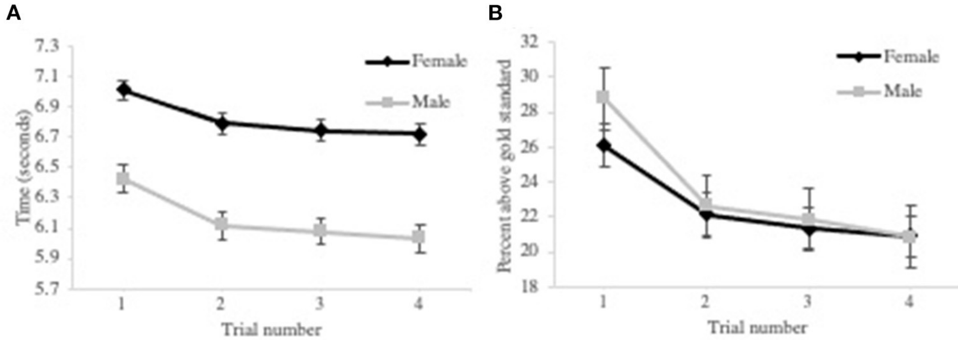

FIGURE 2 | See effect on Phase II participants sprint time. Graph **(A)** represents the original sex-specific mean for each sprint time trial. Graph **(B)** represents the sex-specific sprint time mean expressed as the difference between the subject's sprint time and the gold standard as a percent of the gold standard obtained from the participants for each sprint time trial.

TABLE 6 | One-tailed Pearson correlations after receiving feedback after each sprint trial for Phase II participants.

Variable	1st sprint trial	2nd sprint trial	3rd sprint trial	4th sprint trial	4th−1st sprint trial
Post-trials self-criticism	0.08	0.21	0.16	0.19	0.10
Total negative affect	0.33*	0.28**	0.33*	0.31*	−0.17
Thoughts (set one)					
Catastrophizing	−0.01	0.02	0.01	0.10	0.13
Personalizing	0.23	0.14	0.07	0.07	−0.30*
Equanimity	−0.03	−0.07	0.03	0.02	0.082
Humorous	−0.37**	0.33*	0.25*	0.21	−0.35**
Thoughts (set two)					
I seem to have bigger problems than most people do	0.19	0.28*	0.34**	0.37**	0.14
I'm a loser	−0.04	−0.01	−0.01	0.07	0.15
This isn't any worse than what lots of other people go through	0.11	0.09	0.10	0.07	−0.10
Why do these things always happen to me?	0.00	0.07	0.07	0.09	0.11
In comparison to other people, my life is really screwed up	0.18	0.16	0.20	0.16	−0.11
Everyone has a bad day now and then	0.17	0.10	0.13	0.21	−0.03
Reactions (set one)					
Behavioral equanimity	−0.07	0.01	−0.06	−0.04	0.07
Reactions (set two)					
I tried to be kind to myself	0.10	0.05	0.02	0.02	−0.14
I tried to make myself feel better	0.26*	0.19	0.13	0.12	−0.28*
I was really hard on myself	0.20	0.21	0.18	0.21	−0.08
I kept the feedback in perspective	0.03	0.02	0.01	−0.09	−0.16
I tried to do things to take my mind off of the feedback	0.34**	0.20	0.23	0.15	−0.38**
I expressed my emotions to let off steam	0.18	0.08	0.08	0.05	−0.23
I took steps to fix the problem or made plans to do so	0.14	0.13	0.03	−0.06	−0.30*
I sought out the company of others	0.27*	0.07	0.07	0.02	−0.42**
I gave myself time to come to terms with it	0.36**	0.25*	0.22	0.11	−0.45**

*$p < 0.05$; **$p < 0.01$.

of pre-trials self-criticism. We also found that athletes with higher levels of self-compassion prior to sprint performance experienced less self-critical thoughts following biomechanical feedback and subsequent sprint trials. Though the results of this study provide some support for the effectiveness of self-compassion in promoting adaptive emotions, thoughts, and reactions in response to sprint performance biomechanical feedback, we found little evidence suggesting that high levels of self-compassion led to more effective implementation of the feedback and improved sprint performance. More specifically, self-compassion was not a moderator of change in sprint performance across trials, and the relationship between self-compassion and sprint performance was non-significant.

Our study supported the results of previous studies, in that we found negative correlations among self-compassion and concern over mistakes, pre-trials self-criticism, and post-trials self-criticism (Mosewich et al., 2013). We also found negative correlations between self-esteem and concern over mistakes, pre-trials self-criticism, and post-trials self-criticism (Reis et al., 2019). We also found that both pre-trials self-criticism and concern over mistakes were positively correlated with post-trials self-criticism and emotions (i.e., total negative affect), though the majority of the relationships between pre-trials self-criticism and concern over mistakes, and post-trials thoughts and reactions were non-significant. A possible reason for these non-significant relationships could have been our participants' high mean levels of self-compassion—descriptive analyses revealed that athletes who participated in this study had higher levels of self-compassion ($M = 3.47$, $SD = 0.55$) in comparison to Reis et al.'s ($M = 3.10$, $SD = 0.58$; 2015) and Leary et al.'s (Study 1: $M = 3.15$, $SD = 0.63$; Study 2: $M = 3.03$, $SD = 0.58$; Study 5: $M = 3.08$, $SD = 0.58$; 2007) participants. Self-compassion can be used as a coping strategy to reduce an individual's self-criticism and concern over mistakes (Gilbert and Procter, 2006; Mosewich et al., 2013). Additionally, self-compassion can reduce maladaptive emotions, thoughts, and reactions by fostering more positive perceptions of the self (Neff, 2003b). Perhaps the relatively high self-compassion scores of the athletes included in this study made it less likely that they would experience adverse reactions to the biomechanical feedback, and thus experienced less self-criticism and concern over mistakes. In other words, it seems possible that the athletes simply did not experience the biomechanical feedback as an emotionally difficult setback that fostered self-criticism and concern over mistakes, as they were already viewing the feedback from a more self-compassionate lens.

Previous studies have emphasized the positive role of self-compassion in coping with emotionally difficult sport-related experiences (e.g., being responsible for a team failure: Mosewich et al., 2013; Ferguson et al., 2014; Reis et al., 2015). Some of our results are in line with these findings. Specifically, the correlations between self-compassion and emotions (i.e., total negative affect), and some thoughts (i.e., catastrophizing, personalizing, and equanimity) and reactions (i.e., "I tried to be kind to myself," "I tried to make myself feel better," and "I kept the feedback in perspective") were positive, thus emphasizing the potential positive effects of self-compassion on athletes' responses to challenging sport situations. Conversely, the relationships between self-compassion and the remaining measured thoughts and reactions (e.g., humorous thoughts, behavioral equanimity) were not significant and did not align with extant research. Perhaps athletes are so habituated to receiving performance-related feedback—though not necessarily biomechanical feedback—over their many years of training that they might already have a set of coping skills in place to cope with feedback. This explanation might also support why self-esteem was sufficient for predicting athlete responses to the given feedback. Receiving and implementing biomechanical feedback also may not be a situation in which self-compassion is essential, as it might be one of several effective coping skills that athletes have available to them.

We found that athletes receiving biomechanical feedback for the first time (i.e., after their first sprint set) significantly improved their following sprint performance (i.e., second sprint set). A wide range of studies provide compelling evidence that support the positive effect that biomechanical feedback has on performance enhancement (Broker et al., 1989; Sanderson and Cavanagh, 1990; Mendoza and Schöllhorn, 1993); however, no significant differences were observed in athlete performance between their second, third, and fourth sprint performances. One possible explanation could be the possible misinterpretation of biomechanical feedback by study participants. Preatoni et al. (2013) highlighted the importance of translating biomechanical feedback into easily understandable information for athletes. No prescriptive feedback (i.e., error correction) was provided by our study experimenter which may have limited impact of the biomechanical feedback and the ability of athletes to implement the feedback in subsequent performances.

Another possible explanation for the improvement in athlete performance between only the first and second sprint sets could be attributed to a practice effect. Athletes may have improved their sprint sets due to familiarization with test protocols and equipment (i.e., inertial sensors and timing gate chip) rather than feedback effects. To ascertain if familiarization was indeed responsible for this performance improvement, a secondary experiment was performed on 10 athletes that were not part of the current study. Results revealed that there was no significant effect on sprint performance between first and second trials[2]. The data from this secondary study suggest that the performance improvement seen between the first and second trials in the main study were unlikely to have been caused by practice effects.

In the current study, a combination of KR and KP biomechanical feedback was provided to athletes at the end of each trial. The athletes received KR feedback on sprint time and KP feedback on their step frequency, step length, and trunk sway. As discussed above, we found a significant improvement between the first and second trial sprint times (i.e., KR feedback), but no significant changes were observed in subsequent trials. One possible reason for this lack of continued change could be due to the frequency of providing KR feedback to our participants. Previous studies have shown that less frequent KR feedback (i.e., having some no-feedback trials) can improve the motor skill acquisition more efficiently than when more frequent KR feedback is given (Salmoni et al., 1984). It has been posited that while frequent KR feedback might guide a learner to target behavior, it may later prevent individuals from using intrinsic feedback processing and error detection and might cause learner dependency (Salmoni et al., 1984). Additionally, we found that

[2]Ten athletes (9 male: $M_{age} = 21.8$, $SD = 3.1$) performed four sets of 40-meter sprints with identical biomechanical data collection protocol as Phase II but with no psychological questionnaires. Participants received similar biomechanical feedback after sprinting as the original study, but only began to receive feedback after the second sprint trial (rather than the first). The result of a repeated measure ANOVA analysis revealed no significant sprint performance improvements between the first and second trials.

the effects of KP feedback were also not consistent across each of the measured variables. In the extant literature, evidence suggests that frequently delivered KP feedback is a superior guide during basic skill acquisition, particularly for novices (Wulf et al., 1994; Wulf and Shea, 2002). Frequent KP feedback has also been shown to significantly improve complex movements (i.e., sports-specific skills requiring whole-body movements with many degrees of freedom: Wulf et al., 1994). More specifically, when Wulf et al. (1998) provided KP feedback about force onset to participants participating in ski simulator training, those who received KP feedback at 100% frequency demonstrated superior performance improvement over those who only received KP feedback at 50% frequency. Perhaps providing our main study participants with higher frequency KP feedback may have led to improvements in step frequency, step length, and trunk sway performances over sprint trials. It is also possible that the main study participants may not have known how to implement new techniques to efficiently improve all of the examined metrics of sprint performance, as receiving this specific biomechanical feedback was likely new to them. For example, participants who had less side-to-side trunk sway than the gold standard may have known that they *should* increase their trunk sway to improve their sprint time, but they may not know *how* to do so. Providing additional prescriptive feedback may have facilitated participants' interpretations of biomechanical feedback received about their sprint performance and may have led to continued improvements in these kinematic variables over sprint trials.

Limitations

A limitation of our study was the ability to fully control the testing environment. Sports performance can vary based on the presence of others (Geisler and Leith, 1997). All the data from the main participants were collected on a public indoor running track; however, to help control for potential distraction of other people, testing hours were chosen to minimize the effects of external factors (e.g., number of non-participants present at the location, audience feedback, noise, etc.), and only study participants were allowed to enter the test site. Having said this, despite our best attempts, the testing environment did vary to some degree across participants.

Another limitation of this research may have been the limited variability of our participants' baseline levels of self-compassion. One of the primary objectives of our study was to explore whether self-compassionate athletes had better sprint performance than their less self-compassionate peers after receiving feedback. Purposefully recruiting athletes with varying levels of self-compassion may help in examining the responses to biomechanical feedback in athletes as this would have likely resulted in a greater variance in the other psychological variables we measured and, consequently, the effectiveness of using self-compassion as a coping strategy.

A final limitation involved our inability to stratify our sample based on sex due to the size of our sample pool, which impeded our ability to draw any conclusions about the presence of sex-based differences in the measured psychological variables. Some evidence exists demonstrating that males (and men) and females (and women) score differently on measures of self-compassion (Lizmore et al., 2017), self-esteem (Li et al., 2018), concern over mistakes (Cremades et al., 2013), and self-criticism (Luyten et al., 2007), but other researchers (Gotwals et al., 2003; Anshel et al., 2009; Huysmans and Clement, 2017; Dunn et al., 2021) have found the opposite to be true (i.e., no significant differences between sexes or gender, see footnote 1). Future researchers should consider stratifying their sample by sex, in addition to controlling for possible differences in gender in their analyses.

Future Directions

There are several considerations that can be implemented to develop more effective biomechanical feedback for athletes. The participants in our study were not track and field athletes and it is likely that some of the biomechanical content of the feedback may have been new to them. As such, a combination of coach/expert feedback and tutorial videos might have provided better guidance for athletes (Preatoni et al., 2013) and subsequently improved their performance. Future studies may find different results if both descriptive and prescriptive biomechanical feedback are delivered to study participants. Second, the emotional reactions of individuals during sport performance can change across time points, especially in stressful situations (Crocker et al., 1998; Cerin et al., 2000). Measuring discrete emotions that might be relevant to the experience of receiving biomechanical feedback (including both negative and positive emotions; e.g., pride, joy, and happiness) might garner more information about the emotional experiences of athletes receiving biomechanical feedback about their performance. Next, although the validity and reliability of composite self-compassion scores has been well-documented (Neff, 2003b), analyzing the self-compassion subscales offers an alternative approach to explore the associations among self-compassion and adaptive emotions, thoughts, and reactions in response to biomechanical feedback. Previous studies show the utility of specific subscales—such as Common Humanity and Mindfulness—in decreasing self-criticism (Jopling, 2000) and openness to feedback (Neff, 2003a). It seems reasonable that some self-compassion subscales would be more strongly related to athletes' responses to feedback than a total score. Lastly, further research studying the effects of self-compassion inductions on sport performance and athletes' responses to biomechanical feedback is needed. Practicing self-compassion could increase athletes' awareness of their emotions, thoughts, and reactions (Leary et al., 2007; Reis et al., 2015), and—as a result of being more self-compassionate—might experience more adaptive responses to biomechanical feedback.

CONCLUSION

Our results suggest that self-compassion can play a nuanced role in athlete coping and sport performance. Self-compassion was related to adaptive psychological characteristics prior to sprint trials (i.e., positively correlated with self-esteem, negatively related to self-criticism, and concern over mistakes) and seemed to attenuate athletes' negative responses to biomechanical feedback after sprint performance (e.g., negative affect, negative thoughts, and reactions). However, higher levels of self-compassion did not seem to help athletes perform better after receiving biomechanical feedback. We also found that the specific type and frequency of the objective biomechanical feedback used in this study did not appear to improve sprinting performance in this population. While the first presentation of biomechanical feedback significantly improved the sprint time, step length, side-to-side sway, and front-to-back sway in participants, these enhancements were not continually improved across subsequent attempts. There is a possibility that providing more guidance (i.e., prescriptive coach feedback) might have increased the impact of biomechanical feedback on athlete performance. To further advance the literature, sport researchers might consider different approaches to providing biomechanical feedback to their participants, measuring additional emotions that might be relevant to the experience of receiving feedback, approaching the measurement of self-compassion from a subscale perspective, and implementing self-compassion inductions prior to performance trials. We believe that further efforts made to identify the constellation of psychological skills that can lead to the efficient execution of feedback in competitive environments will ultimately enhance sport performance and increase adaptive psychological outcomes for athletes.

DATA AVAILABILITY STATEMENT

The datasets presented in this article are not readily available due to lack of participant consent. Requests to access group-only data should be directed to joel.lanovaz@usask.ca.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the University of Saskatchewan Research Ethics Board. Written informed consent to participate in this study was provided by the participants.

AUTHOR CONTRIBUTIONS

YAA was supervised by JLL and KCK. YAA, KCK, ARO, LJF, and JLL contributed to conception and design of the study. YAA collected all data and performed the statistical analysis. DLC had significant writing responsibilities for the manuscript version of the research. KCK was responsible for the acquisition of the financial support leading to this publication. All authors read, revised, and approved the submitted version of this manuscript.

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Greater Breast Support Alters Trunk and Knee Joint Biomechanics Commonly Associated With Anterior Cruciate Ligament Injury

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Objective: The female breast is a passive tissue with little intrinsic support. Therefore, women rely on external breast support (sports bras) to control breast motion during athletic tasks. Research has demonstrated that lower levels of breast support are associated with altered trunk and pelvis movement patterns during running, a common athletic task. However, no previous study has identified the effect of sports bra support on movement patterns during other athletic tasks including landing. Therefore, the purpose of this study was to examine the effects of breast support on trunk and knee joint biomechanics in female collegiate athletes during a double-leg landing task.

Methods: Fourteen female collegiate athletes completed five double-leg landing trials in each of three different sports bra conditions: no support, low support, and high support. A 10-camera motion capture system (250 Hz, Qualisys, Goteburg, Sweden) and two force platforms (1,250 Hz, AMTI, Watertown, MA, USA) were used to collect three-dimensional kinematics and ground reaction forces simultaneously. Visual 3D was used to calculate trunk segment and knee joint angles and moments. Custom software (MATLAB 2021a) was used to determine discrete values of dependent variables including vertical breast displacement, knee joint and trunk segment angles at initial contact and 100 ms post-initial contact, and peak knee joint moments. A repeated measures analysis of covariance with *post-hoc* paired samples *t*-tests were used to evaluate the effect of breast support on landing biomechanics.

Results: Increasing levels of breast support were associated with reductions in peak knee flexion (Right: $p = 0.008$; Left: $p = 0.029$) and peak knee valgus angles (Right: $p = 0.011$; Left: $p = 0.003$) as well as reductions in peak knee valgus moments (Right: $p = 0.033$; Left: $p = 0.013$). There were no changes in peak knee extension moments (Right: $p = 0.216$; Left: $p = 0.261$). Increasing levels of breast support were associated with greater trunk flexion angles at initial contact ($p = 0.024$) and greater peak trunk flexion angles ($p = 0.002$).

Conclusions: Lower levels of breast support are associated with knee joint and trunk biomechanical profiles suggested to increase ACL injury risk.

Keywords: ACL, breast, knee, sports bra, biomechanics, landing, injury

INTRODUCTION

Landing tasks in multidirectional sports are associated with a variety of lower extremity injuries for both males and females. However, female athletes have a greater prevalence of traumatic knee injury than males (Arendt and Dick, 1995; NFHS, 2016). Specifically, female athletes are up to eight times more likely to experience an anterior cruciate ligament (ACL) injury than their male counterparts in the same sport (Arendt and Dick, 1995; NFHS, 2016).

The exaggerated rate of ACL injury in female athletes has been attributed in part to distinct differences in lower extremity biomechanical patterns in females compared to males. Pappas et al. (2007) revealed that females exhibit greater peak knee valgus angles and peak vertical ground reaction forces (GRFs) than male athletes (Pappas et al., 2007) when landing from a height of 40 cm. As the mechanical demand increased from 40 to 60 cm, female athletes also exhibited greater peak ankle dorsiflexion, and peak foot pronation than male athletes (Kernozek et al., 2005). Further, during unanticipated side-step cutting, females exhibited greater knee abduction angles at initial contact (IC) and greater peak ankle eversion angles during stance phase than males. Greater ankle eversion angle has been suggested to contribute to greater tibial internal rotation while greater knee valgus prior to cutting may place greater load on the structures of the knee including the ACL, therefore, increasing the risk of injury (Ford et al., 2005). These sex-related differences in lower extremity biomechanics during both landing and cutting may explain the greater rate and incidence of ACL injuries in female compared to male athletes.

An understudied factor known to alter lower extremity biomechanics and ACL injury risk is trunk biomechanics. During a sixty-centimeter vertical double-leg drop-landing task, individuals that landed with greater trunk flexion angles also exhibited greater hip and knee flexion angles (Blackburn and Padua, 2008, 2009). In addition, individuals landing with greater trunk, hip, and knee flexion angles also experienced a decrease in quadricep activity and increase in hamstring muscle force (Blackburn and Padua, 2009; Kulas et al., 2010). This increase in hamstring muscle force is suggested to counteract knee anterior shear forces when landing with greater trunk flexion, as opposed to landing with greater trunk extension (Kulas et al., 2010). Even when the mechanical demand of the task is decreased to a single-leg squat task, individuals squatting with a moderate amount of trunk lean flexion, as opposed to minimal amounts of trunk flexion, still experienced higher hamstring muscle forces and lower peak and mean ACL forces and strains (Kulas et al., 2012). Further, a study investigating the effect of fatigue on landing biomechanics demonstrated that sagittal and frontal plane trunk and lower extremity alignment are altered by fatigue potentially increasing risk of ACL injury during a landing task (Liederbach et al., 2014). While trunk biomechanics may play a role in increased ACL stress and increased ACL injury risk, these studies do not compare trunk biomechanical differences between females and males.

A sex-specific trait that has been shown to alter trunk biomechanics during sport-related movements is the female breast. Breast development occurs with physical maturation (Biro et al., 2013), has been associated with altered lower extremity biomechanics (Hewett et al., 2004; Sigward et al., 2012) and mirrors the sex-based divergence in ACL injury rates (Sanders et al., 2016). Female breasts are a passive tissue that are only supported by connective tissue (Gaskin et al., 2020). Because of this, breasts have limited intrinsic support and often require the use of extrinsic support, typically in the form of sports bras during highly dynamic activities. Without the use of sports bras and sufficient breast support, females can experience increased levels of embarrassment, decreased willingness to exercise, and increased levels of breast discomfort or pain (Risius et al., 2017). By wearing sports bras and sufficient support, females can control for vertical, anteroposterior, and mediolateral breast displacement (Scurr et al., 2009, 2011). Additionally, breast support has been found to create significant changes in running biomechanics including peak pelvis rotation, pelvis range of motion, vertical trunk oscillation, peak trunk rotation, and trunk range of motion as well as peak torso range of motion across all planes of motion (Milligan et al., 2015; Risius et al., 2017). However, a majority of breast support in sports movement research has limited focus to upper extremity and trunk biomechanics specifically during running. Further, this research has primarily investigated large breasted females with a breast size of a D-cup.

While previous literature has determined that both lower extremity and trunk biomechanics can increase the risk of ACL injuries, no previous research has investigated the effect of breast support on lower extremity and trunk biomechanics associated with ACL injury during landing tasks. Therefore, the purpose of this study is to determine the effect of sports bra support on trunk and knee joint biomechanics in female collegiate athletes during a double-leg landing task. It was hypothesized that increasing levels of breast support would be associated with knee and trunk biomechanics less indicative of ACL injury risk.

MATERIALS AND METHODS

Participants

An *a prior* power analysis (G*Power 3.1.5) was conducted based on findings from preliminary data. Using an effect size of 0.40, an alpha level of 0.05 and power (1–β) of 0.80, it was determined that a total sample size of 12 will provide sufficient statistical power for the study (Portney and Watkins, 2009). A total of 14 female athletes were recruited for this study. However, *two* participants did not complete all experimental conditions, and were consequently not included in the data analysis. Inclusion criteria included (1) 18–25 years of age, (2) current or former (< 2 years) female collegiate athlete, (3) self-reported bra size of B-DD cup, (4) no history of prior breast surgeries (reduction or implants), (5) free from a recent history of musculoskeletal injuries (within the past six months), and (6) free from any history of ACL injuries. The experimental protocol (PRO-FY2020-24) was approved by the University of Memphis Institutional Review Board and all participants provided written informed consent prior to data collection.

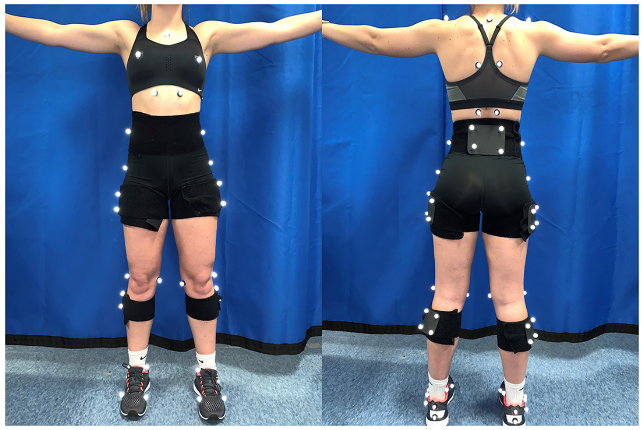

FIGURE 1 | Image of retroreflective marker locations used to define and trunk the skeleton including the trunk, pelvis, right and left thigh, shank and foot. Anatomical markers including the left and right iliac crests, greater trochanters, medial and lateral femoral epicondyles and medial and lateral malleoli as well as the first and fifth metatarsals were removed prior to dynamic testing.

Experimental Equipment

Participants were asked to wear spandex shorts and their preferred athletic shoes for testing. Ground reaction forces (GRFs) and three-dimensional kinematics were recorded simultaneously using a 10-camera motion capture system (250 Hz, Qualisys AB, Goteburg, Sweden) and two force platforms (1,500 Hz, AMTI Inc., Watertown, MA, USA) embedded in the laboratory floor. The skeleton was modeled using 14 mm retro-reflective markers and included trunk and pelvis, as well as left and right thigh, shank, and foot segments. Retro-reflective markers were placed bilaterally on the participant's lower extremity and trunk in order to track individual segment motion during the double-leg landing task. The pelvis, thigh, and shank were tracked using rigid clusters of four retroreflective markers. The rearfoot was tracked using three individual retroreflective markers placed over the superior, inferior and lateral calcaneus. The trunk was defined using individual markers placed over the left and right acromion processes and the right and left iliac crests (**Figure 1**). The trunk segment was tracked using individual markers placed on the skin over the superior sternum, the spinous process of the first thoracic vertebra (T1), the left and right transverse processes of the sixth thoracic vertebrae (T6), the left and right transverse processes of the 12th thoracic vertebra (T12) and the anterior portion of the 10th osteochondral junction. Breast motion was tracked using individual markers placed over the superior sternum and left and right nipples. Anatomical markers were placed over the left and right iliac crest, and trochanters. Anatomical markers were also be placed over the medial and lateral femoral epicondyles, medial and lateral malleoli, and the first and fifth metatarsal heads. After a standing calibration, anatomical markers were removed leaving only the tracking markers on the breast, trunk, pelvis, thigh, shank, and rearfoot.

FIGURE 2 | Anterior, posterior and lateral views of a participant with D-Cup sized breasts in the high support Nike Alpha **(A–C)** and low support Nike Indy **(D–F)**. The athlete was classified as a D-Cup based on the difference between her bust and underbust circumferences (Bust: 84 cm; Underbust: 73.5 cm; Difference: 10.5 cm). The high support sports bra is designed to lift and compress the breast tissue while the low support sports bra is not designed with these features.

Experimental Protocol

Participants visited the Exercise Neuromechanics Research Laboratory at the University of Memphis once for examination and testing. Participants were screened for inclusion criteria, completed a written Physical Activity Readiness Questionnaire (PAR-Q), and provided written informed consent. Each testing session occurred in the following order: (1) measurement of anthropometric variables including age, height (cm), weight (kg), breast size (cm), and rib cage size (cm), (2) warm-up exercises, (3) placement of measurement sensors, and (4) completion of the dynamic testing protocol. The dynamic testing protocol consisted of a double-leg step-off landing task in each of three breast support conditions including low support (LOW), high support (HIGH), and no support (CON).

The LOW conditions required the participant to wear a sports bra that is described by the manufacturer as having "light" support for low-impact workouts. The low support sports bras offered the breasts limited support. The low support sports bra was the Nike Indy (Nike Inc., Beaverton, OR, USA). The fabric of the sports bra includes a body and lining made of 88% recycled polyester and 12% spandex, center back mesh and bottom hem made of 81 percent nylon and 19 percent spandex, elastic made 84 to 85% nylon and 15 to 16% spandex, interlining made of 80% polyester and 20% spandex, pad top fabric and pad back fabric made of 100% polyester, and pad made of 100 percent polyurethane. The HIGH condition required the participant to wear a sports bra that is described by the manufacturer has having their "highest" level of support with a compressive feel for minimal bounce. The high support sports bra was the Nike Alpha (Nike Inc., Beaverton, OR, USA). The fabric of the sports bra includes a body and back lining insets made of 79% nylon and 21% spandex, mesh and mesh lining made of 81% nylon and 19% spandex, pad made of 100-polyurethane, and pad back fabric made of 100 percent polyester. The CON condition required the participant to complete the protocol bare chested with no sports bra and no breast support. CON condition was optional for participants. The purpose of the control condition is to compare data from previous studies to the current study. Sizes of the low and high support sports bras were determined based on fitting described by the manufacturer. **Figure 2** depicts the breast support provided by each sports bra in a representative participant. The protocol was repeated in each randomized support condition (LOW and HIGH) while the CON condition was completed last.

The protocol consisted of a double-leg landing task which required the participant to step-off of a 40-cm box and land bilaterally with one foot on each force platform. A box height of 40 cm was selected as this height is commonly within the maximum vertical jump height of most female athletes. A successful trial was characterized by the participant landing from the box with simultaneous left and right ground contacts with one foot on each of the two force platforms. For foot contact, simultaneous ground contact was defined as having both feet strike the force platforms within a two frame (8 ms) window. Participants completed a total of five successful trials. The participants were allowed to familiarize themselves with the landing task for a period of several minutes until they reported their comfort. Participants performed the familiarization protocol prior to each support condition.

Data Analysis

Landing data were analyzed from IC to an instant 100 ms after contact (INI). The energy absorbed during this period has been associated with injury biomechanics (Norcross et al., 2010) and includes the period in which the ACL is most likely to experience significant injury (Krosshaug et al., 2007; Koga et al., 2010; Bates et al., 2020). IC was determined as the instant at which vertical GRF exceeds a threshold of 20 N and remained above this threshold for a period >0.010 s. A 20 N threshold was selected as it represented a value more than 3 standard deviations above the mean baseline GRF value while the 0.010 s duration represented four frames of data and 10% of the window of analysis, thereby removing potential artifacts being identified as events of interest. Visual 3D (C-Motion Inc., Bethesda, MD, USA) was used to create a six degree-of-freedom kinematic model as well as filter kinematic and GRF data. The six degree-of-freedom model allows each modeled segment to move independently in three translational and three rotational directions. Retroreflective marker trajectories and GRF data were filtered using a fourth-order, zero-lag Butterworth lowpass filter with cutoff frequencies of 10 and 40 Hz, respectively (Smith et al., 2020). Sagittal and frontal plane knee joint angles and moments as well as sagittal plane trunk segment angles were calculated using Visual 3D (C-Motion Inc., Germantown, MD). Vertical breast displacement was calculated as the difference in position of the nipple markers relative to the position of the sternum marker within the plane of the trunk from IC to INI. The adjustment of the local axis system allows for the calculation of vertical breast displacement relative to the trunk. Custom software (MATLAB 2021a, MathWorks, Natick, MA) was used to calculate vertical breast displacement, sagittal and frontal plane knee joint angles at IC and at INI as well as peak sagittal and frontal plane knee joint moments between IC and INI.

Statistical Analysis

A 1 x 3 (task by breast support level) repeated measures analysis of covariance (ANCOVA) was used to determine the effect of breast support level on knee joint and trunk biomechanics while controlling for the effect of breast size. Breast size was quantified as the difference between bust and underbust circumferences (in cm). An ANCOVA was selected to control for the potential effects of breast size on knee joint and trunk biomechanics.

In the presence of a significant main effect of support level, *post-hoc* pairwise comparisons were performed using paired samples *t*-tests to determine source of the significant interaction. A Holm-Bonferroni Correction was performed to adjust the level of significance for multiple comparisons (Holm, 1979). To conduct this correction, the *p*-values for *post-hoc* pairwise comparisons were placed in ascending order (from smallest to largest) and compared to the adjusted level of significance. As three pairwise comparisons were performed, significance for the first *post-hoc* comparison was set at $p < 0.017$ ($p < 0.05/3$) while significance for the second *post-hoc* comparison was set at $p < 0.025$ ($p < 0.05/2$) and significance for the third *post-hoc* comparison was set at $p < 0.05$ ($p < 0.05/1$). The sequential adjustment of the *p*-value is designed to reduce the risk of Type I error associated with multiple comparisons while also maintaining sufficient statistical power. Cohen's d estimates of effect sizes were also reported to further evaluate the effect of breast support on trunk and knee joint biomechanics (Cohen, 1988). Cohen's d values were interpreted as follows: small, $d < 0.2$; moderate, $0.2 < d < 0.8$; large, $d > 0.8$. Significance for omnibus testing was set at $p < 0.05$ while *post-hoc* alpha levels were adjusted as previously described. All statistical comparisons were conducted using SPSS (IBM, Armonk, New York).

RESULTS

Participants

Table 1 presents a summary of participant anthropometrics. Participants had an average age of 20.9 (± 1.7) years, average

TABLE 1 | Participant anthropometric values including age, height, weight, bust and underbust circumferences, breast size and sport participation.

Subject	Age	Height (cm)	Mass (kg)	Bust (cm)	Underbust (cm)	Breast size (cm)	Breast size (Cup)	Sport	Note
S1	24	162	56.6	83	71.5	11.5	D	Track and Field	
S2	21	165	60.8	84	73.5	10.5	D	Soccer	
S3	21	164	53.5	82.5	73	9.5	C	Soccer	
S4	19	167	65	86.5	70	16.5	D	Soccer	Removed
S5	19	172.2	60.6	80	71.5	8.5	C	Soccer	
S6	23	172	65.8	85.5	78	7.5	B	Track and Field	
S7	22	165.1	56.3	79.5	72.5	7	B	Soccer	
S8	23	172	59.87	82.5	76	6.5	B	Volleyball	Removed
S9	21	163.1	60.6	85	72.5	12.5	D	Soccer	
S10	20	181.2	73.3	82.5	75.5	7	B	Volleyball	
S11	20	178.4	74.7	86	80.5	5.5	B	Volleyball	
S12	18	167.7	73.6	88	78.5	9.5	C	Volleyball	
S13	20	181	70.5	83.5	75.5	8	C	Volleyball	
S14	21	170.3	61.5	85.5	71.5	14	D	Softball	
Mean	20.8	170.2	64.0	83.8	74.5	9.3			
SD	1.6	6.9	7.4	2.5	3.1	2.5			

height of 170.1 (± 6.4) cm, average weight of 63.8 (± 6.9) kg, average bust circumference of 83.9 (± 2.4) cm, and average rib cage circumference of 74.3 (± 3.1) cm. No comparisons were made between individuals of different breast sizes.

Breast Displacement

Increasing levels of breast support were associated with reductions in vertical breast displacement (**Figure 3**) during the double-leg landing task for the left ($F = 3.0$, $p < 0.001$) and right breasts ($F = 3.4$, $p < 0.001$). Breast displacement was greater in the CON compared to LOW (Left: $p < 0.001$, $d = 0.92$; Right: $p < 0.001$, $d = 0.74$) and HIGH (Left: $p < 0.001$, $d = 1.37$; Right: $p < 0.001$, $d = 1.20$) breast support conditions while breast displacement was also greater in the LOW compared to HIGH support conditions (Left: $p < 0.001$, $d = 0.66$; Right: $p < 0.001$; $d = 0.067$).

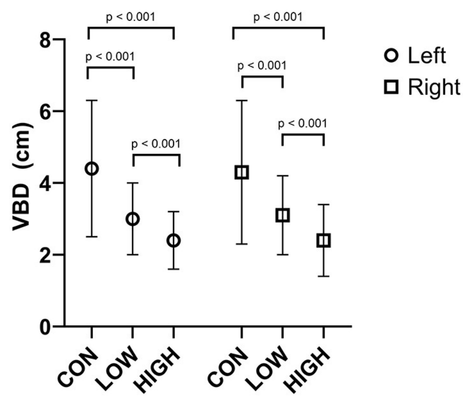

FIGURE 3 | Average vertical breast displacement (VBD) of the left and right breast in the CON, LOW and HIGH support conditions during the double-leg landing task. Displacements are presented in cm.

Knee Joint Angles

At IC, level of sports bra support was not associated with changes in knee flexion angles for either left ($F = 1.25$; $p = 0.166$) or right ($F = 1.42$; $p = 0.146$) legs. Moreover, no effect of sports bra support was observed for knee joint valgus angles for either left ($F = 0.60$; $p = 0.284$) or right ($F = 0.65$; $p = 0.284$) legs.

At INI, level of sports bra support was associated with altered knee joint flexion angles for both left ($F = 3.40$; $p = 0.029$) and right ($F = 6.94$; $p = 0.008$) legs (**Table 2**). For the left leg, no differences were observed in knee flexion angles at INI between the CON and LOW conditions ($p = 0.370$, $d = 0.72$) or the LOW and HIGH conditions ($p = 0.167$, $d = 0.27$) while CON condition was associated with greater knee flexion angles than the HIGH condition ($p = 0.039$, $d = 0.58$). For the right leg, knee flexion angles at INI were greater in the CON compared to the LOW ($p = 0.009$, $d = 0.50$) and HIGH conditions ($p = 0.019$, $d = 0.52$). However, no differences were observed between the LOW and HIGH conditions ($p = 0.493$, $d = 0.01$).

Knee valgus angles at INI (**Table 2**) were altered by increasing levels of sports bra support for both left ($F = 11.01$; $p = 0.003$) and right ($F = 11.0$; $p = 0.011$) legs. The CON condition was associated with greater knee valgus angles than either the LOW (Left: $p = 0.002$, $d = 0.48$; Right: $p = 0.003$, $d = 0.73$) or HIGH conditions (Left: $p = 0.001$, $d = 0.76$; Right: $p = 0.003$, $d = 1.28$). No differences in knee valgus angles were observed between the LOW and HIGH conditions (Left: $p = 0.355$, $d = 0.30$; Right: $p = 0.362$, $d = 0.30$).

Knee Joint Moments

Figure 4 presents peak knee joint extension moments during the double leg landing task. Level of sports bra supports had no effect

TABLE 2 | Knee joint kinematics during the double-limb landing task.

Limb	Condition	Flexion Angle at IC (°)	Valgus Angle at IC (°)	Flexion Angle at INI (°)	Valgus Angle at INI (°)
Left	Control	19.2 ± 4.4	−0.4 ± 3.9	68.8 ± 4.3	−5.1 ± 6.9
	Low	20.4 ± 6.9	0.5 ± 2.9	67.6 ± 7.0	−2.0 ± 6.1[a]
	High	17.9 ± 4.7	0.7 ± 2.8	66.2 ± 4.7[a]	−0.2 ± 6.0[a]
	p-value	0.166	0.284	**0.029**	**0.003**
Right	Control	19.4 ± 4.8	−0.7 ± 2.6	69.0 ± 4.9	−6.5 ± 5.3
	Low	18.3 ± 5.9	0.6 ± 3.2	66.3 ± 5.8[a]	−2.1 ± 6.7[a]
	High	18.5 ± 5.4	0.9 ± 2.0	66.3 ± 5.5[a]	−0.4 ± 4.2[a]
	p-value	0.146	0.284	**0.008**	**0.011**

[a]Denotes significant difference compared to CON support condition.
Presented as mean ± SD. Bold values represent statistical significance ($p < 0.05$).

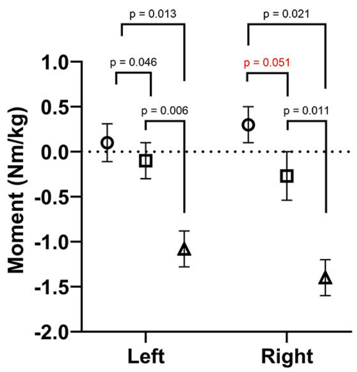

FIGURE 4 | Peak knee extension moments of the left and right leg in the CON, LOW and HIGH support conditions during the double-leg landing task. Joint moments are presented in Nm/kg.

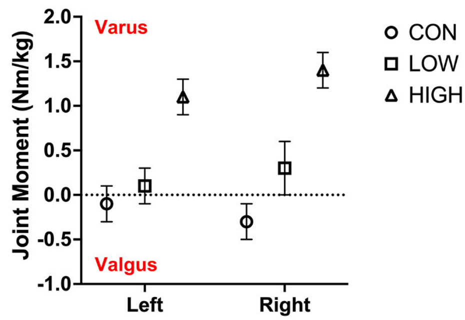

FIGURE 5 | Peak knee valgus moments of the left and right leg in the CON, LOW and HIGH support conditions during the double-leg landing task. Joint moments are presented in Nm/kg.

on peak knee joint moments for left ($F = 0.96$; $p = 0.216$) or right ($F = 4.22$; $p = 0.261$) legs.

Peak valgus moments were reduced with greater levels of breast support during the double-leg landing task (**Figure 5**). For the left leg, peak knee valgus moments were reduced with increasing breast support ($F = 3.91$; $p = 0.033$). *Post-hoc* comparisons revealed greater knee valgus moments in the CON compared to LOW ($p = 0.046$, $d = 0.98$) and HIGH ($p = 0.013$, $d = 5.75$) while the LOW was associated with greater knee valgus moments than the HIGH ($p = 0.006$, $d = 4.90$). For the right leg, increasing levels of breast support were associated with smaller peak knee valgus moments ($F = 4.00$; $p = 0.038$). Pairwise comparisons revealed no differences between the CON and LOW conditions ($p = 0.051$, $d = 2.40$) while the CON was associated with greater peak knee valgus moments than the HIGH support condition ($p = 0.021$, $d = 8.5$). Further, the LOW condition was associated with greater peak knee valgus moments than the HIGH condition ($p = 0.011$, $d = 4.76$).

Trunk Angles

At IC, increasing levels of breast support were associated with greater trunk flexion (**Table 3**; $F = 4.59$; $p = 0.024$). *Post-hoc* analyses revealed no differences in trunk flexion angles between the CON and LOW support conditions ($p = 0.142$, $d = 0.24$) while trunk flexion angles were greater in the HIGH compared to CON ($p = 0.006$, $d = 0.53$) and LOW support conditions ($p = 0.020$, $d = 0.29$). Similarly, increasing levels of breast support were associated with greater trunk flexion at INI ($F = 15.3$; $p = 0.001$). Pairwise comparisons demonstrated that trunk flexion angles were greater in the LOW ($p = 0.001$, $d = 0.58$) and HIGH conditions ($p = 0.001$, $d = 0.99$) compared to CON condition while trunk flexion angles were greater in the HIGH compared to LOW support conditions ($p = 0.003$, $d = 0.38$).

DISCUSSION

The purpose of the current study was to determine the effects of breast support level on knee joint and trunk biomechanics in female collegiate athletes during a double-leg landing task. The major findings of this study were that increasing levels of breast support were associated with smaller peak knee flexion angles, smaller peak knee valgus angles and smaller peak knee valgus moments. Further, greater breast support was also associated with greater trunk flexion at IC and greater peak trunk flexion during the first 100 ms following ground contact.

TABLE 3 | Trunk angles at IC and at INI during the double-limb landing task as well as the statistical results of the omnibus ANCOVA.

Event	CON	LOW	HIGH	*p*-value
IC	−0.5 ± 2.5	0.1 ± 2.5	0.8 ± 2.4[a,b]	**0.024**
INI	−1.4 ± 1.8	−0.2 ± 2.3[a]	0.7 ± 2.4[a,b]	**0.002**

[a]*Denotes significant difference compared to CON support condition.*
[b]*Denotes significant difference compared to the LOW support condition.*
Presented as mean ± SD.

Knee joint flexion is a major contributor to load attenuation during a landing task (Zhang et al., 2008). The current findings demonstrated that greater levels of breast support were associated with reduced knee flexion and knee flexion excursions within the first 100 ms following IC. When considering the lower extremity as a linear spring, knee flexion is a dampening movement (Farley and Morgenroth, 1999; Powell et al., 2014, 2016, 2017). It is postulated that greater knee flexion observed in the low support conditions (CON and LOW) represents a neuromuscular strategy associated with less leg stiffness (greater compliance) which would decelerate the pelvis and trunk along with the passive breast tissue over a longer period of time, decreasing the vertical accelerations of the breast tissue to reduce breast pain. Conversely, in the high support condition, the breast tissue was constrained by the sports bra reducing breast accelerations during the landing task which allowed the participants to land with a preferred landing pattern with greater stiffness. Increased leg stiffness has been suggested to be indicative of better athletic performance (Butler et al., 2003).

Though knee flexion excursions were reduced with increasing breast support, no differences were observed in peak knee extension moments between the breast support conditions. When considering the lower extremity as a torsional spring, the combination of similar knee extension moments and reduced knee flexion excursions would result in greater knee joint stiffness and greater joint loading during the landing task (Butler et al., 2003; Powell et al., 2017). Previous research has shown that greater joint stiffness values are associated with greater vertical loading rates (Butler et al., 2003; Williams et al., 2004; Powell et al., 2017) and greater peak vertical ground reaction forces (Butler et al., 2003; Williams et al., 2004; Powell et al., 2017; Arnwine and Powell, 2020), each of which is associated with an increased risk of musculoskeletal injury (Whiting and Zernicke, 1998). Due to the short duration of the analysis period following IC (100 ms), the biomechanics of the landing task were the result of a predicted mechanical requirement of the landing task and were not the result of a feedback dominant motor pattern. Evidence has demonstrated that long latency reflex control (involving sensory processing by supraspinal structures) of lower leg muscle activation presents with latencies >100 ms (Tsuda et al., 2001, 2003). Therefore, we propose that the greater knee flexion excursions associated with the low breast support conditions (CON and LOW) were the result of a predictive motor control pattern selected to increase lower leg compliance and reduce accelerations of the passive breast tissue during the landing task.

A secondary outcome of greater knee flexion and leg compliance in the lower breast support conditions (CON and LOW) during the landing task is an expansion of the available knee joint range of motion in the frontal and transverse planes (Nordin and Frankel, 2012). The current data demonstrated that in the low breast support conditions (CON and LOW), peak knee valgus angles were greater than in the HIGH breast support condition. Greater knee valgus during a landing task has been associated with reduced neuromuscular control and a greater risk of ACL injury (Hewett et al., 2005; Kernozek et al., 2005; Pappas et al., 2007). Though the differences in knee valgus angles at INI between breast support conditions were small (∼3°-4°), research has suggested that deviations in frontal plane knee joint angle as small as 2° can result in meaningful reductions in the external load required to rupture the ACL (Chaudhari and Andriacchi, 2006). The mechanical effect of greater knee valgus angles is supported by the current findings which demonstrated reduced knee valgus moments in the greater breast support conditions.

Trunk motion has been suggested to modify knee joint biomechanics during load attenuation tasks including single leg squatting and landing tasks (Blackburn and Padua, 2008, 2009; Kulas et al., 2010, 2012). Using a modeling approach, Kulas et al. (2012) revealed that a moderate forward trunk lean was associated with lower peak ACL forces and strains compared to a minimal forward trunk lean during a single-leg squat. Similarly, during a double-leg landing, Kulas et al. (2010) demonstrated that individuals that land with moderate trunk flexion exhibit less knee anterior shear forces and greater hamstrings muscle forces compared to individuals that land with an extended trunk position. Functionally, the hamstrings muscle group acts to protect the ACL by limiting anterior translation of the tibia relative to the femur. Moreover, an intrinsic ACL-hamstrings reflex pathway exists to provide active, muscular support to an ACL that is experiencing strain (Tsuda et al., 2001). The findings of the current study demonstrate that greater breast support was associated with increased trunk flexion angles at IC as well as peak trunk flexion angles. Therefore, these data suggest that the high breast support condition was associated with trunk biomechanics that are indicative of a lower risk of ACL injury compared to low breast support conditions (CON or LOW).

While the current study presents novel findings pertaining to the influence of breast support on knee joint and trunk biomechanics, the authors acknowledge several limitations of the current study. One limitation of the current study is the assumption that the participants are wearing the correct sports bra size and therefore the sports bra support level based on their self-known bra size. However, research has suggested that up to 80% of females are wearing the incorrect bra size (McGhee and Steele, 2010; Hupprich et al., 2020). While the participants were measured for the "correct" bra size using bust and rib cage circumferences, this technique has been criticized for its inaccuracies in measuring for bra size (McGhee and Steele, 2011). However, this specific technique to measure for bra size is commonly used and feasible for the entire female population, which is why it was used. A second limitation of the current study is the small sample size with the power analysis suggesting only 12 participants. The small sample size of only 12 participants may injure any generalizations made to the population. However, even with this small sample size, the study was sufficiently powered to find significant differences between breast support conditions. Furthermore, while 12 participants were required for the power analysis, not all participants completed the CON condition which limited comparisons. Therefore, an additional two participants were recruited to reach the required number of 12, and a total of 14 participants were collected. Another limitation of the current findings pertains to the measurement accuracy of motion capture systems in frontal plane kinematics. Specifically, it is known that skin artifact can negatively affect accuracy of marker-based motion capture systems (Chiari et al., 2005; Leardini et al., 2005). Further, these errors disproportionately effect frontal and transverse plane kinematic calculations. As such, the current findings should be view in light of these limitations in motion capture.

CONCLUSIONS

Greater breast support was associated with a multi-joint biomechanical adaptation characterized by reduced knee flexion, reduced knee valgus and greater trunk flexion angles. These movement profiles are associated with lower risks of traumatic knee injury suggesting that breast support is an important consideration for optimal sport performance and injury prevention. Future research should expand the current analysis to investigate altered contributions of the ankle and hip joint as well. Moreover, lower extremity stiffness and its interaction with trunk biomechanics should also be investigated.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by University of Memphis Institutional Review Board (PRO-FY2020-24). The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

HF, AN, JS, and JH collected data and prepared the initial draft of the manuscript. HF, AN, JS, JH, and DP performed data analysis. All authors were involved in study conception and design. All authors revised, edited and approved the final manuscript.

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Changes in Muscle Activation During and After a Shoulder-Fatiguing Task: A Comparison of Elite Female Swimmers and Water Polo Players

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This study compared female athletes with different aquatic sports expertise in their neuromuscular activation before, during, and after a shoulder internal rotation fatigue protocol. Eleven water polo players, 12 swimmers, and 14 controls completed concentric maximal voluntary external and internal shoulder rotations before and after a fatigue protocol consisting of concentric internal rotations at 50% of maximal voluntary contraction for at least 3 min or until reporting a rating of perceived effort RPE of 8/10 or higher. Muscle activation was measured for the maximal voluntary contractions, as well as for the first (T1), middle (T2), and third (T3) minute of the fatigue protocol using surface electromyography (EMG) on pectoralis major, anterior and posterior deltoid, upper and middle trapezius, and latissimus dorsi. Intramuscular EMG was used for supraspinatus, infraspinatus, and subscapularis. Pre-fatigue internal rotation torque was significantly correlated with shorter task duration ($r = -0.39$, $p = 0.02$), with water polo players producing significantly greater torque than controls but having significantly lower endurance. Swimmers demonstrated decreased latissimus dorsi activation at T3 compared to T2 ($p = 0.020$, $g = 0.44$) and T1 ($p = 0.029$, $g = 0.74$), differing from water polo players and controls who exhibited increased agonist activation and decreased activation of stabilizers. Comparing the pre-fatigue to the post-fatigue maximal shoulder rotations, water polo players had decreased activation in subscapularis ($p = 0.018$, $g = 0.67$); all groups had decreased activation in latissimus dorsi ($p < 0.001$), though swimmers demonstrated a large effect ($g = 0.97$); and controls had decreased activation in supraspinatus ($p = 0.005$, $g = 0.71$). Together, these results suggest that sports expertise may be associated with different muscle activation both while and after fatigue is induced. Further research should continue to explore sports-specific patterns of muscle recruitment and fatigue adaptations, as well as if certain strategies are adaptive or maladaptive. This may have important consequences for injury prevention among athletes who perform repetitive overhead movements in their sports and who are susceptible to overuse injuries.

Keywords: electromyography—EMG, shoulder, female athletes, water polo players, swimmers, neuromuscular activation, neuromuscular fatigue

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INTRODUCTION

Studies suggest that 35–45% of athletes in overhead sports interrupt training due to shoulder problems, with as many as 75% of swimmers having experienced shoulder pain throughout their athletic career (McMaster and Troup, 1993; Joshi et al., 2011; Aliprandi et al., 2013; Matzkin et al., 2016). This high prevalence calls for more research to increase understanding of the mechanisms leading to injury and, in turn, guide improved preventative strategies. The freestyle stroke is frequently used by both competitive swimmers and water polo players during training. The average swimmer executes ≈30,000 shoulder revolutions per week for 50 weeks of the year (Bak and Faunø, 1997; Weldon and Richardson, 2001; Matzkin et al., 2016). Though water polo players swim fewer kilometers, they also perform large numbers of overhead throws during each training session (Wheeler et al., 2013), which also lead to fatigue. Previous research has identified the scapular stabilizers (supraspinatus, infraspinatus, subscapularis, and middle trapezius) and internal rotators (latissimus dorsi and pectoralis major) as important contributors to the freestyle stroke (Nuber et al., 1986; Pink et al., 1991; Weldon and Richardson, 2001; Bedi, 2011). For water polo, the posterior deltoid, supraspinatus, and middle trapezius muscles play active roles in the cocking phase of the throwing task, and the pectoralis major and anterior deltoid in the follow through (Fleisig et al., 2009; Weber et al., 2014; Yaghoubi et al., 2014). When throwing while in water, the lack of a base of support requires the shoulder joint to produce more force than overhead throwing performed on land (Feltner and Taylor, 1997). Water polo players also swim freestyle in a head-up position, requiring more activation from the upper trapezius than in a competitive freestyle stroke.

Fatigue is considered a major contributor to the rate of injury, as it affects shoulder girdle stability (Dale et al., 2007; Joshi et al., 2011; Matthews et al., 2017). Madsen et al. (2011) found that there was a progressively higher rate of abnormal scapular motion (scapular dyskinesis) in a group of healthy swimmers as their training session went on and fatigue developed. Since the occurrence of scapular dyskinesis is linked to shoulder injuries in swimmers (Bak and Faunø, 1997; Kibler and McMullen, 2003), this indicates that fatigue plays a key role in the development of those injuries. Fatigue can be operationalized as decreased maximal force production capacity, or as changes in muscle activation and/or increased rate of perceived effort while maintaining task performance (Vøllestad, 1997; González-Izal et al., 2012; Enoka and Duchateau, 2016). Force and torque outputs provide methods to measure the functional effects of fatigue on performance and can be quantified using an isokinetic dynamometer (Baltzopoulos and Brodie, 1989; Cools et al., 2003). Measures using electromyography (EMG), have shown EMG amplitude (e.g., root-mean square [RMS]) to increase with fatigue under some conditions (Krogh-Lund and Jørgensen, 1991; Smith et al., 2016). This has been interpreted as increases in the recruitment of more and bigger motor units, as well as in increased motor unit discharge rates to continue performing a task. However, this is typically seen in repeated and/or prolonged submaximal efforts only, whereas in short, high-intensity tasks, fatigue is rather linked with decreases in activity amplitude (Vøllestad, 1997; González-Izal et al., 2012).

Research in the workplace setting has demonstrated that EMG can be used to compare two different groups to identify muscle activation patterns that are associated with increased upper limb injury risk (Goubault et al., 2020). Similar comparisons of muscle activation among athletes have not been made, but kinematic studies of landing and cutting among female soccer and basketball players revealed that athletes had sports-specific movement control patterns that, consequently, could indicate sports-specific susceptibility to injury (Cowley et al., 2006; Munro et al., 2012). The repetitive nature of upper limb actions performed by water polo players and swimmers makes it relevant to also consider the role of fatigue. Indeed, internal rotation torque has been shown to decrease after fatiguing throwing tasks (Ellenbecker and Roetert, 1999; Mullaney et al., 2005). As well, throughout swimming-specific tasks, increases in EMG amplitude for the latissimus dorsi and triceps brachii were observed (Stirn et al., 2011). However, there is yet to be research that examines if shoulder muscle activation or subsequent changes in activation with fatigue differ based on sports expertise. Therefore, the objective of this study was to investigate whether female swimming and water polo training experience is associated with patterns of shoulder muscle activation during and following the inducement of fatigue. Performance measures, such as torque production capacity and endurance, were also compared. We hypothesized that water polo and swimmers would have greater endurance during a fatigue protocol compared to a control group, that torque production would decrease when fatigued, but that there would be differences in muscle activation between the groups.

METHODS

Participants

Thirty-seven healthy female volunteers were recruited into one of three groups: a control group (CON), an elite water polo group (WP), and an elite swimming group (SW) (**Table 1**). The control group consisted of individuals with no past training in repetitive overhead sports or activities. The elite athlete groups were recruited through the Institut national du sports du Québec network and local national level aquatics teams. The inclusion criteria for the "elite" groups required athletes to have six or more years of experience competing at the national or international level and be participating in 15 h per week or more of sports-specific training (Swann et al., 2015). Based on the classification criteria outlined by McKay et al. (2022), six WP participants were classified as elite/international level (tier 4) athletes and five as highly trained/national level (tier 3) athletes. For the SW participants, two were classified as elite/international level (tier 4) athletes, seven as highly trained/national level (tier 3) athletes, and three as trained/developmental (tier 2) athletes.

All participants reported having no injuries involving their shoulder joint within the previous 6 months. The research protocol was approved by the Centre de Recherche Interdisciplinaire en Réadaptation (CRIR-1247-0517) and all

TABLE 1 | Participant characteristics.

Group	*n*	Age (years)	Height (cm)	Weight (kg)
CON	14	24.7 ± 2.4	166.7 ± 6.8	63.4 ± 9.8
WP	11	22.2 ± 5.6	171.5 ± 6.9	80.0 ± 14.8
SW	12	20.4 ± 2.5	169.1 ± 4.7	68.0 ± 6.3

participants provided written informed consent before the experimental procedure.

Instrumentation

Wireless surface EMG (sEMG; Delsys Trigno Wireless EMG, Natick, MA, USA) sensors (Hermens et al., 1999) were placed on pectoralis major, anterior deltoid, posterior deltoid, upper trapezius, middle trapezius, and latissimus dorsi according to SENIAM guidelines and those established by Barbero et al. (2012). Intramuscular EMG (iEMG) sensors were used to record activity in three muscles using paired hook fine-wire electrodes (Natus Neurology, Middleton, WI, USA): supraspinatus, infraspinatus, and subscapularis. The insertion points were consistent with those in a study by Gaudet et al. (2018a). The exact locations of the electrode placements and insertion points can be found in **Table 2.** All electrodes were attached to the skin using double-sided tape, with additional tape over the sensor, with additional tape over the electrode to reduce any movement during the protocol.

All shoulder rotations were performed using a CON-TREX Multi-Joint Isokinetic Dynamometer (CON-TREX MJ; CMV AG, Dubendorf, Switzerland), with the participant in a prone position on the CON-TREX. The angular velocity of the CON-TREX motor was set to 120°/s to simulate the speed and positioning of a freestyle swimming stroke at peak velocity (Falkel et al., 1987). **Figures 1A,B** show the positioning and the movement for these trials, respectively. The range of motion varied slightly depending on the participant's flexibility (77.8 ± 5.4°) but remained constant during all of the participant's movements throughout the protocol.

Experimental Procedure

After the participant's height and mass were measured, the CON-TREX seat and motor were adjusted such that the head of the motor would form a straight line with the participant's glenohumeral joint and elbow. The skin was shaved and cleaned with alcohol before EMG electrodes were placed. Using the dominant arm, the participant then performed a series of concentric maximal voluntary contractions (MVC) with 1 min of rest between each. Four sets of three repetitions were performed: two sets of maximal shoulder external rotations with completely passive internal rotation, and two sets of maximal shoulder internal rotations with passive external rotation.

The fatigue protocol consisted of repetitive shoulder internal rotations at ≈50% of the individual's maximal internal rotation torque. Participants were provided visual feedback throughout: a screen in front of the participant showed a bandwidth of 42.5–57.5% of their maximal internal rotation torque. The instructions were to perform repeated concentric internal rotations, keeping the torque output within the 15% bandwidth, and to allow the CON-TREX to move the arm back to the starting position during the external rotation. The participant performed two rounds of familiarization prior to starting the fatigue protocol. For the first 20 s of each minute, the participant was given verbal feedback by the researchers to remain within the bandwidth. At the 20-s mark, the researchers told the participant to remain constant within the bandwidth for the following 30 s without verbal feedback. In the last 10 s the participant was prompted to rate their perceived effort using the modified Borg CR10 Scale (Borg, 1998). The researcher specified that the perception of effort should be in relation to only the neck/shoulder. The protocol continued for a minimum of 3 min until the participant reported an RPE of 8/10 or higher, or until the torque output fell significantly outside of the bandwidth three times consecutively. This termination criterion was unknown to the participant. Immediately after completing the fatigue protocol, the participant repeated the four sets of maximal voluntary contractions, this time without rest between sets to limit recovery.

Data Processing

Pre-fatigue and post-fatigue maximal internal rotation and external rotation torque values were identified for each participant. All EMG data were filtered using a zero-lag 2nd order Butterworth bandpass filter (sEMG was bandpassed between 10 and 450 Hz, and iEMG between 10 and 1,000 Hz) and full wave rectified before heartbeats were removed. Additionally, iEMG data underwent a notch filter to remove frequency harmonics. Using a moving window of 100 ms, a peak RMS value was identified for each muscle using the single pre-fatigue MVC contraction that elicited the highest RMS amplitude for the given muscle.

EMG data during the fatigue protocol was partitioned into internal and external rotation using the position data from the CON-TREX. For each internal rotation of the fatiguing task, RMS values were calculated using a 100 ms moving window, and for each muscle, were normalized to the previously determined pre-fatigue maximal rotation peak RMS value. The RMS values for the last five internal rotations before the 50-s mark of the first (T1), middle (T2), and last (T3) minute of the task were averaged to obtain one RMS value for each of the nine muscles. If the task was performed for an even number of minutes, the average of the two middle minutes was taken for T2.

Statistical Analyses

Because assumptions of normality and homogeneity of variance were violated, a Brown–Forsythe test was conducted to compare

TABLE 2 | EMG electrode placements.

Muscle	Electrode Position
Surface electrodes	
Pectoralis Major	Three finger widths medial to the coracoid process of the shoulder.
Anterior Deltoid	Three finger widths distal to coracoid process of the shoulder, in line with the humerus.
Posterior Deltoid	Two finger widths distal to the angle of the acromion, on the line between the acromion and the pinky finger.
Upper Trapezius	Midpoint between the C7 spinous process and the anterior acromion process.
Middle Trapezius	50% between the medial border of the scapula and the spine, at the level of T3.
Latissimus Dorsi	Three finger widths distal and slightly lateral to inferior angle of the scapula.
Intramuscular electrodes	
Subscapularis	Three finger widths superior to the inferior angle of the scapula on the medial side. Insertion occurred with the scapula winged to be able to insert the electrode under the scapula.
Infraspinatus	2.5 cm inferior to the midpoint of the spine of the scapula.
Supraspinatus	1.5 cm superior to the midpoint of the spine of the scapula.

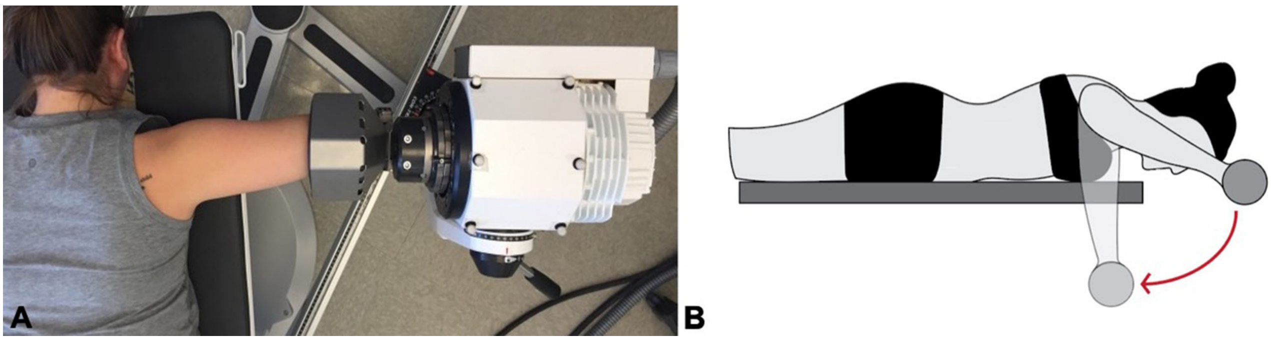

FIGURE 1 | (A) Participant setup on the isokinetic dynamometer. **(B)** Movement pattern of the shoulder internal rotation fatigue protocol.

task duration between groups, followed by Bonferroni *post-hoc* testing. Separate Generalized Estimating Equations (GEE) were performed on external rotation torque, internal rotation torque, the RMS recorded during the pre- and post-fatigue MVC for each muscle, and the RMS recorded during the fatigue protocol. The GEE models for torque included one within-subject variable (Condition, two levels: pre- and post-fatigue), one between-subject variable (Group, three levels: controls, water polo players, swimmers), and included weight as a covariate. For RMS during the fatigue protocol, the GEE model had two within-subject variables (Time, three levels: T1, T2, T2; Muscle, nine levels) and one between-subject variable (Group, three levels). The GEE model for MVC RMS included two within-subject variables (Condition, two levels; Muscle, nine levels: one for each level) and one between-subject variable (Group, three levels: CON, WP, SW). Pairwise comparisons, with Bonferroni correction, of estimated marginal means were carried out when there were statistically significant main effects and interactions. The magnitudes of differences were evaluated with Hedges' g effect sizes.

All analyses were conducted with SPSS (IBM SPSS Statistics for Windows, Version 27.0).

RESULTS

For maximal external rotation torque, there was a main effect of Condition ($X_1^2 = 10.216$, $p = 0.001$) and of Group ($X_2^2 = 6.826$, $p = 0.033$) (**Figure 2A**). All groups demonstrated decreased torque in the fatigued condition ($p = 0.001$, $g = 0.26$) and WP produced higher torque than both CON ($p = 0.031$, $g = 1.64$). Similar main effects for Condition ($X_1^2 = 13.269$, $p < 0.001$) and Group ($X_2^2 = 7.414$, $p = 0.025$) existed for maximal internal rotation torque (**Figure 2B**), with decreased values after the fatigue protocol ($p < 0.001$, $g = 0.33$) and WP producing greater torque than CON ($p = 0.020$, $g = 1.81$).

The task duration for CON, WP, and SW was 8.5 ± 6.0 (mean ± SD), 4.0 ± 0.8, and 6.5 ± 3.6 min, respectively. The Brown–Forsythe test indicated differences between groups ($F(2, 21.828) = 3.917$, $p = 0.035$, **Figure 3**) with *post-hoc* comparisons indicating that CON performed the task significantly longer than WP before reaching the termination criterion ($p = 0.040$, $g = 0.95$). Furthermore, a significant correlation ($r_s(35) = -0.39$, $p = 0.02$) between maximal pre-fatigue internal rotation torque and task duration indicated that higher internal rotation strength was associated with a shorter time to reach the termination criterion.

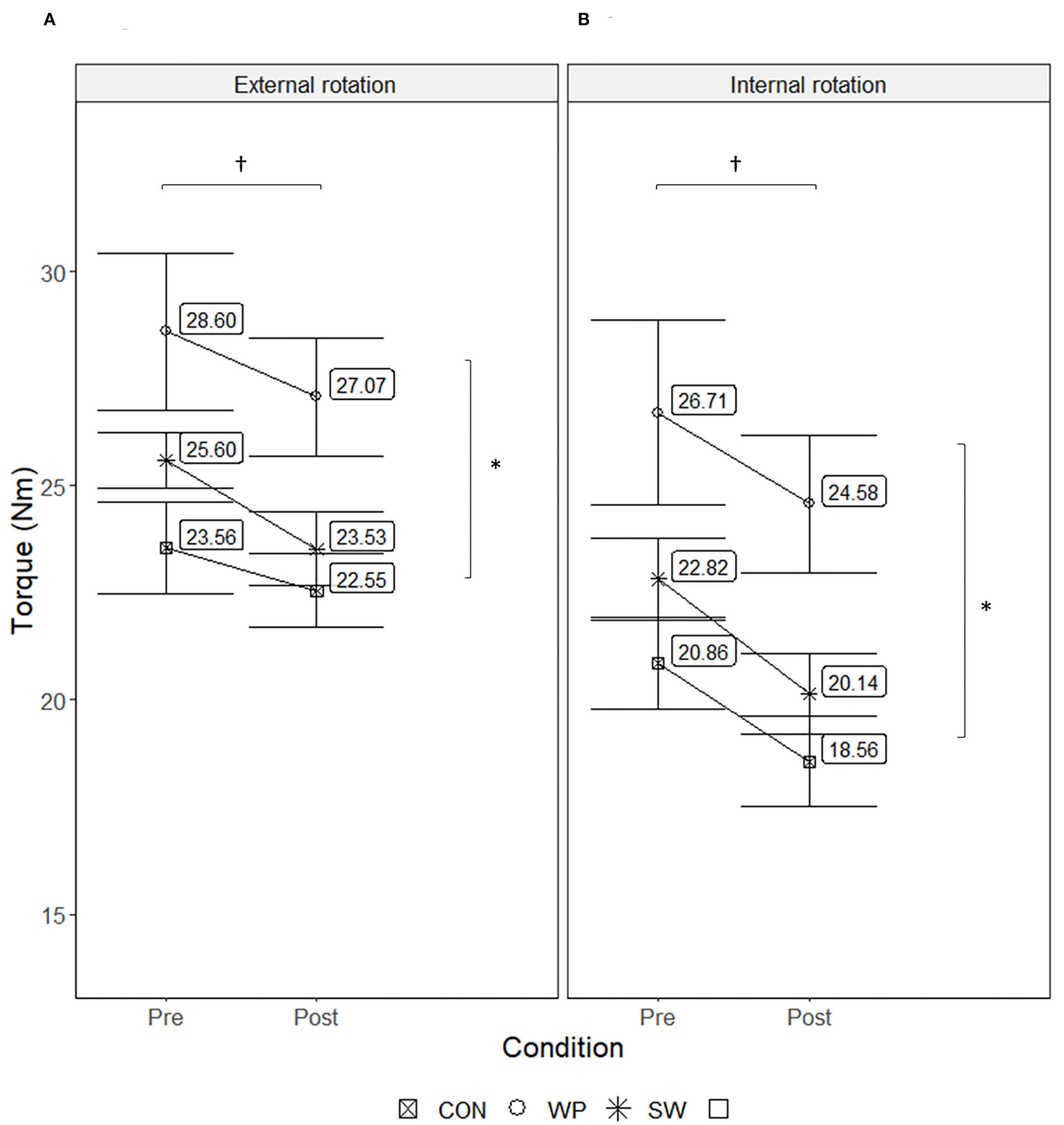

FIGURE 2 | Maximal concentric external **(A)** and internal **(B)** shoulder rotation torque produced by water polo athletes (WP), swimmers (SW), and a control group (CON) before and after a concentric shoulder internal rotation fatigue protocol. Values displayed are estimated marginal means ± SE. †Indicates a significant main effect of the condition, with decreased torque values after the fatigue protocol ($p < 0.05$). *Indicates significant group difference, with WP producing significantly greater torque than CON ($p < 0.05$).

During the shoulder internal rotation fatigue protocol, there were significant Group × Time × Muscle interactions for EMG RMS ($X^2_{32} = 315.904$, $p < 0.001$) (**Figure 4**). Among CON, pectoralis major RMS was lower at T2 ($p = 0.22$, $g = 0.72$) and T3 ($p = 0.015$, $g = 0.85$) compared to T1. Also compared to T1, CON had higher T3 RMS for posterior deltoid ($p = 0.013$, $g = 0.91$) and middle trapezius ($p = 0.023$, $g = 0.72$). Among WP, there were increases in RMS from T1 to T2 for pectoralis major

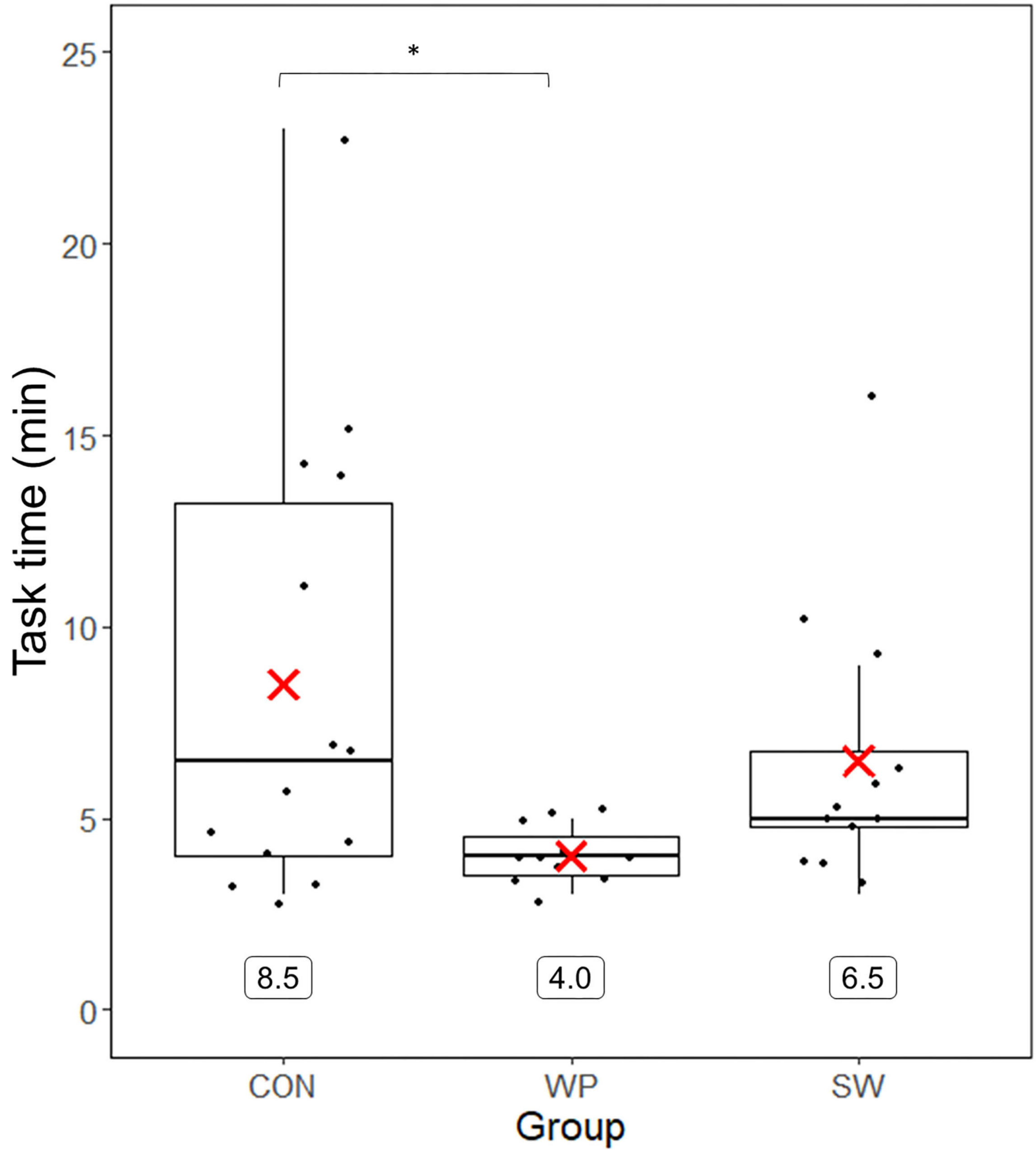

FIGURE 3 | Duration of the fatigue protocol for water polo athletes (WP), swimmers (SW), and a control group (CON). Group means are represented by the red X and displayed numerically below the boxplot. *Significant group difference between CON and WP.

($p = 0.009$, $g = 0.60$) and subscapularis ($p = 0.012$, $g = 0.16$). WP demonstrated decreased posterior deltoid RMS at T2 ($p = 0.005$, $g = 0.74$) and T3 ($p < 0.001$, $g = 0.90$) relative to T1, decreased middle trapezius RMS at T3 compared to T1 ($p = 0.024$, $g = 0.55$) and T2 ($p = 0.004$, $g = 0.20$), as well as decreased supraspinatus RMS at T3 compared to T1 ($p < 0.001$, $g = 0.85$) and T2 ($p = 0.003$, $g = 0.55$). Among SW, only latissimus dorsi RMS changed significantly, with lower RMS at T3 than at T2 ($p = 0.020$, $g = 0.44$) and T1 ($p = 0.029$, $g = 0.74$). At T1, CON had greater RMS for posterior deltoid than SW ($p = 0.008$, $g = 1.04$) and higher subscapularis RMS than WP ($p = 0.030$, $g = 0.98$). At T2, CON had greater anterior deltoid RMS than WP ($p = 0.048$, $g = 0.45$).

For EMG RMS measures during the MVCs, there were also significant Group × Condition × Muscle interactions ($X^2_{16} = 80.033$, $p < 0.001$) (**Figure 5**), indicating group-specific changes in activation. CON demonstrated a decrease in RMS for supraspinatus after the fatigue protocol ($p = 0.005$, $g = 0.71$); WP had lower post-fatigue RMS for anterior deltoid ($p < 0.001$, $g =$

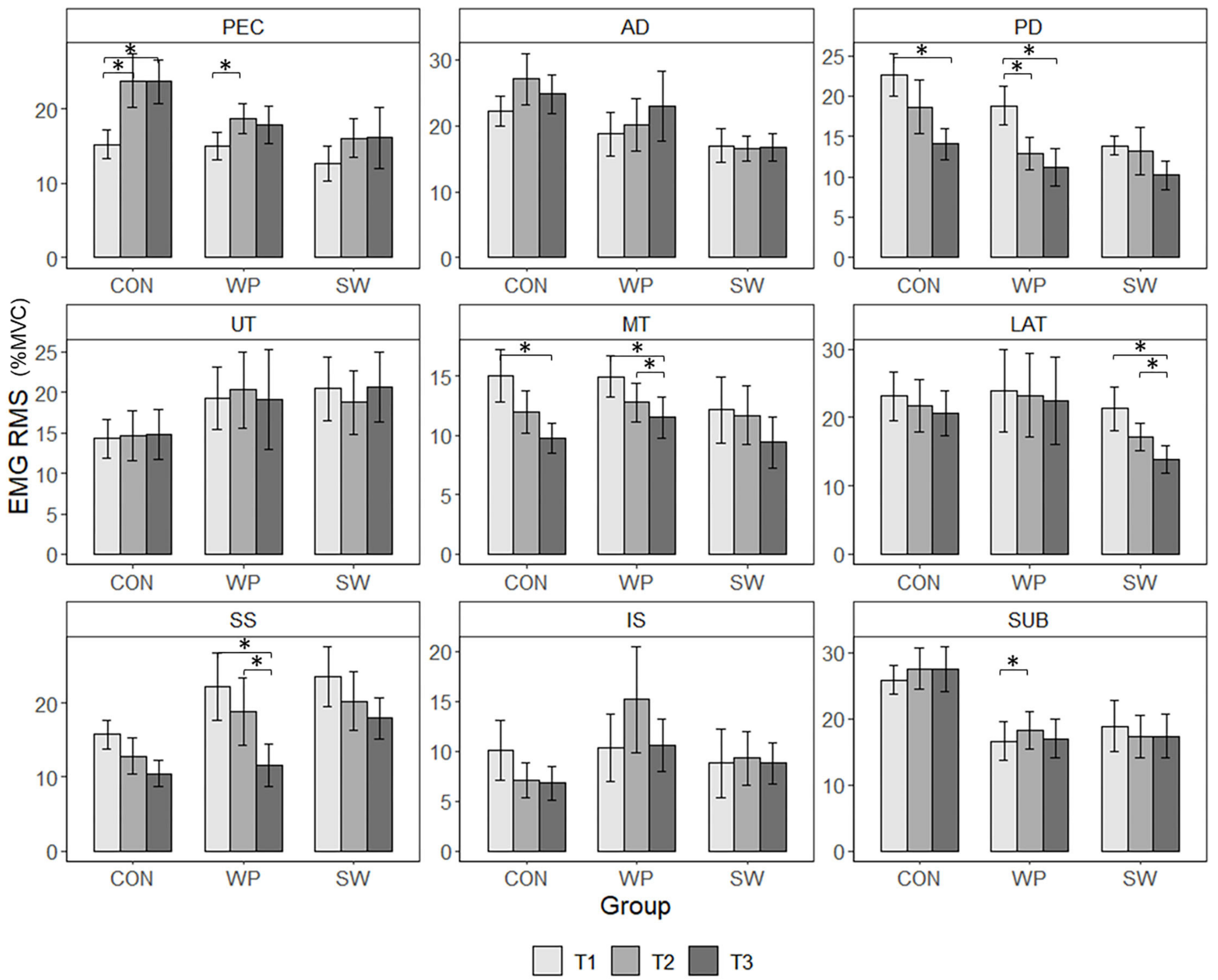

FIGURE 4 | EMG RMS of water polo athletes (WP), swimmers (SW), and a control group (CON) measured during the last five rotations before the 50-s mark of the first (T1), middle (T2), and last (T3) minute of a concentric shoulder internal rotation fatigue protocol. Values are estimated marginal means ± SE measured for pectoralis major (PEC), anterior deltoid (AD), posterior deltoid (PD), upper trapezius (UT), middle trapezius (MT), latissimus dorsi (LAT), supraspinatus (SS), infraspinatus (IS), and subscapularis (SUB). *Indicates a significant group-specific changes in RMS ($p < 0.05$).

0.91), upper trapezius ($p = 0.046$, $g = 0.29$), and subscapularis ($p = 0.018$, $g = 0.67$); and SW had lower post-fatigue RMS for anterior deltoid ($p = 0.001$, $g = 1.37$). Significant Condition × Muscle interactions ($X^2_8 = 111.863$, $p < 0.001$) for EMG amplitude followed by pairwise comparisons indicated that, post-fatigue, all participants had lower RMS for posterior deltoid ($p < 0.001$, $g = 0.731$), latissimus dorsi ($p < 0.001$, $g = 0.69$), and infraspinatus ($p < 0.001$, $g = 0.64$). Hedges' g effect sizes for EMG RMS changes during the fatigue protocol and comparing the pre- and post-fatigue MVCs are listed in **Table 3**.

DISCUSSION

This study was the first to compare fatigue-related changes in shoulder muscle activation patterns between female athletes of different aquatic sports specialties. Among water polo athletes, swimmers, and controls, fatigue was induced by a repetitive shoulder internal rotation task performed at 50% MVC. WP had a shorter task duration than CON but also had higher pre-fatigue torque production than CON. SW and WP showed different patterns of muscle activation during the repetitive shoulder internal rotation task and as a result of the fatigue induced, and all groups demonstrated decreased shoulder stabilizer activation in the post-fatigue MVCs.

Both subjective and objective measures confirmed that fatigue was induced by the internal rotation task. All participants finished the task with higher ratings of perceived effort than when they started. Consistent with traditional definitions of muscle fatigue resulting from physical exertion (Gandevia, 2001),

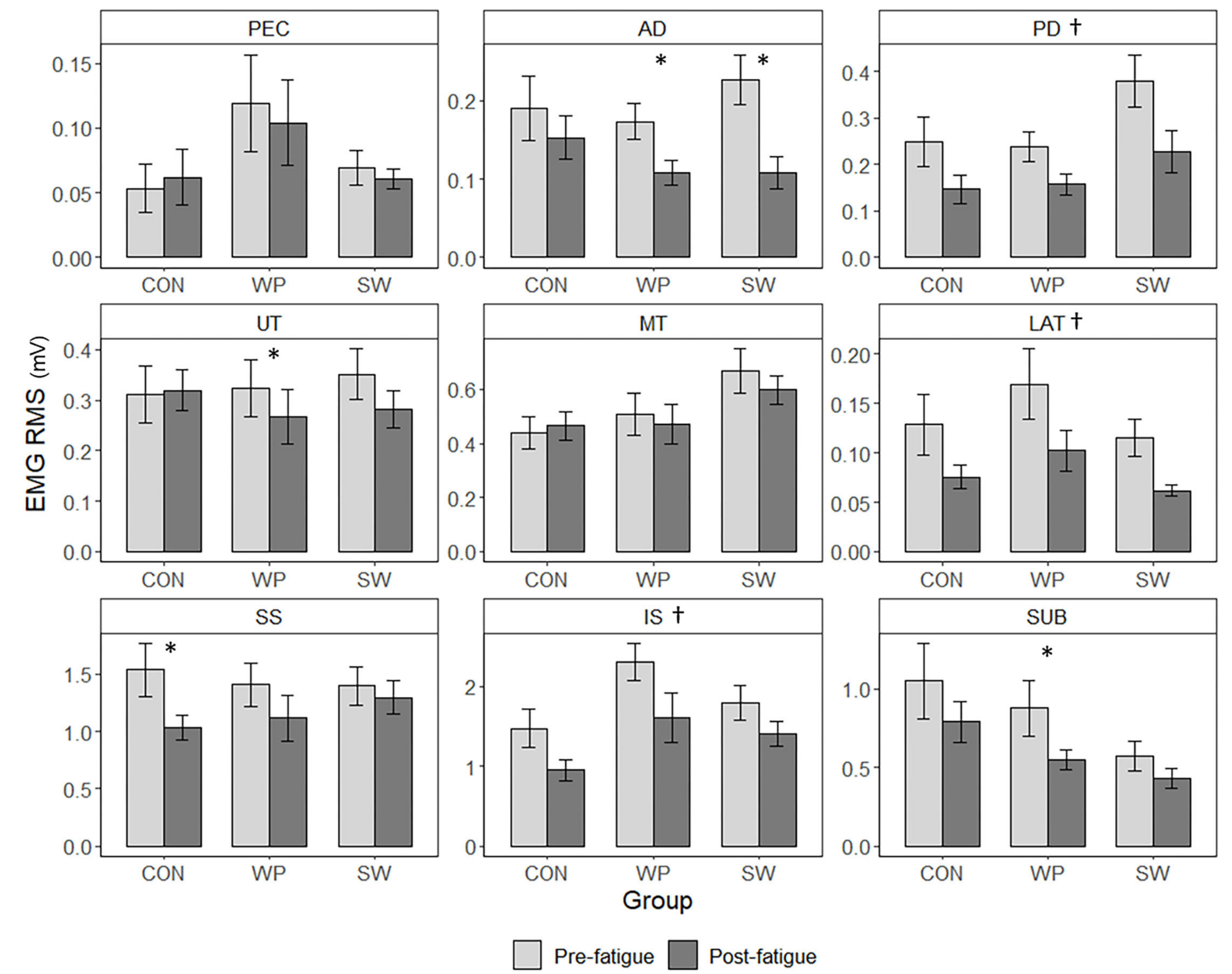

FIGURE 5 | EMG RMS of water polo athletes (WP), swimmers (SW), and a control group (CON) were measured during maximal concentric external and internal shoulder rotations performed before and after a concentric shoulder internal rotation fatigue protocol. Values are estimated marginal means ± SE measured for pectoralis major (PEC), anterior deltoid (AD), posterior deltoid (PD), upper trapezius (UT), middle trapezius (MT), latissimus dorsi (LAT), supraspinatus (SS), infraspinatus (IS), and subscapularis (SUB). †Indicates a significant interaction of Muscle × Condition, with decreased RMS values after the fatigue protocol ($p < 0.05$). *Indicates a significant group-specific decrease in RMS ($p < 0.05$).

there was a reduction of force production capacity, as evidenced by lower maximal internal and external rotation torque values after the fatiguing protocol. Interestingly, CON performed the task longer than WP before reaching the termination criteria. While it may be initially counterintuitive that CON had greater endurance compared to high-performance athletes, this result may be attributable to strength differences between the groups, with WP producing significantly higher internal rotation torque than CON at baseline, even when accounting for weight. Water polo involves forceful and ballistic actions, and this type of training can promote shifts toward type II fiber types (Wilson et al., 2012; Plotkin et al., 2021), in addition to overall increases in muscle fiber cross-sectional area (Folland and Williams, 2007). On top of morphological adaptations to sports-specific training, neurological adaptations, such as enhanced motor unit firing rates, greater motoneuron excitability, and changes in inter-muscle coordination, may allow athletes to produce peak forces closer to their true maximum (Folland and Williams, 2007). In contrast, even when motivated, healthy but untrained individuals are less likely to be able to fully activate their agonists (Westing et al., 1988; Dudley et al., 1990). Thus, the target force of 50% MVC at which the fatigue protocol was performed may have been a lower intensity for CON compared to WP, relative to their true maximum. This would in turn result in a less physically demanding task for CON than it was for WP, allowing them to continue for

TABLE 3 | Hedges' *g* effect sizes for changes in EMG RMS between the first minute (T1), middle minute (T2), and last minute (T3) of the internal rotation fatigue task, as well as between the pre- and post-fatigue maximal voluntary contractions (MVCs).

	CON				WP				SW			
	T1 vs. T2	T1 vs. T3	T2 vs. T3	MVC	T1 vs. T2	T1 vs. T3	T2 vs. T3	MVC	T1 vs. T2	T1 vs. T3	T2 vs. T3	MVC
Pectoralis major	**−0.72***	**−0.85***	0.01	−0.16	**−0.60***	−0.41	0.12	0.13	−0.37	−0.28	0.00	0.54
Anterior deltoid	−0.37	−0.24	0.16	0.31	−0.10	−0.26	−0.17	**0.91***	0.05	0.03	−0.03	**1.37***
Posterior deltoid	0.33	**0.91***	0.42	**0.66†**	**0.74***	**0.90***	0.21	**0.81†**	0.08	0.63	0.32	**0.80†**
Upper trapezius	−0.03	−0.05	−0.01	−0.04	−0.07	0.01	0.06	**0.29***	0.11	−0.01	−0.12	0.42
Middle trapezius	0.38	**0.72***	0.35	−0.12	0.36	**0.55***	**0.20***	0.13	0.04	0.29	0.27	0.27
Latissimus dorsi	0.10	0.18	0.07	**0.56†**	0.03	0.07	0.03	**0.69†**	0.41	**0.74***	**0.44***	**0.97†**
Supraspinatus	0.33	0.72	0.28	**0.71***	0.26	**0.85****	**0.55***	0.42	0.22	0.43	0.18	0.17
Infraspinatus	0.30	0.34	0.04	**0.66†**	−0.30	−0.02	0.30	**0.76†**	−0.04	0.00	0.05	**0.54†**
Subscapularis	−0.16	−0.14	0.01	0.33	**−0.16***	−0.04	0.13	**0.67***	0.11	0.11	0.00	0.49

A negative value reflects an increase in EMG RMS.
†Significant interaction of Muscle × Condition.
**Significant group-specific change in RMS.*

longer. Although there have yet to be studies within female groups investigating if greater initial strength can be linked to higher fatigability, such associations have been made within the context of sex differences research. This suggests the presence of a strength-related mechanism to explain sex differences in fatigability and associated neuromuscular mechanisms (West et al., 1995; Hunter and Enoka, 2001; Hunter et al., 2006; Hunter, 2014).

EMG RMS changes during the fatiguing task were similar between CON and WP. Greater pectoralis major RMS for both groups, and greater subscapularis RMS for WP, at T2 compared to T1, suggest that there was an initial increase in agonist activation. There were no further increases at T3 for these muscles. Instead, there were decreases in RMS for posterior deltoid and middle trapezius for both groups, and supraspinatus among WP. Posterior deltoid, middle trapezius, and supraspinatus are all muscles that are more active in external rotation than in internal rotation (Boettcher et al., 2010; Gaudet et al., 2018a). Thus, as the fatiguing task continued, CON and WP may have used a strategy that first involved increased agonist activation, which was then followed by reduced antagonist activity. While this may have been a strategy to maintain task performance, concurrent activation of stabilizing muscles (co-contraction) is important for joint stability of the shoulder (Sangwan et al., 2014). Interestingly, there was comparatively little change in muscle activation among SW. Based on their sports expertise, they are also the group that would have been most habituated to such a repetitive cyclic task. Movement variability in locomotion is thought to be an adaptive mechanism that could reduce the risk of overuse injury (Bartlett et al., 2007). If this was a strategy more heavily relied upon by SW—who has expertize in repetitive, locomotor upper limb movements—fatigue-related changes in muscle activation among SW may not have been captured by comparing just the last five repetitions of the first, middle, and last minute.

There were also differences and similarities in EMG RMS changes for the MVCs. Decreases in RMS in maximal efforts may represent the fatigue of type II fibers (Cifrek et al., 2009) and/or an inability to sustain initially high firing rates (Gandevia, 2001). In the post-fatigue MVCs, although CON, WP, and SW all shared significantly lower EMG RMS for latissimus dorsi compared to the pre-fatigue condition, the largest effect size for this change was among SW. Interestingly, during the fatiguing task, SW was the only group to exhibit decreases in latissimus dorsi RMS—a muscle that, in swimming, is highly active in the pull-through (i.e., propulsive) phase of all four competitive swimming strokes (Nuber et al., 1986; Pink et al., 1991). In throwing, subscapularis is a major contributor to generating the forces needed for rapid acceleration, but also plays an important role in stabilizing the shoulder joint (Gowan et al., 1987; Escamilla and Andrews, 2009). Only WP demonstrated a significant decrease in subscapularis RMS and similarly, were the only group for which there were indications of subscapularis fatigue during the fatiguing task. Although more research is required to examine EMG parameters and their variability in parallel with fatigue development, these results hint that there is sports specificity underlying group differences in muscle activation between water polo players and swimmers during and after a fatiguing task.

It is also interesting to note that anterior deltoid RMS decreased among WP and SW, supraspinatus RMS significantly decreased among CON, and posterior deltoid and infraspinatus RMS were lower for all groups after the fatiguing task. Shoulder stabilizers have previously demonstrated neuromuscular fatigue induced by a repetitive, maximal, concentric internal and external rotation task (Gaudet et al., 2018a). The current findings suggest that a submaximal shoulder rotation task may also induce fatigue among the shoulder stabilizers. Infraspinatus and subscapularis activity is important for countering the superior shear force produced by supraspinatus and deltoid and minimizing the risk of subacromial impingement (Payne et al., 1997). Consequently, decreased anterior deltoid, posterior deltoid, and supraspinatus activation that occur in conjunction with decreases in infraspinatus and subscapularis RMS could potentially indicate a protective mechanism to maintain balance at the shoulder joint.

One limitation of this study is that the fatigue protocol may have been performed at different intensities relative to participants' true maximal strength. The length of time (∼44 s) that the task was performed until the T1 data, and what occurred during the first 30 s (verbal feedback to help the participant maintain consistent torque output within the bandwidth) may have had a variable impact on T1 data, and as such, may also be considered as a limitation. Additionally, EMG has inherent limitations due to factors such as electrode placement, electrode movement, and individual anatomical differences. However, the same trained individual placed all electrodes in a secure fashion. As well, within-subject changes in EMG RMS were compared between groups, rather than one-time measures being compared between groups. Together, these steps reduce the impact of such factors on our measured outcomes. Group comparisons were also limited by the comparatively small sample sizes in each group. Finally, because of constraints related to athlete availability, the time since the last workout and the training phase of the season could not be controlled.

In conclusion, this was the first study to investigate fatigue-related changes in shoulder muscle activation among female groups with different aquatic sport specializations. When comparing the start, middle, and end of a repetitive internal rotation task, swimmers exhibited fewer changes in muscle activation compared to water polo players and controls, who appeared to use a strategy of initially increasing agonist activation and then reducing stabilizer activation to maintain performance. Patterns of EMG RMS in maximal shoulder rotations before and after fatigue was induced, as well as during the fatigue protocol, suggest that water polo players and swimmers may have had sports-specific ways of performing the same tasks. Whether one way was superior to another, or if they were simply different, remains to be determined with repeated testing and examining associations with injury development. As this is the first study to focus exclusively on female overhead athletes and their neuromuscular control, more research on these populations is needed to ultimately understand if there are sports-specific neuromuscular adaptations that are associated with injury development among female athletes, and in turn identify training and injury prevention strategies to promote healthy and continued sports participation among women.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

This study, which involved human participants, was reviewed and approved by the Centre de Recherche Interdisciplinaire en Réadaptation (CRIR-1247-0517). Written informed consent to participate in this study was provided by the participants' legal guardian/next of kin. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

AUTHOR CONTRIBUTIONS

SK was the lead on research design, setup, recruitment, data collection, collaborated on analysis, and writing. LD led the data analysis and writing with the assistance of the other authors. MC assisted during research study design, recruitment, data collection, and analysis. JC actively supervised and advised on every step and decision made regarding the research study and completion of the project. All authors contributed to the article and approved the submitted version.

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An Investigation of Bilateral Symmetry in Softball Pitchers According to Body Composition

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Introduction: High body fat percentage (bf%) is considered a potential injury risk factor for softball pitchers amidst the already high rates of pitching-related injury. Similarly, research points out that large bilateral asymmetries are another risk factor for softball pitchers. As softball pitching is a highly asymmetric sport and the repetitive nature of the windmill pitch places high stress on the body while pitchers are in unbalanced and asymmetric positions, research examining body composition and asymmetry is necessary.

Purpose: The purpose of this study was to compare functional characteristics of softball pitchers with a healthy and a high bf%. Bilateral symmetry was assessed for pitchers' hip and shoulder isometric strength (ISO) and range of motion (ROM) between the following two groups of softball pitchers: (1) those with a high bf% ($\geq$32%) and (2) those with a healthy bf% (<32%).

Methods: A total of 41 high school female softball pitchers from the southern United States agreed to participate (1.69 $\pm$ 0.07 m, 76.14 $\pm$ 17.08 kg, 15.1 $\pm$ 1.1 years). Pitchers completed a dual-energy X-ray absorptiometry (DEXA) scan and were grouped into one of the following two categories based on their bf%: healthy (<32 bf%) and high ($\geq$32 bf%). Bilateral symmetry was assessed for pitchers' hip and shoulder ISO and ROM using a handheld dynamometer and inclinometer, respectively. Bilateral arm bone and lean mass was also measured *via* the DEXA.

Results: Mixed analyses of variance revealed a significant interaction between bf% groups and side dominance for internal rotation shoulder ROM, $F_{(1,39)} = 14.383$, $p < 0.001$, $\eta^2_p = 0.269$. Main effects for side dominance were also observed for shoulder external rotation ISO, $F_{(1,39)} = 8.133$, $p = 0.007$, $\eta^2_p = 0.173$, hip internal rotation ISO, $F_{(1,39)} = 4.635$, $p = 0.038$, $\eta^2_p = 0.106$, arm bone mass, $F_{(1,39)} = 38.620$, $p < 0.001$, $\eta^2_p = 0.498$, and arm lean mass, $F_{(1,39)} = 101.869$, $p < 0.001$, $\eta^2_p = 0.723$.

Conclusion: Asymmetries and slight differences in functional characteristics exist between bf% groups. Altered functional characteristics may influence pitchers' windmill pitch movement and should be acknowledged by support staff to improve softball pitchers' health and longevity.

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Implications: Insight into asymmetries can help researchers and clinicians understand the implication of excess body fat and further theorize mechanisms of injury among this athlete population.

Keywords: asymmetry, softball, pitch, body fat percentage, range of motion, isometric strength

INTRODUCTION

Softball pitching is a highly asymmetric sport, with dominant and nondominant limbs performing drastically different motions (Fuchs et al., 2019). The regular asymmetry and vast amount of repetition present in softball pitching (Corben et al., 2015; Skillington et al., 2017) can lead to adaptations in physical and functional characteristics, as well as altered bilateral movement patterns (Friesen et al., 2019; Hellem et al., 2019). Research highlights the high demand of baseball and softball pitching and reports physical adaptations [e.g., glenohumeral internal rotation deficit (GIRD)] that ensue, due to the repetitious, irregular motions, and high amounts of force repeatedly stressing the body while in specific positions (Kettunen et al., 2000; Robb et al., 2010; Shanley et al., 2011; Li et al., 2015; Zeppieri Jr et al., 2015; Picha et al., 2016; Greenberg et al., 2017; Camp et al., 2018). While research consistently points out that the GIRD present among baseball throwers as a result of frequent positioning of the throwing shoulder into maximal external rotation (ER) and the high-velocity internal rotation (IR) of the shoulder joint (Garrison et al., 2012; Chou et al., 2018), softball pitchers experience similar motions whereby they would experience similar soft tissue and functional adaptation. During the acceleration phase of the softball pitch, pitchers experience high shoulder IR velocity (4,500 ± 1,200°/s), similar to their baseball counterparts (6,703 ± 5,770°/s) (Barrentine et al., 1998). Considering the specific nature of the underhand windmill pitch, understandably there will be adaptations that may vary accordingly. Research dedicated to specifically softball shows that pitchers exhibit greater adaptation in their stride leg than push leg due to the repetitive overloading during each aggressive landing during foot contact of the pitch (Fuchs et al., 2019). Evidence of bilateral differences among pitchers, in conjunction with the various roles of bilateral limbs, highlights the physical adaptations that ensue as a result of repetition.

Altered functional characteristics, such as strength and range of motion (ROM) of major joints, due to high repetition are also widely reported in throwing literature. While altered functional characteristics can bring about necessary adaptations to benefit the athlete, research also shows that some adaptations that evoke drastic asymmetry, such as large bilateral deficits in shoulder internal ROM, may lead to an increased risk of injury (Scher et al., 2010; Shanley et al., 2011; Saito et al., 2014; Tainaka et al., 2014; Bedi et al., 2016; VandenBerg et al., 2017). Therefore, while adaptations might be necessary from a performance perspective, large-scale adaptations noticeable *via* bilateral comparison might warrant caution for athlete safety. Understanding what might lead to greater bilateral deficits is important for ensuring player health and development.

A particular body trait that might pose further threat to asymmetry is body composition. Body fat percentage (bf%) and body mass index have been linked with altered hip and shoulder ROM (Kettunen et al., 2000; Friesen et al., 2020a). Similarly, recent research also suggests that bf% alters softball pitching kinematics and kinetics (Friesen et al., 2020b, 2022; Friesen and Oliver, 2021; Friesen K. et al., 2021). Given the higher rate of injury among those pitchers with greater mass (Oliver et al., 2019a), excess body fat among softball pitchers is a concern. Coincidentally, collegiate softball pitchers display the greatest amounts of bf% among their teammates and reports also show that, on average, they obtain more body fat throughout a competitive season (Czeck et al., 2019; Peart et al., 2019). Pitchers who possess more body fat tissue exhibit higher forces at injury-susceptible joints (Friesen, 2020; Friesen and Oliver, 2021), namely, the shoulder, and therefore, we would expect that there may be more significant joint adaptations for those pitchers with greater mass.

Recent reports show that those pitchers who are injured most often are typically heavier, taller, and have a higher body mass index or bf% (Greenberg et al., 2017; Oliver et al., 2018; Friesen K. B. et al., 2021). Therefore, amidst the already high rates of softball pitching-related injury (Oliver et al., 2019c; Valier et al., 2020), examination into the functional asymmetries of those with various bf% is necessary, especially given that research suggests dramatic bilateral asymmetries can predispose an athlete to greater risk of injury (Shanley et al., 2011). Therefore, the purpose of this study was to compare functional characteristics of softball pitchers with a healthy bf% and a high bf%. Bilateral symmetry was assessed for pitchers' hip and shoulder isometric strength (ISO) and ROM. Furthermore, side-to-side bone mass and lean mass symmetry was assessed for the arms between the two groups of pitchers. It is hypothesized that pitchers within the high bf% group may accrue greater asymmetries in joint ROM and ISO. It was also hypothesized that pitcher groups would display different mean values of bone and lean tissue in their arms.

MATERIALS AND METHODS

Participants

A total of 41 high school female softball pitchers from the southern United States agreed to participate in the study (1.69 ± 0.07 m, 76.14 ± 17.08 kg, 15.1 ± 1.1 years, $n = 41$, $n = 35$ right-hand dominant). Pitchers were grouped into one of two categories based on their total bf%. The cutoff value determining pitchers with high fat% from pitchers with healthy fat% was set at 32% body fat according to the American College of Sports Medicine criterion-based reports, which define 20–32% body

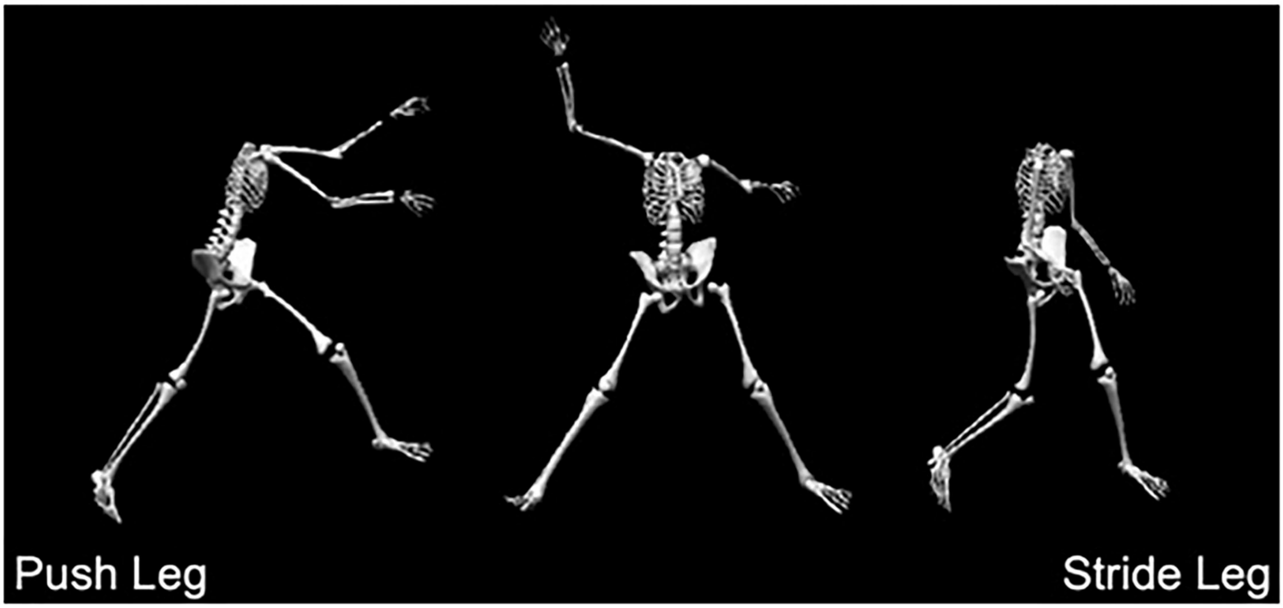

FIGURE 1 | Push and stride leg illustration during the windmill pitch.

fat as being satisfactory for health in women (ACSM, 2013). Therefore, pitchers were grouped in the healthy bf% group if their bf% was $< 32\%$ and grouped into the high bf% group if their bf% was $\geq 32\%$. There were 18 pitchers grouped into the healthy bf% group (1.70 $\pm$ 0.07 m, 64.54 $\pm$ 9.11 kg, 16 $\pm$ 2 years) and 27 pitchers grouped into the high bf% group (1.70 $\pm$ 0.07 m, 84.04 $\pm$ 15.78 kg, 15 $\pm$ 2 years).

Prior to participation, all participants were explained the study protocol and informed consent was signed by the participants' parent/guardian while the participant signed assent documentation. To be eligible for participation, pitchers needed to be injury- and surgery-free for the past 6 months and on a current softball roster. They also needed to have pitched in a game within the past 6 months and have reported to the laboratory fully rested for the previous 24 h.

Procedures

All protocols were approved by the Institutional Review Board. Pitchers first completed a dual-energy X-ray absorptiometry (DEXA) whole-body scan, which collected whole-body and segmental composition measurements, including fat tissue, lean tissue, and bone mineral content (GE Healthcare, Madison, WI, USA). The standard error of estimate for the DEXA is $\pm$1.8%. Following the dual-energy X-ray measurement, pitchers' bilateral hip and shoulder ISO and ROM were assessed in both the IR and ER directions. Regarding the hips, the push hip was part of the leg that pushes the pitcher off the ground and the stride hip referred to the leg that made foot contact during the pitch (**Figure 1**).

A handheld dynamometer (Lafayette Instruments, Lafayette, IN, USA) and inclinometer (Fabrication Enterprises, Inc., White Plains, NY, USA) were used to measure ISO and ROM, respectively (Dwelly et al., 2009; Sauers et al., 2014; Oliver et al., 2016, 2019b; Friesen et al., 2019, 2020a). For hip data measurement, pitchers sat on an athletic training table with their hips and knees flexed at 90°. A rolled towel was placed under the distal femur to allow for smooth rotation of the hip joint (see **Figure 2**). ROM was assessed by having the examiner rotate the shank either toward the contralateral leg (ER measurement) or away from the contralateral leg (IR measurement). The inclinometer was placed on the shaft of the fibula just proximal to the lateral malleolus for IR and on the shaft of the tibia just proximal to the medial malleolus for ER (Friesen et al., 2019). End ROM was determined just prior to when the participants' hip would lift off of the table while seated and with the examiner feeling for firm capsular end feel (Friesen et al., 2019). Hip ISO was also measured in this position (**Figure 3**). The dynamometer was placed at the same location as the inclinometer and resistance was applied while the participant pushed against the dynamometer and the hip remained in a neutral position during testing.

In gathering shoulder measurements, participants lay supine on an athletic training table with their upper arm abducted 90° and their elbow flexed to 90°. Again, a rolled towel was placed under the distal humerus to ensure smooth shoulder movement. The tester used one hand to limit scapular movement while the other hand slowly rotated the forearm either in the direction of the feet (IR measurement) or in the direction of the head (ER measurement). End ROM was determined just prior to when the participants' scapula would lift off the table for IR and at firm capsular end feel for ER (**Figure 4**). The inclinometer was positioned on the forearm above the styloid

FIGURE 2 | Hip ER ROM measurement.

process of the ulna for both measurements. ISO was measured in the same way, with the tester maintaining a neutral arm position for the participant, while they applied pressure to the dynamometer (**Figure 5**). The tester ensured intrarater reliability, with an intraclass correlation coefficient [ICC(3,k)] of 0.92–0.95 for all measurements. Minimal detectable change (MDC) was calculated with a 95% CI to determine clinical significance. Glenohumeral joint IR and ER ROM MDCs were 6.2 and 7.5, respectively. Hip IR and ER ROM MDCs were 6.6 and 4.9, respectively. IR and ER MDCs for ISO measurements were 3.5 and 3.6, respectively, for the glenohumeral joint, and 9.0 and 2.3 for hip IR and ER.

Statistical Analyses

All statistical analyses were completed in SPSS software package (SPSS Statistics 26 Software, IBM Corp., Armonk, NY, USA). Prior to statistical analyses, all data were checked for normality, linearity, and outliers (defined as those >2 SDs away from the mean). Data were considered normal and linear, with a few outliers present. Analyses were conducted with and without outliers to which no significant differences were observed; therefore, original data with outliers included were used for analyses. Levene's test for equality of variance was conducted and equal variance was consistently observed. Mixed analyses of variance (ANOVA) were conducted to assess the difference

FIGURE 3 | Hip ER ISO measurement.

between bf% groups and side dominance on several variables, including shoulder and hip external and IR ROM and ISO. Two subsequent mixed ANOVAs were also used to assess arm bone mass and lean mass. Alpha level was set *a priori* to 0.05.

RESULTS

Means and SDs for each variable are presented in **Table 1**. The ANOVA assessing IR shoulder ROM revealed a statistically significant interaction between bf% groups and side dominance, $F_{(1,39)} = 14.383$, $p < 0.001$, $\eta^2{}_p = 0.269$, with an observed power of 0.959. The high bf% group had more dominant shoulder IR ROM than the healthy bf% group (mean difference = 4°) (**Table 1**). Examination of the healthy bf% group shows that the dominant shoulder displays less IR ROM than the nondominant shoulder (mean difference = 4°), while the high bf% group reveals more IR ROM in their dominant than nondominant shoulder (mean difference = 2°) (**Table 1**). It is noted that none of these differences are of clinical significance as they are less than the MDC previously calculated.

The ANOVA assessing shoulder ER ISO was significant and revealed a main effect for side, $F_{(1,39)} = 8.133$, $p = 0.007$, $\eta^2{}_p = 0.173$, with an observed power of 0.794. The

FIGURE 4 | Shoulder IR ROM measurement.

dominant/throwing shoulder displayed significantly more ER ISO than the nondominant/glove side (mean difference = 10 kgf). There was also a main effect for side in the ANOVA assessing hip IR ISO, $F_{(1,39)} = 4.635$, $p = 0.038$, $\eta^2_p = 0.106$, with an observed power of 0.556. The dominant/push hip displayed significantly more IR ISO than the nondominant/stride side (mean difference = 10 kgf). No other assessments regarding ROM and ISO were statistically significant.

Examination of bone and lean tissue of the dominant and nondominant arms revealed significant main effects for side dominance. The ANOVA examining bone reported $F_{(1,39)} = 38.620$, $p < 0.001$, $\eta^2_p = 0.498$, with an observed power >0.999. The ANOVA examining lean tissue reported $F_{(1,39)} = 101.869$, $p < 0.001$, $\eta^2_p = 0.723$, with an observed power >0.999. Both bone (mean difference = 0.5 lbs) and lean tissue (mean difference = 0.6 lbs) were heavier in the dominant arm of the pitchers.

DISCUSSION

The constant asymmetric motions associated with softball pitching and the additional risk for injury associated with body asymmetries (Shanley et al., 2011) warrant investigation to the extent of functional and bodily side-to-side differences among softball pitchers. Furthermore, the added risk of injury for those pitchers who possess more body fat tissue, and the increased injury rates among those who do have

FIGURE 5 | Shoulder ER ISO measurement.

excess body fat, emphasizes the need to understand how functional characteristics differ between those with a healthy and high bf%. The main findings of this article show that (1) the side-to-side differences in IR shoulder ROM vary differently between pitcher bf% groups; (2) the dominant side shoulder exhibits greater ER ISO than the nondominant shoulder; (3) the push hip exhibits greater IR ISO than the stride hip; and (4) the dominant side arms display heavier bone and lean tissue. These findings provide insight into the asymmetries of high school softball pitchers and the slight difference in functional characteristics that exist between bf% groups.

Body Fat Percentage Group Differences

The only functional characteristic to present differences according to bf% group was IR ROM of the shoulder, although it is important to note that these differences were smaller than the MDC calculated previously. Discussion of results should therefore be viewed with this in mind. Data revealed that pitchers with a healthy bf% had less dominant arm IR ROM than the high bf% group (albeit not of clinical significance). The difference between bf% groups suggests that the amount of body fat a pitcher possesses might have a slight influence on functional characteristic adaptations. Loss of IR ROM of the dominant shoulder is common among those athletes who regularly perform

TABLE 1 | Means ± standard deviations for both pitcher groups' hip and shoulder ISO, ROM, and arm lean tissue and bone.

Variable	Healthy-fat%		High-fat%	
	Dominant	Non-dominant	Dominant	Non-dominant
Hip IR ROM (°)	30.4 ± 4.0	30.1 ± 3.4	27.3 ± 5.3	28.7 ± 4.6
Shoulder IR ROM*	40.7 ± 5.1[a,b]	44.5 ± 4.2[b]	44.8 ± 5.8[a,c]	42.4 ± 6.8[c]
Hip ER ROM (°)	38.0 ± 4.0	38.1 ± 3.8	36.9 ± 4.6	36.3 ± 3.3
Shoulder ER ROM (°)	100.6 ± 14.5	100.7 ± 15.1	94.7 ± 12.2	94.5 ± 17.0
Hip IR ISO†	178.2 ± 35.1	170.0 ± 42.8	188.3 ± 42.2	176.1 ± 43.4
Shoulder IR ISO (kgf)	155.7 ± 18.3	155.9 ± 17.8	159.1 ± 24.2	154.9 ± 33.4
Hip ER ISO (kgf)	136.5 ± 26.6	135.3 ± 23.6	138.7 ± 30.4	146.5 ± 33.9
Shoulder ER ISO†	168.0 ± 26.0	163.7 ± 23.8	186.1 ± 43.0	171.0 ± 47.0
Arm Lean Tissue (lbs)†	5.6 ± 1.0	5.0 ± 1.0	6.3 ± 1.0	5.7 ± 1.1
Arm Bone Tissue (lbs)†	0.4 ± 0.1	0.3 ± 0.1	0.4 ± 0.1	0.4 ± 0.1

*Denotes significant interaction (within a row, same letters denote significant differences).
†Denotes significant main effect for side.

throwing tasks (Shanley et al., 2011), but the healthy bf% group displaying less throwing shoulder IR ROM than the high fat% was not expected. Originally, it was hypothesized that the high fat% group might display decreased ROM in general, due to the impingement issues that arise from exhibiting increased fatty tissue. Research points out that those pitchers with a higher body mass index possess less bilateral hip ROM, hypothesized to be the result of greater impinging tissue (Friesen et al., 2020a). It was also expected that the greater mass associated with those in the high bf% group might accrue greater adaptation due to increased loading on the joints. With current data revealing the opposite, it might suggest that the pitchers with a healthy bf% might be accruing greater functional adaptations than the high fat% group. Perhaps healthy fat% pitchers encounter higher repetition and greater pitch counts that might result in greater adaptation (e.g., more throwing shoulder ER ROM and less IR ROM). Conversely, this might suggest that the high fat% group performs less repetitions, which could be either, or both, a cause and effect of carrying extra fat mass. Those who possess extra fat mass might be less active and, as a result, perform fewer and less intense repetitions. Interestingly, a previous report also found that pitchers with higher body mass index did not achieve as much dominant shoulder ER ROM as those with lower body mass index (Friesen et al., 2020a). Therefore, adaptation maybe does not occur the same in a population of pitchers with increased mass and body fat. Future studies are necessary to understand why those softball pitchers with more body fat accrue less throwing shoulder IR ROM as the discussion of causation is merely speculation. It should also be noted that since the acquisition of less IR ROM is a common adaptation of throwers, and our two groups displayed opposite trends of adaptation, future research should account for body mass or bf% as a potential covariate when analyzing functional adaptations of throwers.

Side-To-Side Differences

Data revealed that there were side-to-side asymmetries in IR ROM between the throwing and glove shoulders. Interestingly, the high bf% group displayed more IR ROM in their throwing shoulder compared to the glove shoulder, opposite of the healthy fat% group. While drastic shoulder asymmetries can lead to injury among throwers (Shanley et al., 2011), the varied asymmetry between bf% groups suggests a potential effect of body fat on shoulder adaptations. However, as asymmetry was present within both populations, we cannot assume that body fat alters symmetry. We can, however, state that body fat might influence functional adaptations, namely, at the shoulder joint. As we know that higher bf% has been demonstrated to be related to higher injury risk (Chalmers et al., 2015; Oliver et al., 2018), this also provides some evidence that perhaps the altered functional characteristics within those with higher body fat could be a potential mechanism for injury; however, more studies are needed to examine mechanisms of injury among those with higher body fat.

Notable side-to-side differences were observed in shoulder ER ISO and hip IR ISO. Based on the pitching motion, asymmetry in IR ISO of the shoulder was expected as another typical adaptation for throwers. During the windmill pitch, the throwing arm shoulder completes rapid IR during the acceleration phase of the pitch (Maffet et al., 1997; Barrentine et al., 1998). To slow the rapid IR, ER strength is necessary to bring the pitcher to a safe stop to finish the pitch. The repeated pitching exposure can strengthen the musculature of the throwing shoulder while in this position, as opposed to the glove arm that performs less overall motion. It was also reported that there was more IR strength in the push hip than in the stride hip. Again, this was expected as the drive off the ground occurs while the pitchers' push hip is internally rotated (Friesen et al., 2020a). In contrast, during the stride, the stride hip needs ER strength more than IR strength to keep the stride leg hip from collapsing and exhibiting greater IR upon aggressive ground contact.

The final findings noted asymmetry in throwing arm lean tissue and bone for both groups of pitchers. As might be expected, bone and lean tissue was heavier within the throwing arm. As this did not vary between pitcher bf% groups, this again highlights the importance of other factors that might influence adaptation to functional characteristics. However, descriptive analysis of the mean and SD data reported in **Table 1** shows that the high fat% pitchers did possess greater mean weight of both lean tissue and bone than the healthy fat% group. Perhaps this is an adaptation required to move the generally larger limb of those who are bigger in stature and carry increased body fat.

While research notes that pitchers with increased bf% exhibit higher rates of pitching-related injury (Oliver et al., 2019a; Friesen K. B. et al., 2021) and altered biomechanics during the windmill pitch (Friesen et al., 2020b), there were minimal differences in functional characteristics between bf% groups. There were, however, many asymmetries reported, as could be expected. As bf% does not seem to greatly influence player adaptation, but injury reports suggest an increased injury-susceptibility for larger athletes, additional research is needed to understand the mechanism of injury among softball pitchers. Similarly, more studies are needed to examine the influence of specifically fat mass on softball pitch biomechanics in effort to best inform practitioners working with the development of softball pitchers. Also, this might suggest that other factors such as biomechanics, effort, and practice/game exposure influence functional characteristic adaptations more than the presence of body fat.

Limitations

This study has some limitations. First, there are potential inconsistencies in maximum effort required during the ISO measurement, and similarly the consistency of the applied rater pressure to the device can vary given the handheld nature of the device. However, all participants were given the same instruction, "to push as hard as possible," and the rater too gave maximal effort during each measurement. Second, the expertise and development of participants was not controlled; therefore, pitchers with various mechanics and pitching success rates might influence the aggregate data analysis. However, the recruitment sample did come from one general area of the southern United States; therefore, it might be assumed that these players were of a similar skill and competition level. Similarly, the study sample involved a relatively small age range, and it might be expected that older pitchers with more substantial bf% differences could present different and more substantial findings. It is also important to keep in mind that there were no drastic functional characteristic differences between the bf% groups, nor was there a significant number of other variables that differed between bf% groups. Finally, our functional characteristics were measured in static positions and will therefore be limited in their application of the dynamic pitching motion under consideration.

DATA AVAILABILITY STATEMENT

Data inquiries can be directed to the corresponding author.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Auburn University Institutional Review Board. Written informed consent to participate in this study was provided by the participants' legal guardian/next of kin.

AUTHOR CONTRIBUTIONS

KF and GO: study conception and design and data collection. KF, AL, KC, and GO: data analysis and interpretation, drafting manuscript, and critical revision. All authors contributed to the article and approved the submitted version.

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