POSTURAL BALANCE CONTROL IN SPORT AND EXERCISE

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POSTURAL BALANCE CONTROL IN SPORT AND EXERCISE

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EDITORIAL
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Editorial: Postural Balance Control in Sport and Exercise

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Editorial on the Research Topic

Postural Balance Control in Sport and Exercise

Postural balance control enables maintenance, achievement, or restoration of a state of balance in any posture or activity. It is, therefore, a determinant in achieving goals in daily life, increasing the quality of life, and maximizing performance in recreational and professional sports. Indeed, in a sports context, balance is one of the performance-limiting factors, and no sport-technique gesture can be efficiently achieved without an effective postural balance control. In sports activities, balance control can also be associated with injury risk. Postural balance control is governed by automatic processes in which the individual is unaware of the adjustment of postural muscle tone and by cognitive processes in which the information about the individual's own body and the localization of objects in extrapersonal space are required. The complexity of these mechanisms makes postural balance control assessment with a rational approach challenging. In sports and exercise, the assessment of static postural balance provides objective data based on conventional techniques (e.g., measurements of centre of pressure trajectory). However, among athletes or healthy active people, this condition may not be adequately challenging. Conversely, functional and ecological tests provide information on postural ability but may lack objectivity in scoring. Moreover, the assessment of postural balance can be influenced by the postural conditions adopted. For instance, balance performance can change when assessed in a sport-specific context rather than under context-independent postural conditions. Hence, the study of the influence of postural balance control on sports performance or injury risk should necessarily include assessments under dynamic and environmental conditions besides assessments under static and decontextualized conditions. Taking together these considerations, the present Research Topic explored the role of static and dynamic postural balance control on sport and exercise performance, focusing on both static and dynamic assessment methods.

In this Research Topic, three studies addressed the effects of fatigue on postural balance performance. Kozinc et al. investigated the effects of whole-body fatigue on postural sway in single-leg stance and its transient characteristics. They showed that the fatiguing protocol did not increase postural sway, although it tended to be more variable across the trial. Interestingly, postural sway was less affected in females than in males. Marcolin et al. aimed to investigate in male adults whether a fatigue protocol on calf muscle could affect muscle activation strategies and dynamic balance performance over an oscillating platform. They found that the reduction of the electromyographical activity of the soleus during the dynamic balance task after fatigue did not affect the global dynamic postural balance performance, likely due to an overall increase in the calf muscles stiffness. Camargo da Silva et al. extended knowledge of the mechanisms underlying the interplay between fatigue and postural performance associated with the neuromuscular adaptations induced by sports practice. Specifically, they compared gymnasts and age-matched non-gymnasts and showed that fatigue significantly increased medio-lateral postural oscillations in all subjects.

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Moreover, gymnasts did not present a better postural balance control than non-gymnasts. Based on electromyographic parameters, the former employed other neuromuscular control strategies to maintain their postures in single-legged quiet standing.

Gorjan et al. investigated the motion of the centre of mass while standing with external stabilization, without constraining the joint movements, using a method based on an inverted cart-pendulum system. Results showed inter-individual variability in postural responses during the external stabilization, with less than half of all subjects showing a stabilizing effect.

Kacem et al. examined the effect of neuromuscular fatigue on static and dynamic postural balance performance in female athletes during the premenstrual phase and the menstrual cycle. Results indicated that the disruptive effect of neuromuscular fatigue on static and dynamic postural control was more accentuated in the premenstrual phase than during the menstrual cycle.

Pojskic et al. investigated the influence of acute hypoxic exposure on balance ability in highly trained basketball players through repeated single-leg balance tests over a multiaxial tilting platform. The findings of this study highlighted how the acute effects of normobaric hypoxia on balance performance were modulated by the playing level. Indeed, elite players had better resistance to the adverse effects of normobaric hypoxia than their sub-elite counterparts.

Deng et al. showed the effect of hip joint angle on quadriceps recruitment pattern and stiffness in healthy individuals during isometric knee extensions.

Heredia-Elvar et al. focused on some of the most popular isometric core stability exercises. They provided new observational screening guidelines based on body alignment, postural sway, and pelvic acceleration thresholds calculated using a smartphone and its inbuilt accelerometer. These guidelines could help decide if a core stability exercise variation represents an adequate training intensity level for a given participant.

In a first review article, Zemkova noted that of the various studies that examined the effects of exercise on postural balance control, only a few were conducted under actual sport-specific conditions. Addressing this gap should have positive implications for developing specific exercise programs in those disciplines where success is linked to postural performance. In a second review article, Zemkova and Zapletalova highlighted that of the numerous studies that investigated the role of neuromuscular control of postural and core stability in sports performance, only a few demonstrated a relationships between body balance and stability of the core.

Ketterer et al. investigated whether continuous sensory conflicts induced by optic flow perturbations induced by virtual reality could challenge the postural system sustainably. The study results showed that sinusoidal optic flow perturbations reduced postural stability only in the first 5 s of the test, thus being not sufficiently suitable for balance training as they cannot trigger persisting sensory conflicts.

Kiers et al. presented a dynamic postural stability index based on the forces recorded during a ski-specific single-leg landing on a dynamometric platform after a forward double-leg drop jump from a box over a hurdle. The results suggested that this index could be a reliable and sensitive measure of dynamic postural control in youth skiers.

In conclusion, the Research Topic points to the need for further investigation in sports and daily-living contexts to understand the role of specific environmental conditions on postural balance performance. Given the specificity of balance concerning the external context, researchers are encouraged to develop new tests in addition to standardized laboratory tests. An interesting perspective is the employement of wearable technologies to enable the development of context-specific tests and overcome the lack of objectivity in the assessment of current ecological tests. An open Research Topic is how to effectively quantify the intensity of destabilizing exercises. Expanding this knowledge will support the development of tailored balance training programs to enhance athletic performance and reduce the risk of falls in daily life.

AUTHOR CONTRIBUTIONS

All authors listed have made a substantial, direct, and intellectual contribution to the work and approved it for publication.

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ORIGINAL RESEARCH
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The Effect of Fatigue on Single-Leg Postural Sway and Its Transient Characteristics in Healthy Young Adults

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Neuromuscular fatigue is known to impair balance ability, which is reflected in increased postural sway during quiet standing tasks. Recently, quantifying transient characteristics of postural sway has been suggested as an approach to obtain additional information regarding postural control. However, this approach is currently vastly unexplored. The purpose of this study was to investigate the effects of fatigue (induced by a repeated change of direction task) on postural sway and its transient characteristics during single-leg standing, including whole-trial estimates and indexes of transient behavior in young healthy active adults. The study involved 28 physically active students (14 females). Single-leg postural sway was recorded for 30 s before and after a fatiguing protocol, which consisted of a repeated change of direction tasks. We calculated the traditional whole-trial estimates of postural sway [center-of-pressure (CoP) velocity and amplitude in anterior-posterior (AP) and medial-lateral (ML) directions] and corresponding transient behavior indexes, based on three 10-s intervals. Statistically significant sex × fatigue interaction with medium effect sizes was found for whole-trial CoP velocity in AP ($p = 0.028$; $\eta^2 = 0.17$) and ML directions ($p = 0.019$; $\eta^2 = 0.19$). *Post-hoc* test showed that both variables substantially decreased in female participants ($p = 0.041$–0.045; $d = 0.54$–0.56), but remained similar in males ($p = 0.194$–0.294). There were small to medium statistically significant main effects of fatigue on transient index for CoP amplitude in both directions ($p = 0.042$–0.049; $\eta^2 = 0.02$–0.14). Notably, CoP AP amplitude increased in the first 10-s interval for males (before fatigue: 5.6 ± 1.3 mm; after fatigue: 6.3 ± 1.6 mm), while the CoP AP amplitude in the third interval remained similar after fatigue (before fatigue: 5.5 ± 1.4 mm; after fatigue: 5.1 ± 1.2 mm). In conclusion, the responses to fatigue in terms of postural sway were time interval specific, and there were certain sex-differences in responses to fatigue, which could be related to better ability to adapt balance strategies in females. Moreover, our results demonstrate that the indexes of transient behavior could perhaps detect smaller fatigue-induced changes in postural sway that are seen in whole-trial estimates.

Keywords: body sway, posture, stability, transient behavior, fatiguing

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INTRODUCTION

Balance and postural control are fundamental abilities underpinning normal human movement function (Pollock et al., 2000; Ivanenko and Gurfinkel, 2018). Assessments of balance are routinely performed in research and clinical practice, studying patients with neuromuscular impairments (Gunn et al., 2015; Shen et al., 2016), children (Geldhof et al., 2006; Alsakhawi and Elshafey, 2019), and older adults (Park, 2018; Kozinc et al., 2020). In athletes, dynamic balance tests have been used to assess injury risk (De Noronha et al., 2006; Plisky et al., 2006); however, the utility of static balance tests (e.g., assessment of postural sway during quiet standing) for athletic populations has not been explored as much. In a literature review, Hrysomallis (2011) reported that static and dynamic balance are related to athletic performance and that balance exercise can contribute to improvements in the performance of movement tasks such as jumping and agility tests. Clearly, the role of balance assessment should not be neglected in athletes, both from injury prevention and performance enhancement aspects. It has been postulated that neuromuscular fatigue, owing to its effects on movement biomechanics, plays an important role in injury risk in sport (Barber-Westin and Noyes, 2017; Benjaminse et al., 2019; Bourne et al., 2019). Therefore, it could be of particular interest to understand how neuromuscular fatigue affects balance.

Static balance is traditionally assessed by measuring postural sway during quiet stance, most typically through center-of-pressure (CoP) movement recordings. The negative effect of neuromuscular fatigue on balance is very well-known. These effects may be driven by various mechanisms, such as altered position sense (Forestier et al., 2002), changes in visual, vestibular, and proprioceptive sensory inputs (Lepers et al., 1997), and reductions in muscle output (Enoka and Stuart, 1992); for review, see Paillard (2012). Increases of postural sway during parallel and single-leg stance have been documented after locally fatiguing lower limb (Lin et al., 2009; Bisson et al., 2011; Zech et al., 2012; Boyas et al., 2013; Barbieri et al., 2019) and after fatiguing the trunk muscles (Pline et al., 2006; Lin et al., 2009; Ghamkhar and Kahlaee, 2019). Similarly, postural sway during single-leg stance is increased after global fatiguing protocols, based on running (Zech et al., 2012; Steib et al., 2013; Beurskens et al., 2016), rowing (Springer and Pincivero, 2009), load lifting (Bannon et al., 2018), and repeated sprinting (Pau et al., 2014). In addition, isolated fatiguing of inspiratory muscles was also reported to result in increased postural sway during parallel stance (Janssens et al., 2010). Since sports training and competitions mostly include full-body movement tasks, it seems reasonable to assume that investigating the effect of global fatigue on balance has greater ecological validity and practical relevance for athletes. In the study performed by Zech et al. (2012), the global fatigue had a significantly larger effect on single-leg postural sway (+47%) than local fatigue (+10%), whereas similar impairments in dynamic balance (assessed through star excursion balance test) were observed after local and global fatigue in another study (Garcia-Gallart and Encarnacion-Martinez, 2019). Greig and Walker-Johnson (2007) investigated the effect of a 90-min running protocol on single-leg postural sway in male soccer players and reported that postural sway was not meaningfully affected. On the other hand, an increase in single-leg postural sway was produced by a zig-zag task in male and female athletes from field-based team sports (Lacey and Donne, 2019). Interestingly, aerobic fatigue had a more pronounced effect on single-leg postural sway than anaerobic fatigue in a study conducted on female soccer players (Güler et al., 2020). Recently, Bedo et al. (2020) used a complex fatiguing protocol, consisting of multiple tasks, such as jumping, sidestep cutting, sprinting, and landing. This protocol resulted in a large increase in single-leg postural sway, along with a decrement in maximal strength. Both abilities, however, returned to baseline within 5 min (Bedo et al., 2020). On the contrary, Baghbani et al. (2016) observed no effect of a complex seven-station fatigue protocol on dynamic balance (star excursion test in athletes), while the non-athlete control group did perform worse after fatiguing as expected.

The evidence regarding the differences between the sexes in terms of postural sway are somewhat ambiguous, as previous experiments showed either no differences between males and females (Cruz-Gómez et al., 2011; Olchowik et al., 2015) or that females exhibit lower postural sway (Greve et al., 2013; Sell et al., 2018). Caution is needed when examining sex differences, as anthropometric variables may cofound the results (Era et al., 1996; Bryant et al., 2005; Villarrasa-Sapiña et al., 2016; Plandowska et al., 2019). Studies have also suggested that postural sway in males is more sensitive to fatigue (Bannon et al., 2018) and vision elimination (Masui et al., 2005). Therefore, when examining the effects of fatigue on postural sway, sex, and anthropometric variables should be considered as factors in statistical analyses.

Having in mind that balance has been linked to sports performance (Hrysomallis, 2011) and injury risk (Hrysomallis, 2007); the effect of fatigue-induced balance deterioration on injury risk and sports performance is unknown. Recently, a new approach to analyze transient behavior of CoP data has been proposed by Reed et al. (2020). Instead of averaging the CoP data across the trial, this approach separates the data into several time intervals and evaluates the changes in CoP behavior among these intervals. Reed et al. (2020) reported that indexes of the transient behavior of postural sway were independent of the whole-trial estimates (in other words, they were not correlated with the whole-trial estimates), and were sensitive to age and vision elimination. In our subsequent studies, we confirmed the independence of transient indexes and the whole-trial variables, and demonstrated their sensitivity (albeit with small effect sizes) to leg preference (Kozinc and Šarabon, 2021a). Moreover, we found different transient behavior (i.e., quicker reduction of postural sway throughout the trial) in ballet dancers compared to young adults (Kozinc and Šarabon, 2021b). Admittedly, averaging the data is superior in terms of the reliability of the outcomes, however, indexes of transient behavior provide additional information regarding an individual's balance control that are masked (e.g., initial less stable period) by averaging the whole-trial data. This approach could be relevant for injury risk screening in sport. However, we found no differences in terms of the transient behavior of postural sway in single-leg stance between athletes with and without ankle sprain history (Kozinc et al., 2021). In this study, we investigated whether indexes of transient behavior of single-leg postural sway could be used to detect deterioration in balance, induced by fatigue. Since fatigue is a major determinant of injury risk in athletes (Benjaminse et al., 2019; Bourne et al., 2019), sensitive and valid approaches to monitor it, are needed. Indexes of transient behavior could be particularly useful if they showed higher sensitivity to fatigue in comparison to whole-trial estimates postural sway. Therefore, the purpose of this study was to investigate the effects of whole-body fatigue on postural sway during single-leg standing, including whole-trial estimates and indexes of transient behavior in young healthy active adults. Based on the large body of literature on the effects of fatigue on postural sway, we considered the single-leg stance as the most appropriate task, being more challenging than bipedal tasks and consequently being expected to be more sensitive to fatigue. We hypothesized that the fatiguing protocol will result in increased whole-trial CoP amplitude and velocity. The lack of previous studies precluded a formation of hypothesis regarding the effects of fatigue on indexes of transient behavior. The secondary aim of this study was to examine if the effects of fatigue on postural sway are sex-specific.

MATERIALS AND METHODS

Participants

Based on the previous studies (Zech et al., 2012; Steib et al., 2013; Pau et al., 2014; Lacey and Donne, 2019; Güler et al., 2020), we expected to observe an increase in CoP velocity and amplitude with an effect size (Cohen's *d*) of ~0.75–1.00. We calculated that 19 participants would be needed to confirm such an effect with 90% statistical power and alpha level of 0.05. However, since the fatiguing protocols as well as postural tasks were different in previous studies, we aimed for a higher sample size to detect potential smaller effect. Thus, the study sample comprised of 28 kinesiology students (14 females, age: 22.1 ± 2.1 years; body mass: 60.5 ± 5.7 kg; body height: 166.9 ± 8.4 cm; body mass index: 21.8 ± 2.4; and 14 males, age: 26.0 ± 5.1 years; body mass: 76.6 ± 6.7; body height: 181.5 ± 4.6 cm; body mass index: 24.2 ± 2.1). The participants were required to be free from musculoskeletal injuries in the previous 6 months, pain and other medical conditions that could affect postural control. All participants reported being physically active at least three times a week, with different preferred exercises of choice (e.g., resistance training and aerobic exercise). They also completed several study courses that required them to exhibit decent levels of aerobic endurance, strength, flexibility, and balance. For instance, they were required to reach the norms for 2,400 m run test (males: <10 min 30 s; females: <11 min 30 s), hamstring flexibility [reaching to the floor with fingertips (males) or full palms (females) from a standing position, without flexing the knees], and full body strength (lifting 50% of their body weight three times with an overhead squat exercise technique), and correctly perform several drills or movement skills (swimming techniques, basic elements of soccer, volleyball, basketball and handball, and athletic drills). The procedures of this study were approved by National Medical Ethics Committee (approval number: 0120-690/2017/8) and were conducted in accordance to the Declaration of Helsinki. After the explanation of the experimental procedures, a written informed consent was obtained from all participants.

Study Design and Experimental Tasks

The study was conducted in a single session, lasting approximately 1 h. Before the measurements, the participants performed a 15-min warm up, consisting of 10 min of light-intensity running on an indoor track and 5 min of dynamic stretching exercises. Postural sway was assessed during single-leg quiet stance before and after the fatiguing protocol (see below), using a force plate positioned on a parquet floor. Before the main trials, a standardized 20 s familiarization trial was performed to introduce the task requirements to the participants. For the main trials, the participants were required to maintain a single-leg stance for 30 s. The knee of the stance leg was in an extended position, but it was not allowed to completely lock (i.e., hyperextend). The hip of the other leg was in a neutral position (0°), whereas the knee was flexed at 90° and was not allowed to be touching the standing leg. Participants looked at a fixed point at an eye level and ~3 m away from them. The hands were placed on the hips throughout the trials. A 30-s repetition was performed before and after the fatigue. The participants assumed the single-leg position, and the examiner started the acquisition after ~1 s. After the fatiguing protocol was finished, the postural sway assessments were repeated immediately. We analyzed the postural sway on the preferred leg, which was determined as the leg that the participant would use to kick a ball. It has been shown that postural sway is slightly smaller during standing on a preferred leg compared to the non-preferred leg (Kozinc and Šarabon, 2021a).

Fatiguing Protocol

The fatiguing protocol consisted of two repeated change-of-direction (CoD) tasks. Both tasks encompassed a 5 + 5 m run with either a 90 or 180° CoD maneuvers in between, and were performed in both left and right directions. These procedures are producing reliable outcomes (ICC = 0.84–0.96; inter-trial variation <2%) in our laboratory (Šarabon et al., 2020). The protocol was performed in a gym (parquet floor). The time to complete each repetition was measured with photocell timing gates (Brower Timing Systems, Draper, UT, United States). The gates were placed at about hip height, with the start line being 0.5 m behind the first timing gate to avoid early triggering. Before the fatiguing protocol, the participants first completed two familiarization trials for each task (two trials for left and right directions each) at 50 and 75% of their maximal effort. Then, they performed three repetitions with the maximal effort for each task to establish baseline times. The fatiguing protocol involved performing the CoD tasks in an alternating order (e.g., 90° – left, 90° – right, 180° – left, and 180° – right), with 5 s breaks in between. The protocol was stopped when the time to complete the task increased by 20% or more in the 180° turn task for the dominant direction (the direction for which shorter time was recorded at baseline). Right after the protocol stopped, the participants were asked to report their rate of perceived exertion using a 0–10 scale as suggested by Foster et al. (2001).

Data Processing and Outcome Measures

The ground reaction force data were collected (sampling frequency: 1,000 Hz) by a force platform (model 9260AA; Kistler, Winterthur, Switzerland) and were low-pass filtered (Butterworth, second order, 10 Hz), using the manufacturer's MARS Software (Kistler, Winterthur, Switzerland). The data were then processed in the same software in order to obtain CoP velocity [total, anterior-posterior (AP) and medial-lateral (ML)], CoP amplitude (AP and ML), and CoP frequency. The CoP amplitude was defined as the mean amount of the CoP sway in AP or ML direction, calculated as the common length the COP sway trajectory only in the given direction, divided by the number of changes of movement direction. Firstly, we calculated the traditional whole-trial estimates (i.e., averaged CoP characteristics over the whole 30 s of the trial). In addition, we also calculated the relative differences between the first and the second (DIF_21) and first and third (DIF_31) 10-s time intervals within the whole trial. For each interval, that data were locally demeaned. The relative differences between the intervals were expressed as percentages (100% representing no change; >100% indicating an increase in time; and <100% indicating a decrease in time).

Statistical Analysis

Statistical analyses were done in SPSS (version 25.0; SPSS Inc., Chicago, United States). Descriptive statistics were calculated and reported as mean ± SD. The normality of the data distribution was verified with Shapiro-Wilk tests (all $p \leq 0.065$). A two-way ANOVA with one between-participant factor (i.e., sex) and one within-participant factor (fatigue) was run to explore the effects of sex, fatigue, as well as sex × fatigue interaction for CoP variables and indexes of transient behavior (i.e., the DIF_21 and DIF_21). The effect sizes were expressed as partial eta-squared (η^2) and interpreted as small (<0.13), medium (0.13–0.26), and large (>0.26; Bakeman, 2005). Bonferroni-corrected *t*-test were used to explore the effect of sex separately before and after fatigue, the effects of fatigue for each sex, and the effects of sex on rating of perceived exertion and differences fatiguing protocol duration between the sexes. The effect sizes for *t*-tests were calculated as Cohen's *d* (0.0–0.2 – trivial; 0.2–0.6 – moderate; 0.6–1.2 – large; >1.2 – very large; Bernards et al., 2017). We also examined the associations between whole-trial variables and corresponding indexes of transient behavior (i.e., the DIF_21 and DIF_21) before and after the fatigue. Correlations were assessed by Pearson's correlation coefficients and interpreted as negligible (<0.1), weak (0.1–0.4), moderate (0.4–0.7), strong (0.7–0.9), and very strong (>0.9; Schober and Schwarte, 2018). For all analyses, the threshold for statistical significance was set at $p < 0.05$.

RESULTS

There were statistically significant differences between the sexes regarding body height ($t = 6.1$; $p < 0.001$), body mass ($t = 8.6$; $p < 0.001$), and body mass index ($t = 2.9$; $p = 0.007$).

Pre-fatigue Analyses

In the pre-fatigue state, there was a statistically significant moderate correlation between whole-trial CoP ML velocity and its DIF_21 index ($r = 0.52$; $p = 0.005$). This association was also statistically significant when assessed separately for females ($r = 0.61$; $p = 0.018$), but not for males ($r = 0.43$; $p = 0.125$). The associations between the remaining variables and their corresponding DIF_21 or DIF_31 were not statistically significant, regardless if whole sample or separate gender subgroups were analyzed ($r = 0.09$–0.33; $p = 0.090$–0.667). Before the fatigue, the difference between the genders was only observed for whole-trial CoP amplitude in AP ($p = 0.041$; $d = 0.71$) and ML direction ($p = 0.023$; $d = 0.91$), with the females exhibiting lower values (see **Table 1** for descriptive statistics). However, when CoP variables were normalized to body height or body mass, there were no differences between the sexes ($p = 0.158$–0.434 and 0.112–0.496, respectively). There were no statistically significant correlations between height and CoP variables ($r = 0.03$–0.31; $p = 0.095$–0.893). None of the transient behavior indexes was different between sexes (all $p \leq 0.288$).

Effects of Fatigue on Whole-Trial Estimates

After the fatiguing protocol, the average rating of perceived exertion score was 8.7 ± 0.8 and was not statistically significantly different ($p = 0.376$) between males (9 ± 1) and females (9 ± 1). On average, the males and the females performed 30.8 ± 6.1 and 28.8 ± 6.5 CoD maneuvers, respectively, with the group difference not being statistically significant ($p = 0.411$). The descriptive statistics for all outcome variables related to postural sway before and after fatigue is available in **Table 1**. Statistically significant gender × fatigue interaction with medium effect sizes was found for whole-trial CoP velocity in AP ($p = 0.028$; $\eta^2 = 0.17$) and ML directions ($p = 0.019$; $\eta^2 = 0.19$). Both interactions were still present when the outcomes were normalized to body height ($p = 0.018$–0.027; $\eta^2 = 0.17$–0.19) or body mass ($p = 0.015$–0.019; $\eta^2 = 0.19$–0.21). *Post-hoc* test showed that the both variables substantially decreased in female participants ($p = 0.041$–0.045; $d = 0.54$–0.56), but remained similar in males ($p = 0.194$–0.294). Although not reaching statistical significance, there were also notable decreases in CoP AP and ML amplitude in females ($p = 0.056$–0.058; $d = 0.61$–0.62).

Effects of Fatigue on Transient Behavior of Postural Sway

None of the transient behavior indexes showed statistically significant gender × fatigue interaction. There were small to medium statistically significant main effects of fatigue for DIF_31 for CoP AP as well as DIF_21 for CoP ML amplitude ($p = 0.042$–0.049; $\eta^2 = 0.02$–0.14). The fatigue decreased the values of DIF_31 and DIF_21, which suggests that after fatigue, the CoP amplitude dropped to a larger

TABLE 1 | Postural sway variables and their transient behavior indexes before and after the fatiguing protocol.

Outcome variable	Before fatigue				After fatigue				Effect size (η^2)		
	Male		Female		Male		Female				
	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Gender	Fatigue	Interaction
CoP VEL AP (mm/s)	27.8	3.9	28.3	6.4	29.7	5.0	24.8	5.8	0.06	0.02	0.17*
CoP VEL ML (mm/s)	32.2	5.4	31.2	6.3	34.5	7.3	27.8	6.0	0.12	0.01	0.19*
CoP AMP AP (mm)	5.6	1.5	4.7	0.9	5.6	1.0	4.1	1.0	0.34*	0.06	0.06
CoP AMP ML (mm)	7.9	2.1	6.3	1.2	8.1	2.6	5.4	1.6	0.30*	0.03	0.07
CoP VEL AP – DIF_21 (%)	100.1	22.3	105.3	24.8	94.1	20.3	92.1	16.1	0.00	0.12	0.02
CoP VEL AP – DIF_31 (%)	93.4	14.4	92.1	16.1	86.4	17.9	88.0	21.6	0.00	0.05	0.00
CoP VEL ML – DIF_21 (%)	97.9	18.7	102.8	20.4	98.8	24.7	87.4	18.5	0.01	0.10	0.12
CoP VEL ML – DIF_31 (%)	88.6	18.0	93.0	24.4	90.1	28.1	85.6	15.5	0.00	0.01	0.02
CoP AMP AP – DIF_21 (%)	107.6	38.9	109.6	27.0	96.3	33.7	91.2	18.6	0.00	0.11	0.01
CoP AMP AP – DIF_31 (%)	99.6	18.5	92.3	17.0	83.8	27.9	83.2	22.9	0.02	0.14*	0.01
CoP AMP ML – DIF_21 (%)	101.7	26.3	107.2	24.7	94.5	30.5	85.9	26.1	0.00	0.14*	0.04
CoP AMP ML – DIF_31 (%)	95.2	29.4	93.7	36.5	90.6	39.5	84.9	27.8	0.01	0.02	0.00

CoP, centre of pressure; VEL, velocity; AMP, amplitude; AP, anterior-posterior; ML, medial-lateral; DIF_21, relative differences between the first and the second 10-s time intervals within the whole trial; and DIF_31, relative differences between the first and the third 10-s time intervals within the whole trial.
*Denotes statistically significant effect ($p < 0.05$).

extent throughout the 30-s trial. *Post-hoc t*-tests revealed that DIF_31 for CoP AP amplitude was decreased after fatigue in males ($p = 0.044$; $d = 0.65$), but not in females ($p = 0.310$). Looking at the descriptive statistics for interval-specific data for males (provided in **Table 2**) reveals that the effect of fatigue on in DIF_31 was as a result of an increased CoP AP amplitude in the first 10-s interval in males (before fatigue: 5.6 ± 1.3 mm; after fatigue: 6.3 ± 1.6 mm), while the CoP AP amplitude in the third interval was slightly decreased after fatigue (before fatigue: 5.5 ± 1.4 mm; after fatigue: 5.1 ± 1.2 mm).

In contrast to DIF_31 for CoP AP amplitude, the DIF_21 for CoP ML amplitude decreased after fatigue for females ($p = 0.024$; $d = 0.81$), but not for males ($p = 0.533$). This difference in females was a result in slightly increased CoP ML amplitude in first interval after fatigue (before fatigue: 6.6 ± 1.8 mm; after fatigue: 6.9 ± 1.8 mm) and concomitant drop during the second interval (before fatigue: 6.2 ± 1.5 mm; after fatigue: 5.3 ± 2.0 mm). Although no other statistically significant gender-specific effects were found, all data tended to indicate that after fatigue, males exhibit increased postural sway in the early phases of the trial and relatively unchanged postural sway in later phases of the trial. On the other hand, females decreased the postural sway in later phases of the trial, while their early postural sway was relatively unaffected by fatigue.

After fatigue, the correlation between whole-trial CoP ML velocity and its DIF_21 index ($r = 0.59$; $p = 0.002$) was still present and was similar in males only ($r = 0.55$; $p = 0.043$) and females only ($r = 0.57$; $p = 0.034$). There was also a correlation between whole-trial CoP ML amplitude and its DIF_21 index ($r = 0.65$; $p = 0.001$), which was also present both in males ($r = 0.76$; $p = 0.002$) and females ($r = 0.54$; $p = 0.048$).

DISCUSSION

The purpose of this study was to explore the effects of whole-body fatigue, induced by repeated CoD task on postural sway in single-leg stance and its transient characteristics. Our hypothesis was rejected, as the fatiguing protocol did not result in an increase of postural sway. On the contrary, postural sway was even reduced in females (the difference for CoP velocity, but not CoP amplitude being statistically significant), but did not change in males. Some of the indexes of transient postural sway behavior were influenced by fatigue. A closer examination of the data reveals that the changes are in accordance with our hypothesis, as the postural sway was more variable across trial after fatigue (in turn, the indexes of transient behavior were lower), with males being significantly affected in particular during the early phases of the trial. Although we expected larger deteriorations in postural sway, this study nevertheless generated two important findings. Firstly, the response to fatigue was time interval specific, which warrants further investigation of the transient behavior in relation to fatigue. Secondly, there were considerable differences between male and female participants in responses to fatigue, which points out that gender-specific effect of fatigue on postural sway.

Contrary to previous findings, some (positive) correlations were present between whole-trial-estimates and corresponding indexes of transient behavior. This implies that individuals with higher postural sway showed less stabilization (i.e., their sway was reduced less or was even increased throughout the trial). However, given that the associations were moderate, and the fact that four previous studies on larger sample sizes found no associations between whole-trial-estimates and corresponding indexes of transient behavior (Reed et al., 2020; Kozinc and Šarabon, 2021a,b; Kozinc et al., 2021), it is still reasonable to conclude that calculating and analyzing indexes of transient behavior can contribute new information to the traditional whole-trial analysis. Future studies investigating this topic should include correlational analysis to resolve this issue further.

The absence of the main effect of fatigue was a surprising finding, given that studies have reported increases in postural sway following aerobic and anaerobic (Fox et al., 2008; Güler

TABLE 2 | Interval-specific descriptive statistics.

Outcome variable	Before fatigue				After fatigue			
	Male		Female		Male		Female	
	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*
CoP VEL AP first interval (mm/s)	28.7	4.5	28.7	6.2	32.5	8.8	26.6	5.5
CoP VEL AP second interval (mm/s)	28.3	5.4	30.1	9.1	29.6	5.7	24.5	6.7
CoP VEL AP third interval (mm/s)	26.5	4.2	26.2	6.2	27.0	3.6	23.3	7.1
CoP VEL ML first interval (mm/s)	34.0	5.8	31.9	5.9	36.2	6.8	30.6	5.8
CoP VEL ML second interval (mm/s)	33.0	7.3	32.7	8.8	35.5	9.8	26.8	7.7
CoP VEL ML third interval (mm/s)	29.7	5.9	29.0	7.7	31.9	8.6	26.0	6.2
CoP AMP AP first interval (mm)	5.6	1.3	4.8	1.0	6.3	1.6	4.5	1.0
CoP AMP AP second interval (mm)	6.0	2.2	5.1	1.4	5.7	1.4	4.1	1.1
CoP AMP AP third interval (mm)	5.5	1.4	4.3	1.0	5.1	1.2	3.7	1.3
CoP AMP ML first interval (mm)	8.2	2.3	6.6	1.8	8.7	2.1	6.2	1.5
CoP AMP ML second interval (mm)	8.2	2.9	6.9	1.8	8.3	3.7	5.3	2.0
CoP AMP ML third interval (mm)	7.5	2.2	5.7	1.7	7.6	3.2	5.1	1.9

CoP, centre of pressure; VEL, velocity; AMP, amplitude; AP, anterior-posterior; and ML, medial-lateral.

et al., 2020), local and global (Zech et al., 2012), and sport-specific fatiguing (Greig and Walker-Johnson, 2007; Bedo et al., 2020). The effects of fatigue on postural sway could be related to reductions in muscle force producing capacity (Thorlund et al., 2008; Rampinini et al., 2011), which is observed after both central (Gandevia, 2001) and peripheral fatigue (Fitts, 1994). Other possible underlying causes of increases postural sway include changes in proprioception (Proske, 2019), and specifically for global fatigue, the perturbations introduced due to intensified breathing (Janssens et al., 2010). The most reasonable explanation for the lack of effect of fatigue on postural sway in the present study is that the fatiguing protocol was not sufficiently exhaustive to elicit the expected changes in males. The reduced postural sway in females after fatigue is the most difficult to interpret. The reductions of postural sway did not always imply better balance (Cho et al., 2014). It could be that females changed the postural control strategy. Anyway, while the type of fatiguing protocol is an important limitation of this study, at the same time, our results demonstrate that the indexes of transient behavior could perhaps detect smaller fatigue-induced changes that are masked by whole-trial estimates. Looking only at the whole-trial estimates, one could conclude that our protocol did not affect (males) or reduced (females) the postural sway, while the examination of interval-specific values showed increases sway (males) in the early phase of the trial. It is difficult to speculate what effect would be seen if a more exhausting protocol was used or if the participants were professional athletes. Likely, the effects of fatigue would be even smaller, considering that athletes' balance is less influenced by fatigue compared to general population (Baghbani et al., 2016).

We observed considerable differences regarding the responses in postural sway to fatigue between males and females. Previous studies have suggested that there are no differences in postural sway between the genders in the adult population (Hageman et al., 1995; Cruz-Gómez et al., 2011; Olchowik et al., 2015), or that females exhibit lower postural sway (Greve et al., 2013; Błaszczyk et al., 2014; Sell et al., 2018). Smaller postural sway in females has been often attributed to lower body height (Era et al., 1996; Bryant et al., 2005). However, smaller postural sway in females has been found compared to males after adjusting for height (Frandin et al., 1995), and studies have reported no association between anthropometric variables and postural sway (Ekdahl et al., 1989; Ageberg et al., 2001). While the gender differences in postural sway in the rested state could be at least partially influenced by body height, the causes for different responses to fatigue are likely elsewhere. Masui et al. (2005) reported that the postural sway in older adult males was more influenced by the removal of visual information, compared to older adult females. However, since our fatiguing protocol presumably affected proprioception (leaving vison unperturbed), it would be expected that males would be less affected by fatigue if they placed more importance to visual information. One possible explanation for our results is that females are in general better in sensory integration and sensory reweighing, although the available data on adult populations suggest no gender differences (Faraldo-García et al., 2012). Studies in children suggest that females are better at integrating their sensory inputs, while males treat each sensory input more independently (Smith et al., 2012). It could be that when the balance is challenged (e.g., in fatigued conditions, and/or after the transition to a more challenging posture, such as single-leg stance), the females are better and quicker in adapting their sensory integration in order to stabilize the body. Bannon et al. (2018) observed statistically significant increases in postural sway in males, but not in females, after fatiguing load-lifting task. They concluded that females are able to better adapt their balance strategies to fatiguing conditions than males. Indeed, Adlerton et al. (2003) found increased CoP amplitude and trunk accelerations after local ankle fatiguing in females, whereas CoP velocity was decreased, reflecting a change in the postural control strategy.

Some limitations of the study should be acknowledged. The fatigue protocol was somewhat atypical in comparison to previous studies, which warrant further investigation with different (e.g., local and global, aerobic, and anaerobic) fatiguing protocols. The participants in our study were physically active young adults, which means that our results cannot be generalized to other populations such as elite athletes or older adults. In addition, it has to be stressed that our results cannot be extrapolated to the dynamic tasks. Namely, some of the previous studies have observed an effect of fatigue on static balance (i.e., increased postural sway in quiet stance; Zech et al., 2012) without concomitant changes in dynamic balance. Moreover, the contribution of individual sensory systems, as well the sensory integration, is probably dependent on the dynamics and difficulty of the task (Bent et al., 2002, 2005). Finally, the reliability of the interval-specific CoP data and indexes of transient behavior is questionable. It is known that trial duration increases the reliability of CoP metrics (Carpenter et al., 2001). Since only one trial was conducted for this study in each condition, poor reliability could have affected the results. While we cannot calculate the reliability for this specific sample, we found relatively high coefficients of variation (among three repetitions of the 30-s trial) on a sample of 73 male soccer players (unpublished data). Specifically, the coefficients of variation ranged from 9 to 14% for whole trial CoP velocity and amplitude variables, and from 12 to 27% for interval-specific data. Indeed, postural sway assessed through CoP movements is highly variable, and coefficient of variation >10% is often observed (Hébert-Losier and Murray, 2020). We used one repetition in our protocol because the fatigue could subside quickly after the completion of the fatiguing protocol (Aboodarda et al., 2019; Güler et al., 2020). Nevertheless, more repetitions should be used prior to the fatiguing protocol to increase the rigor of the results. Before the research on the transient behavior of postural sway is advanced, a methodological study is urgently needed to determine the influence of variable choice, postural task, and trial duration (as well as the duration of individual intervals, used to assess transient behavior) on within- and between-session reliability.

CONCLUSION

The purpose of this study was to explore the effects of whole-body fatigue on postural sway in single-leg stance and its transient characteristics. Contrary to previous studies, postural sway did not increase with fatigue; rather it was even reduced in females. Some of the indexes of transient postural sway behavior were influenced by fatigue. Postural sway tended to be more variable across the trial after fatigue. This that the indexes of transient behavior could perhaps detect smaller fatigue-induced changes that are masked by whole-trial estimates. Males were particularly affected during the early phases of the trial. Thus, we can conclude that the response to fatigue in terms of postural sway is time interval specific, and that there are certain gender differences in responses to fatigue, which could be related to a better ability to adapt balance strategies in females. Transient characteristics of postural sway represent a potentially useful approach to analyze balance and postural control. Further studies will be needed to assess the utility of this approach for assessing sports performance and injury risk.

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 Republic of Slovenia National Medical Ethics Committee. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

ŽK, NT, and NŠ conceptualized the idea. NT and DS carried out the measurements. NŠ and DS were overviewing the measurement procedures and administration. NT and ŽK analyzed the collected data. ŽK wrote the manuscript. NŠ, NT, and DS finalized the manuscript. All authors contributed to the article and approved the submitted version.

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The Disturbing Effect of Neuromuscular Fatigue on Postural Control Is Accentuated in the Premenstrual Phase in Female Athletes

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This study explored the fatigue effect on postural control (PC) across menstrual cycle phases (MCPs) in female athletes. Isometric maximal voluntary contraction (IMVC), the center of pressure sway area (CoParea), CoP length in the medio-lateral (CoP_{LX}) and antero-posterior (CoP_{LY}) directions, and Y-balance test (YBT) were assessed before and after a fatiguing exercise during the follicular phase (FP), mid-luteal phase (LP), and premenstrual phase (PMP). Baseline normalized reach distances (NRDs) for the YBT were lower ($p = 0.00$) in the PMP compared to others MCPs, but the IMVC, CoParea, CoP_{LX}, and CoP_{LY} remained unchanged. After exercise, the IMVC and the NRD decrease was higher at PMP compared to FP ($p = 0.00$) and LP ($p = 0.00$). The CoParea, CoP_{LX}, and CoP_{LY} increase was higher in the PMP compared to FP ($p = 0.00$) and LP ($p = 0.00$). It was concluded that there is an accentuated PC impairment after exercise observed at PMP.

Keywords: athletes, static postural control, dynamic postural control, menstrual cycle, muscle fatigue

INTRODUCTION

Postural control (PC) is fundamental for human motor abilities not only to perform daily living activities (Taheri et al., 2019; Sabashi et al., 2021), but also to perform a high level of physical performance during sports training or competition (Geddam et al., 2014). PC assessment was usually conducted in athletes to identify or to prevent the risk of musculoskeletal injuries (Yim et al., 2018). In this context, it has been documented that postural balance impairments may affect physical performance (Zemková, 2014) and increase the risks for sports injuries (Paterno et al., 2010). Importantly, it has been reported that female athletes are more prone to the risk of injuries, such as anterior cruciate ligament injuries and lateral ankle sprains, compared to males while practicing the same sports activities (Ristolainen et al., 2009; Stijak et al., 2015). This higher injury rate in females compared to males has been linked to hormonal differences and fluctuations (Hewett et al., 2007). To this end, previous studies explored PC in women through different menstrual cycle phases (MCPs) governed by hormonal changes, but results remain controversial (Ekenros et al., 2011; Lee and Yim, 2016; Lee et al., 2017). In fact, in sedentary women, greater postural sway during or around ovulation compared to the other phases were observed (Petrofsky and Lee, 2015; Lee and Yim, 2016; Lee et al., 2017). As well, impaired static PC in the luteal phase

(LP) was found in moderately active women with premenstrual symptoms (PMS), but not in those without PMS (Fridén et al., 2003, 2005; Ekenros et al., 2011). Kaya and Çelenay (2016) indicated that the worst dynamic PC was during the menses [early follicular phase (FP)], when estrogen level was low, compared to mid-FP and mid-LP in active women. However, other studies did not find any significant effect of MCPs on PC (Fridén et al., 2005; Hertel et al., 2006; Abt et al., 2007; Ekenros et al., 2011; Ericksen and Gribble, 2012).

PC results from the integration of sensory information (visual, vestibular, and somatosensory) by the central nervous system (CNS) to recruit adequately the muscles (Horak, 2009). Hence, it could be impaired by the dysfunction in any part of these systems (Horak, 2009). Accordingly, it has been shown that neuromuscular fatigue of the lower limb muscles impairs PC (Konstantopoulos et al., 2021). It seems that neuromuscular fatigue impairs the muscle mechanical properties (Bilodeau et al., 2001) and the proprioceptive system (Hiemstra et al., 2001) required for postural balance regulation. In addition, it has been reported that PC is deteriorated due to both local and general fatigue, increasing the risk for traumatic musculoskeletal injuries (Gribble and Hertel, 2004; Pau et al., 2014). Since athletic performances need simultaneously musculoskeletal system integrity and well-developed PC (Ohlendorf et al., 2020), investigating the effect of neuromuscular fatigue on PC in athletes has been an interest for many studies (Thiele et al., 2015; Troester and Duffield, 2019). In this context, PC was shown to be more disturbed by neuromuscular fatigue in females compared to males (Gribble and Hertel, 2004; Whyte et al., 2015). Muscle force, as well as neuromuscular fatigue, could be also influenced by MCP. In fact, it has been found that the muscle force and endurance were higher during the FP compared to LP and menstrual phase (Pallavi et al., 2017). Recently, higher peripheral fatigue, associated with higher muscle damage, was observed in the premenstrual phase (PMP) compared to the MCP (Graja et al., 2020).

While the deteriorating effects of neuromuscular fatigue and MCP on PC have been widely explored independently, to our best knowledge, no previous study has been focused on the interaction effects of these factors on PC in female athletes. In fact, one of the most sport performance limits in female athletes is the higher knee injury rate specifically involving the anterior cruciate ligament compared to male athletes (Lee et al., 2013; Stijak et al., 2015), and the main underlying factor of this higher injury rate is the altered PC (Kjær and Hansen, 2008) linked to the hormonal fluctuations (Hewett et al., 2007). Both MC-related postural impairments and neuromuscular fatigue were known to result in increased sport-related injury risk. Thus, there is a great need for extending research in this area as the alterations in neuromuscular control may result in poorer control of movement, leading to an increased risk of non-contact lower limb injuries. Athletes and coaches would better be counseled regarding the MC effects on neuromuscular fatigue, PC, and risk of injury. By investigating the effect of MC hormonal fluctuations on PC following a fatiguing exercise, female athletes, coaches, and medical professionals could better understand the risk factors and find successful interventions for limiting this eventual impairment. To this end, they can suggest periodization of training programs for female athletes to take advantage of any optimal hormonal fluctuations and reduce the risk of musculoskeletal injury. Therefore, our study aimed to investigate the effect of MCP on static and dynamic PC after high-intensity fatiguing exercise in eumenorrheic female handball players.

MATERIALS AND METHODS

Participants

The sample size was *a priori* calculated as suggested by Beck (2013) using the software G*power for Windows (version 3.1.9.2; Heinrich Heine University Düsseldorf, Northrhine-Westphalia, Germany) as recommended by Faul et al. (2007) to determine the number of participants necessary to identify PC changes with fatigue. For computing sample size, we used the effect size Cohen's f (calculated based on a partial η_p^2) estimated at 0.86 based on a previously published work by Boyas et al. (2013). Values for an alpha, power, correlation among repeated measures, and the non-sphericity correction (ε) were set at 0.05, 0.8, 0.5, and 1, respectively. In total, to reach the desired power, data from at least four participants were deemed to be sufficient to minimize the risk of Type II statistical error. To accommodate a possible dropout of some participants, we recruited 12 female athletes. Our recruitment strategy consisted of a three-stage screening process to delineate the sample. In the first stage, we randomly selected 22 female handball players from the regional senior handball team. Seven out of these 22 players were excluded from the study because they did not meet all inclusion criteria. Females included were handball players from the regional senior handball team and had a regular menstrual cycle (28 ± 1 day). Exclusion criteria were clinical ankle instability, vestibular or visual impairments, lower limb musculoskeletal injury in the previous 6 months, premenstrual syndromes, pregnancy, contraception or hormonal supplementary, and injections through the previous 3 months. In the second stage, 15 out of a total number of players who met the inclusion criteria were selected. Three of the screened players did not accomplish the three test sessions for different causes. As a result, 12 female team handball players (age: 21.0 ± 1.6 years, height: 1.72 ± 0.05 cm, weight: 65.0 ± 5.6 kg, weekly training volume: 11 ± 1 h, training experience: 6 ± 1 years) with the regular menstrual cycle (28 ± 1 day) participated in this study. They practiced in five training sessions of 2 h each and participated in one competitive match weekly in the regional senior handball series. Following an explanation of all the experimental procedures as well as their risks and benefits, all participants provided their written informed consent prior to the experiment. The study protocol was conducted in accordance with the Declaration of Helsinki and approved by the Southern Committee Protection of Persons (CPP SUD: 0042/2017).

Experimental Design

Participants were invited to come four times to the institute laboratory. The first session was opted to familiarize participants with the experimental protocol including isometric maximal voluntary contraction (IMVC), static and dynamic balance tests, and the fatiguing exercise. This session was performed 3 days before the experimental protocol execution to eliminate any fear of using new equipment, and to ensure high-quality results. Moreover, anthropometric measurements were performed in this familiarization session.

The three test sessions corresponded to the three MCPs: FP (day 13), LP (day 21), and PMP (day 27). These three sessions were conducted in a randomized crossover design. In fact, some athletes started the first session in the FP. However, others started in the LP or PMP during the same experimental month. MCP was verified by an ovulation test on day 12–13 of the menstrual cycle. At each session, participants were required to successively perform postural tests, 5 min warm-up, three IMVC, the fatiguing exercise, an IMVC, and the postfatigue postural tests (Bizid et al., 2009; Chaubet et al., 2012; da Costa et al., 2019; Martins et al., 2020). For each test, the mean of the three preexercise trials was calculated and then statistically analyzed as the baseline value. It is necessary to achieve all tests in a short time period after the fatiguing exercise to avoid recovery, and based on previous studies investigating the effect of fatigue on muscle force and PC, only one trial for each test was performed at the postfatigue condition (Boyas et al., 2011; Bisson et al., 2014; Larson and Brown, 2018). The center of pressure (CoP) measurement and the Y-balance test (YBT) trials were randomized both in the pre and postexercise conditions. For all measurements, all participants were able to perform correctly the tasks in both prefatigue and postfatigue conditions. We did not have any discard/ repeat failed trials. The whole postfatigue test procedure (Zech et al., 2012) did not exceed 4 min as it has been previously demonstrated that PC is still significantly affected until 8 min postfatigue (Lin et al., 2009). The assessors were specialists who had significant experience in the physical activity field. All PC tests were taken by the same examiner, a PhD expert in balance measurement, and all MVCs measurements were made by the same PhD expert in dynamometric assessment.

MCP Discrimination

Rectal temperature was recorded every morning before arising from bed for three consecutive months before and during the experiment. The beginning of the FP was indicated by the onset of menses, and the beginning of ovulation was indicated by an increase in temperature of 0.5°C. In addition, on days 12 and 13 of the menstrual cycle, we used the ovulation test strips (OVU Test Pro LH) measuring the luteinizing hormone (LH) surge meaning that the ovulation was reached within 24–48 h after detection of the LH surge, then the LP started.

Muscle Force Assessments

Participants practiced a 5-min warm-up consisting of several submaximal contractions of knee extension muscles at a self-selected intensity. Then, they performed three IMVCs of the dominant leg (the leg used to kick the ball) knee extensor muscles under strong verbal encouragement. They were seated on an isometric dynamometer (Good Strength, Metitur, Finland) equipped with a cuff attached to a strain gauge. The hip and knee angles were set at 90° during all measurements. The highest value of these three IMVCs was considered as the reference IMVC to calculate the 80% IMVC for the fatiguing exercise. IMVC was tested once again immediately after the fatiguing exercise.

PC Assessment

Posturographic Measurements

Posturography is a method that assesses postural stability (Toppila et al., 2006), which is considered to be the gold standard for postural balance assessment (Huurnink et al., 2013). Moreover, it is non-invasive, objective, rapid, and easy to administer. The CoP excursions were collected before and after the fatiguing exercise using a static stabilometric platform (PIMVCostureWin, Techno Concept, Cereste, France; 12-bit A/D conversion), composed of a steel plate supported by three strain gauges. The CoP excursions were collected at a sampling rate of 40 Hz with respect to the French Association of Posturography recommendations (Association Française de Posturologie, 1985). The platform was leveled with the surrounding floor and placed at a 3 m distance from the dynamometer. Participants were asked to stand as still as possible on the force platform with barefoot on one leg (dominant leg) while the other leg flexed by 45° at the hip and knee so as to resemble the starting position of a front kick. Their arms are comfortably placed downward at either side of the body. A plastic device was provided with the platform to maintain the same foot positions for all the measurements. This posturographic test was performed in two visual conditions. In the eyes open (EO) condition, participants were instructed to look straight ahead at a white cross placed onto the wall 2 meters away at eye level. In the EC conditions, in order to ensure the absence of visual information and to perform the same task, they wore a blindfold. They were asked to keep their gaze horizontal in a straight-ahead direction. Trials were randomized to eliminate learning or fatigue effects. Three trials were performed in each experimental condition for each participant before the fatiguing exercise whereas only one trial was performed for the postfatigue assessment. Each trial lasted 25.6 s (French Posturography Association norms) and 1 min of rest was allowed between two consecutive trials. Data collection was initiated after participants adopted the required posture on the platform, stabilized their postural sway, and signaled the experimenter that they were ready to begin. Then, the CoP excursions were computed from the ground reaction forces and their associated torques. During the upright standing postures, participants oscillate with their body relatively rigid, and the reaction force applied to the body is relatively constant. Thus, the associated torque variations depend mainly on the CoP excursions (Asseman et al., 2008). The muscular torque that controls the body oscillations is represented by CoP excursions (Winter et al., 1998; Asseman et al., 2008). CoP signals were smoothed using a second-order Butterworth filter with a 10 Hz low-pass cutoff frequency. To evaluate the static PC of our participants, three CoP sways parameters were analyzed in this study: The CoParea (CoParea; 90% confidence ellipse) was analyzed to evaluate the stability performance of participants. The lower the CoP sway, the better the PC will be (Caron et al., 2000). Furthermore, the CoP lengths corresponding to the sum of CoP displacement in the medio-lateral (CoP_{LX}) and in antero-posterior (CoP_{LY}) directions reflecting the topographical features of the plantar pressure distribution (Han et al., 1999) were analyzed. These parameters were averaged for the three trials per condition and per participant for statistical analysis. In case of a failed trial, data collection was stopped and the trial was repeated. A trial was considered as a failure if the participant put the contralateral foot down, stepped out of position, changed her feet or arms from the starting position, touched something for support, missed the stopwatch pad, or lost balance before completing the 25.6 s (Alsubaie et al., 2019).

Dynamic PC Assessment

The YBT is a commercially available tool for assessing dynamic PC, and it possesses excellent intra-tester (0.85–0.89) and inter-tester (0.97–1.00) reliability (Plisky et al., 2009). In each MCP, athletes performed the YBT with three trials before exercise to consider the mean as the preexercise value and only one trial immediately after exercise. This test was performed by the dominant leg randomly in three directions derived from the Star Excursion Balance Test (SEBT): the anterior, the posteromedial, and the posterolateral directions. This test is considered to be an indicator of lower limb injury risk (Plisky et al., 2006; Smith et al., 2015). Participants have to maintain their balance on the dominant leg while sliding a block as far as possible in each direction with the contralateral leg. The criteria denoting a failed trial were chosen in line with previously published literature. A trial was discarded and repeated if participants (i) took the weight on the reaching foot; (ii) failed to bring back the reaching foot to the starting position without losing control; (iii) failed to keep both hands on hips; (iv) failed to keep the stance foot at the same place; or (v) kicked the reach-block to gain more distance (Plisky et al., 2009). Measures of the YBT reach distances were normalized for leg length to calculate the normalized reach distance (NRD) using the following formula:

$$NRD\,(\%) = [Reach\ Distance\,(cm)/\ Leg\ Length\,(cm)] \times 100$$

Leg length was obtained by measuring the distance between the anterior-superior iliac spine and the most distal aspect of the medial malleolus (Gribble and Hertel, 2003). A composite score of the overall YBT was calculated by averaging the individual normalized scores for each direction (Butler et al., 2013).

Fatiguing Exercise

Our fatigue protocol is a modified version from that previously described by Smith-Ryan et al. (2014). Participants were required to follow their force production on a computer screen placed in front of them, which displayed their real-time digitized force signal. Visual feedback was provided on the screen at 80% of their IMVC with a red line. All participants were asked to track this line for 7 s followed by a 3-s recovery. This protocol, which lasted for 4 min, involved four sets of six intermittent isometric contractions. In order to facilitate the timing of the exercise, they have to hear two different audible metronomes that were repeated during the 4 min of the fatiguing exercise, the first beep was to start the contraction and the second beep was to stop it.

Statistical Analysis

All data were presented as mean ± SD and were analyzed by the software Statistica 10 (Statsoft, France). Data distribution normality was confirmed with the Shapiro-Wilk test. As well, the homogeneity of the variance was verified with the Leven test. CoParea, CoP_{LX}, and CoP_{LY} values were analyzed using a three-way ANOVA with repeated measures (phase × exercise × vision). IMVC; NRD for anterior, posteromedial, and posterolateral directions; and the overall YBT score were analyzed using two-way ANOVA with repeated measures (phase × exercise).

For all analyses, when ANOVA showed a significant effect, a *post-hoc* test (Bonferroni) was performed. Effect sizes (η_p^2) were calculated to assess the practical significance of our findings as recommended by Mullineaux et al. (2001). The level of significance for all statistical analyses was set at $p < 0.05$.

RESULTS

Isometric Maximal Voluntary Contraction

The statistical analysis revealed significant main phase effect [$F_{(2, 20)} = 28.65$; $p < 0.001$; $\eta_p^2 = 0.72$], main exercise effect [$F_{(1, 10)} = 506.25$; $p < 0.001$; $\eta_p^2 = 0.97$], and main phase × exercise interaction [$F_{(2,20)} = 16.95$; $p < 0.001$; $\eta_p^2 = 0.60$] on IMVC values. The *post-hoc* test (Bonferroni test) showed that there was no significant difference of IMVC at the preexercise among the three MCPs ($p = 1.00$ for FP compared to LP; $p = 0.08$ for FP compared PMP; $p = 1.00$ for LP compared to PMP). After exercise, IMVC values decreased significantly ($p < 0.001$) in all phases. This decrease was significantly higher in PMP compared to FP ($p < 0.001$) and LP ($p < 0.001$) and in FP ($p < 0.01$) compared to LP (**Figure 1**).

Static PC

Results of CoP parameters were presented in **Figures 2–4**. Statistical analysis showed significant phase, exercise, and vision main effects on CoParea, CoP_{LX}, and CoP_{LY} values. Moreover, phase × vision, phase × exercise, vision × exercise, and phase × vision × exercise interactions were revealed on these parameters (**Table 1**). The *post-hoc* test showed that before exercise, there is no significant difference in CoParea, CoP_{LX}, and CoP_{LY} values between MCPs neither in the EO ($p = 1.00$) nor in the EC ($p = 1.00$ for CoParea and CoP_{LY} between all MCP; for CoP_{LX} $p = 1.00$ between FP and LP; $p = 0.15$ between FP and PMP; $p = 1.00$ between LP and PMP) condition. Following exercise, CoParea, CoP_{LX}, and CoP_{LY} increased significantly ($p < 0.01$) in all MCPs in both vision conditions. In addition, after exercise, our results showed that these values were significantly higher in the PMP compared to FP ($p < 0.001$) and LP ($p < 0.001$) in the EC condition but not in the EO one ($p = 1.00$ for CoParea and CoP_{LY} between all MCPs; for CoP_{LX} $p = 1.00$ between FP and LP; $p = 0.45$ between FP and PMP; $p = 1.00$ between LP and PMP).

Concerning the visual effect, participants demonstrated significantly ($p < 0.001$) higher CoParea, CoP_{LX}, and CoP_{LY} in EC compared to EO condition regardless of exercise or MCP factors.

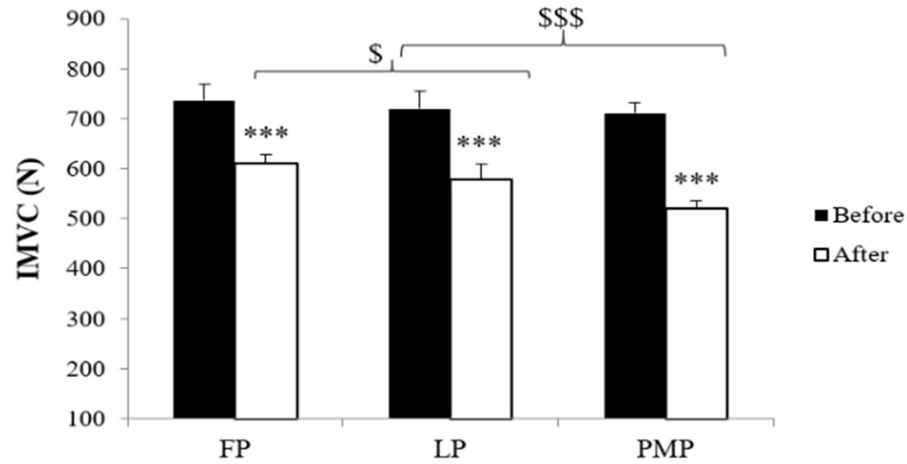

FIGURE 1 | Isometric maximal voluntary contraction (IMVC) (means ± SD) during the follicular phase (FP), mid-luteal phase (LP), and premenstrual phase (PMP) before and after exercise. ***Significant difference compared with before exercise ($p < 0.001$), $significant difference between FP and LP ($p < 0.05$), $$$significant difference compared with PMP ($p < 0.001$).

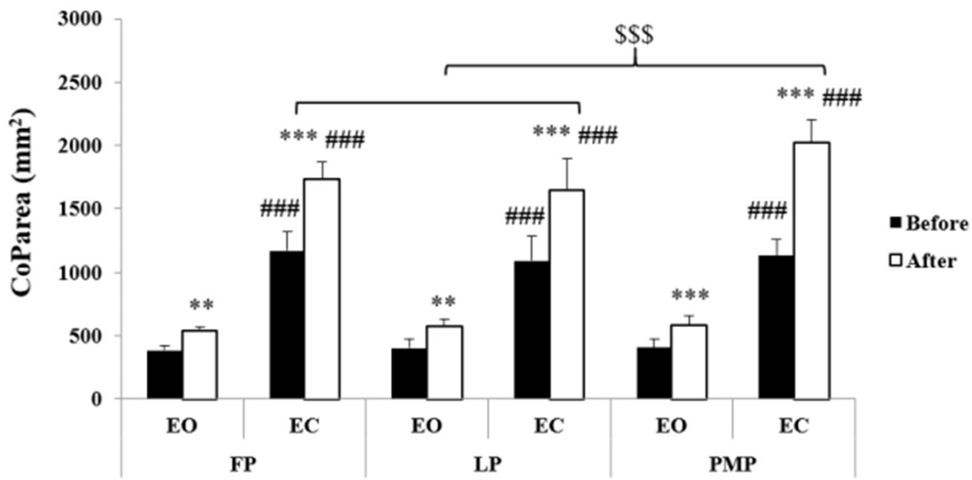

FIGURE 2 | Center of pressure area (means ± SD) during the follicular phase (FP), mid-luteal phase (LP), and premenstrual phase (PMP) in the eyes open (EO) and the eyes closed (EC) conditions before and after exercise. **Significant difference compared with before exercise ($p < 0.01$), ***significant difference compared with before exercise ($p < 0.001$), ###significant difference compared with EO ($p < 0.001$), $$$significant difference compared with PMP ($p < 0.001$).

YBT Performance

Statistical analysis showed significant main phase, exercise, and phase × exercise interaction effects on NRD for anterior, posteromedial, and posterolateral directions and the overall YBT score (**Table 2**). Before exercise, these NRD values for all directions as well as the overall YBT score were significantly lower in the PMP compared to FP ($p < 0.001$) and LP ($p < 0.001$) with a non-significant difference between FP and LP (NRD values for all directions, $p = 1.00$; the overall YBT reach direction score, $p = 0.59$). After exercise, all NRD values decreased significantly ($p < 0.001$) in all MCPs (**Table 3**). This decrease was higher in the PMP compared to FP ($p < 0.001$) and LP ($p < 0.001$) without significant difference between FP and LP (NRD for anterior and posteromedial directions, $p = 1.00$; NRD for posterolateral direction, $p = 0.56$; the overall YBT score, $p = 0.25$).

DISCUSSION

The current study is the first to investigate the effect of neuromuscular fatigue on PC during the different MCPs in eumenorrheic female athletes. The main results of this study showed that the disturbing effect of neuromuscular fatigue on static and dynamic PC was accentuated in the PMP compared to FP and LP. Nevertheless, the present findings demonstrated that IMVC and static PC baseline levels were not influenced by the

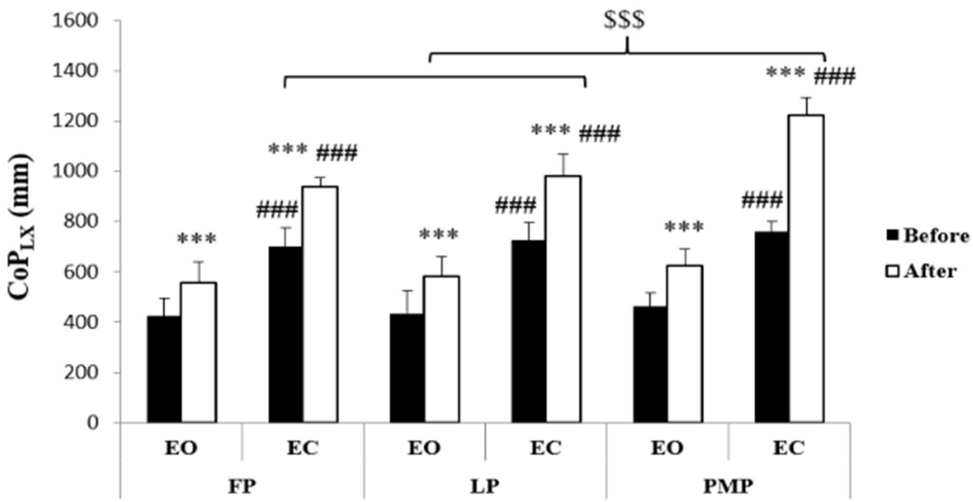

FIGURE 3 | Center of pressure medio-lateral length (means ± SD) during the follicular phase (FP), mid-luteal phase (LP), and premenstrual phase (PMP) in the eyes open (EO) and the eyes closed (EC) conditions before and after exercise. ***Significant difference compared with before exercise ($p < 0.001$), ###significant difference compared with EO ($p < 0.001$), $$$significant difference compared with PMP ($p < 0.001$).

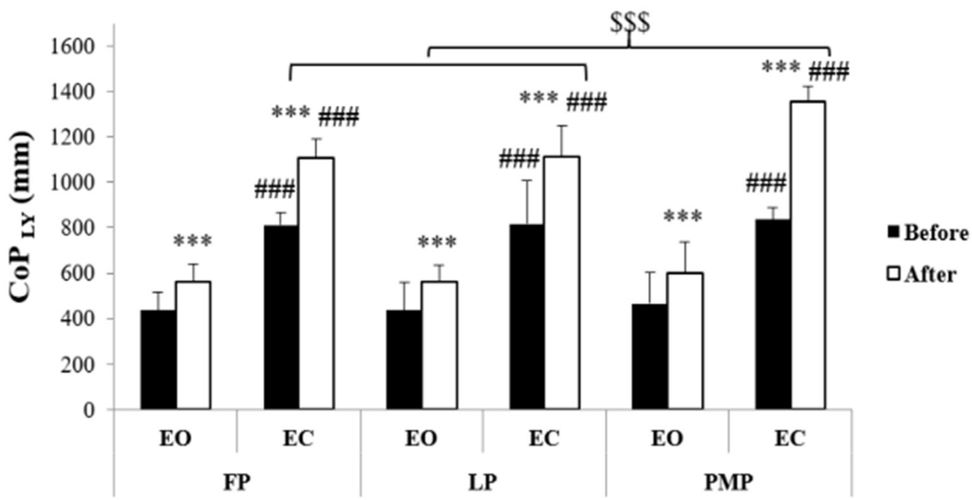

FIGURE 4 | Center of pressure antero-posterior length (means ± SD) during the follicular phase (FP), mid-luteal phase (LP), and premenstrual phase (PMP) in the eyes open (EO) and the eyes closed (EC) conditions before and after exercise. ***Significant difference compared with before exercise ($p < 0.001$), ###significant difference compared with EO ($p < 0.001$), $$$significant difference compared with PMP ($p < 0.001$).

MCP. This study's results are in line with some previous studies showing that MCs did not affect neither muscle strength (Graja et al., 2020) nor static PC (Hertel et al., 2006; Abt et al., 2007; Kaya and Çelenay, 2016) in eumenorrheic women. However, a few studies demonstrated that muscle strength (Pallavi et al., 2017; Rodrigues et al., 2019) and static PC (Fridén et al., 2003, 2005) fluctuate across the MCP. Concerning dynamic PC, this study showed that baseline YBT performance was altered in the PMP compared to FP and LP. This alteration could be explained by the impact of estrogen and progesterone concentration variations during MCP on the CNS function *via* binding to related neurotransmitters and altering their interactions (Ishii et al., 2009). In fact, female sex hormones may influence the role of the CNS and consequently the dynamic PC (Demir and Rasmi, 2015). This hormonal change may compromise the homeostasis of labyrinthine fluids, which might alter the balance ability (Demir and Rasmi, 2015). Furthermore, the decrease in estrogen level in the PMP may induce a decrease in serotonin levels and cause impaired mood, which negatively affects PC (Kaya and Çelenay, 2016). In fact, our findings are in accordance with Kaya and Çelenay (2016) who reported a higher dynamic PC alteration in menses (low estrogen and progesterone levels) compared to

TABLE 1 | ANOVA results of center of pressure area (CoParea), CoP medio-lateral length (CoP_{LX}), CoP antero-posterior length (CoP_{LY}) values.

	CoParea (mm^2)			CoP_{LX} (mm)			CoP_{LY} (mm)		
	F	*p*	η_p^2	*F*	*p*	η_p^2	*F*	*p*	η_p^2
Phase (P)	8.51	<0.01	0.43	35.5	<0.001	0.76	12.71	<0.001	0.53
Exercise (E)	1276.63	<0.001	0.99	978.78	<0.001	0.98	2016.30	<0.001	0.99
Vision (V)	1218.73	<0.001	0.99	788.97	<0.001	0.98	752.15	<0.001	0.98
P × E	19.30	<0.001	0.63	31.90	<0.001	0.74	35.33	<0.001	0.76
P × V	6.13	<0.01	0.35	20.64	<0.001	0.65	5.13	<0.05	0.31
E × V	632.26	<0.001	0.98	122.67	<0.001	0.91	198.93	<0.001	0.94
P × E × V	15.62	<0.001	0.58	23.17	<0.001	0.67	17.18	<0.001	0.60

TABLE 2 | ANOVA results of normalized reach distance (NRD %) for anterior, posteromedial, and posterolateral directions and the overall Y-balance test (YBT) score.

	NRD for anterior direction			NRD for posteromedial direction		
	F	*p*	η_p^2	*F*	*p*	η_p^2
Phase (P)	36.08	<0.001	0.76	29.37	<0.001	0.72
Exercise (E)	301.6	<0.001	0.96	284.12	<0.001	0.96
P × E	8.83	<0.01	0.44	11.03	<0.001	0.50
	NRD for posterolateral direction			**The overall YBT score**		
	F	*p*	η_p^2	*F*	*p*	η_p^2
Phase (P)	31.27	<0.001	0.73	83.85	<0.001	0.70
Exercise (E)	86.41	<0.001	0.88	451.48	<0.001	0.92
P × E	11.93	<0.001	0.52	33.22	<0.001	0.48

FP and LP without any significant difference between FP and LP. In addition, Emami et al. (2019) demonstrated that the NRD of posteromedial direction in YBT was significantly better in the ovulation phase (days 12–14; high estrogen level) in comparison to the early FP (days 3–5).

After exercise, our results showed that IMVC, CoParea, and all NRD values were altered in all MCPs suggesting that neuromuscular fatigue impaired both static and dynamic PC. Muscle force loss seems to be one of the contributing factors of static and dynamic postural impairment resulting from neuromuscular fatigue (Paillard, 2012). In fact, it was evidenced that neuromuscular fatigue deteriorates both sensory input and motor output of the postural system (Paillard, 2012). Moreover, neuromuscular fatigue is known to induce proprioceptive, vestibular, and visual sensory information disturbance (Paillard, 2012). This result is in agreement with previous studies showing that unilateral fatiguing exercises degraded static (Paillard et al., 2014) and dynamic (Whyte et al., 2015; Johnston et al., 2018) PC abilities.

The major results of our study demonstrated that the disturbing effect of neuromuscular fatigue on both static and dynamic PC was accentuated in the PMP compared to LP and FP. The accentuated amount of neuromuscular fatigue in the PMP manifested by the greater force loss after exercise could in part explain these results. Indeed, Graja et al. (2020) found a greater decrease in quadriceps muscle strength in the PMP compared to the FP and LP after high-intensity intermittent exercise and reported a significant negative correlation between the muscle force loss and the estrogen levels in eumenorrheic female athletes. As well, the correlation between force loss and PC perturbation was already demonstrated in a previous study (Borji et al., 2017). However, whether this impaired PC is due to neural or muscular figure mechanisms is still less documented. In this context, accentuated peripheral fatigue and decreased conduction velocity after fatiguing isometric contractions have been shown in PMP than other MCPs (Soares et al., 2011). According to Paillard (2012), local muscle fatigue induces muscle properties modifications including the action potential, intracellular, and extracellular ions, and other intracellular metabolites. These modifications decrease muscle excitability leading to strength loss (Paillard, 2012). The conduction velocity of afferent inputs decelerates and induces a propagation velocity decrease of the motor output, which is needed for maintaining postural balance (Paillard, 2012). Earlier researches revealed significant changes in quadriceps contractile properties and fatigability throughout the menstrual cycle (Sarwar et al., 1996). Moreover, the changes in estrogen and progesterone levels during the menstrual cycle may have an effect on neurological function (Woolley, 1999). To the best of our knowledge, the current study is the first one exploring the interaction effect of neuromuscular fatigue on PC across the different MCPs. Thus, we speculated that this accentuated alteration of CoP sway at the PMP at postexercise could be related

TABLE 3 | Normalized reach distances (NRDs %) for anterior, posteromedial, and posterolateral directions and the overall Y-balance test (YBT) score in the follicular phase (FP), mid-luteal phase (LP), and premenstrual phase (PMP) before and after exercise.

	FP		LP		PMP	
	Before	**After**	**Before**	**After**	**Before**	**After**
NRD for anterior direction (%)	68.35 ± 4.35	65.91*** ± 4.54	67.86 ± 4.24	65.39*** ± 3.37	63.98$$$ ± 4.00	59.79***$$$ ± 3.67
NRD for posteromedial direction (%)	111.49 ± 3.99	108.62*** ± 3.73	111.33 ± 4.02	108.52*** ± 4.64	109.35$$$ ± 3.70	104.70***$$$ ± 4.12
NRD for posterolateral direction (%)	108.97 ± 3.84	106.11*** ± 3.45	108.48 ± 3.23	105.39*** ± 3.27	106.62$$$ ± 3.03	101.73***$$$ ± 3.87
The overall YBT reach direction score	96.27 ± 20.43	93.55*** ± 20.21	95.89 ± 20.48	93.10*** ± 20.25	93.32$$$ ± 21.36	88.74***$$$ ± 21.14

*** *Significant difference compared with before exercise ($p < 0.001$),* $$$ *significant difference compared with PMP ($p < 0.001$).*

to low estrogen level. More precisely, the low estrogen level in the PMP is associated with negative mood, which may affect the central interactions of visual, vestibular, and somatosensory inputs needed for PC (Ekenros et al., 2011). Accordingly, one could argue that the low progesterone level in cerebellar pathways occurring in the PMP may increase postural sway (Darlington et al., 2001).

Concerning the dynamic PC, the lower YBT baseline performance reported in the PMP could explain the accentuated exercise-induced impairment in this phase compared to others. The YBT has been generally used to prospectively assess injury risk (Plisky et al., 2006; Wassinger et al., 2014; Smith et al., 2015). Female athletes with an overall score of <94% are considered to be at high risk to lower extremity injury (Plisky et al., 2006). The results of the current study demonstrated that the overall YBT baseline score was lower than 94% only in the PMP and that this score decreased after fatigue in all MCPs with a greater extend in the PMP. These results suggest that the decrease of both estrogen and progesterone levels may place female athletes at a high injury risk at fresh muscle condition as well as fatigued muscle. Importantly, it has been documented that female athletes taking hormonal contraceptives had a lower injury rate, because of the high estrogen and progesterone levels in oral contraceptive pills inhibiting their variations (Nielsen and Hammar, 1991). In accordance with our results, Wojtys et al. (2002) demonstrated that the risk of injury increases a week before the start of the menstrual period in female handball players. Nevertheless, our study is the first to explore the combination between MCPs and neuromuscular fatigue effects on dynamic PC. Therefore, further studies are needed to better understand this issue. An important result could be immerged from the current study revealing that the interaction between neuromuscular fatigue and MCP effect on static PC depends on the postural task difficulty, especially the visual input availability. In fact, the disturbing effect of neuromuscular fatigue was greater in the PMP compared to the other MCP only in the EC condition. It is well-known that altered visual input increases the challenge to balance and is associated with PC perturbation (Tse et al., 2013). This is demonstrated by the higher CoP sway in the EC condition compared to the EO. Furthermore, the higher increase of CoP sway in the EC condition compared to the EO one suggests that the disturbing effect of neuromuscular fatigue on static PC was accentuated with visual input removal. Our results are in line with those of Bisson et al. (2011), who reported that the postural balance impairment related to neuromuscular fatigue was higher when vision was removed. This result could be explained by a change in sensory inflow or integration as well as a possibly altered central processing of proprioceptive input. In fact, muscle fatigue is known to hamper sensory information processing (Vuillerme et al., 2001, 2005). Besides, the visual input appears to compensate for the inordinate reafference from fatigued muscles during maintenance of upright stance (Boyas et al., 2011). Thus, vision removal increases the disturbing effect induced by neuromuscular fatigue on PC. In this context, it has been demonstrated that neuromuscular fatigue increases the Romberg's index (RI) (Borji et al., 2017), which evaluates vision contribution in PC (Njiokiktjien and Van Parys, 1976).

Our study presents some limitations. First, despite the fact that neuromuscular fatigue can be assessed by the decrease of muscle force as conducted in this current study, it will be more efficient to determine the central and peripheral contributions in the accentuated disturbing effect of neuromuscular fatigue on PC across MCPs using the twitch interpolation technique. Second, some techniques like serum tracking and profiling might be a better option to check estrogen and progesterone concentrations; but it could not be possible to use this technique due to ethical reasons. Further, in the current study, we implemented a local fatiguing exercise consisting of intermittent isometric contractions of the quadriceps muscle at 80% IMVC to explore the effect of MCP and neuromuscular fatigue on PC in female athletes. Even though local quadriceps muscle exercises were widely used in this issue (Paillard, 2012), it seems interesting to investigate other types of fatiguing exercise such as a general whole-body fatiguing protocol in further studies. Finally, our measurements were limited to the fatigued limb. Future investigations are warranted to determine if hormonal fluctuations had similar effects on both the fatigued limb and non-fatigued one or during the bipedal stance.

CONCLUSION

Results of the current study suggest that baseline muscle force and static PC were not affected by the MCP hormonal variations, whereas dynamic PC was worse in the PMP compared to the other MCP in female athletes. Moreover, the disturbing effect of neuromuscular fatigue was accentuated in the PMP than the other phases, which may represent a risk factor for musculoskeletal injury in female athletes. It is worth noting to consider PC fluctuations across different MCPs when prescribing exercise programs for female athletes.

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 Southern Committee for the Protection of Persons (CPP SUD) at the University Hospital (CHU) of Habib Bourgiba Sfax Tunisia. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

MK and RB conceived, designed the research, analyzed the data, edited the draft, and finalized the manuscript. MK performed the experiments. RB and SS helped to write the draft of the manuscript. HR was involved in collecting additional and further editing of the final manuscript. All authors read and approved the final version of this manuscript.

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Observational Screening Guidelines and Smartphone Accelerometer Thresholds to Establish the Intensity of Some of the Most Popular Core Stability Exercises

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The lack of training load control, mainly exercise intensity, is one of the main limitations of core stability (CS) programs, which makes the training individualization and the analysis of the dose-response relationship difficult. The objectives of this study were to assess the inter-and intra-rater agreement when using new observational screening guidelines to decide if a core stability exercise represents an adequate training intensity level for a given participant. Besides, the relationship between experts' ratings based on these criteria and pelvic accelerations recorded with a smartphone accelerometer was also analyzed. Ten healthy physically active participants with a smartphone accelerometer placed on their pelvis were video-taped while performing a progression of seven variations of the front bridge, back bridge, side bridge and bird-dog exercises. Two expert and four non-expert raters watched the videos and used the observational screening guidelines to decide for each exercise variation if it represented an adequate training intensity level or not. In order to analyze the inter-and intra-rater agreement, several Kappa (κ) statistics were used. Receiver operating characteristic (ROC) curves to explore if the accelerometry allowed to establish pelvic acceleration thresholds representing the minimum level of exercise intensity for CS training. Cut-off acceleration values were calculated balancing sensitivity (Se) and 1-specifity (1-Sp) indexes (i.e., Youden index) or minimizing 1-Sp. The intra-and inter-rater analysis showed a substantial-high level of agreement with a prevalence-adjusted bias-adjusted Kappa > 0.69. The ROC curves showed that the acceleration thresholds for the bridging exercises were very similar, with global cut-off values of 0.35 m/s^2 (Se = 82%; 1-Sp = 15%) when using the Youden Index and of 0.50 m/s^2 when minimizing 1-Sp (Se = 31%), whilst the bird-dog exercise showed lower cut-off values (Youden Index: 0.21 m/s^2, Se = 90%, 1-Sp = 16%; minimizing 1-Sp: 0.32 m/s^2, Se = 40%). Overall, this study provides observational screening guidelines and smartphone accelerometer thresholds to facilitate the decision-making process when setting the intensity of some of the most popular core stability exercises in young physically active individuals.

Keywords: trunk exercise, training load, pelvic acceleration, expert rater, postural control

INTRODUCTION

Based on the results of previous studies, exercises for improving core stability (CS) have frequently been used as an additional training routine for professional and amateur athletes to improve athletic performance (Sato and Mokha, 2009; Sandrey and Mitzel, 2013; Trecroci et al., 2020) and to prevent and rehabilitate musculoskeletal injuries (Gouttebarge and Zuidema, 2018; Khaiyat and Norris, 2018). In addition, CS exercises have been effective in improving balance, functional performance and preventing the risk of falls in older adults (Granacher et al., 2013) and in reducing pain and disability in chronic low back pain patients (Mueller and Niederer, 2020). Most CS exercises, such as *bridge/plank* and *bird-dog* exercises, consist of maintaining different lying or quadruped postures that challenge the participants' ability to hold a neutral lumbopelvic position (Okubo et al., 2010; Vera-Garcia et al., 2014, 2020; Barbado et al., 2018; El-Gohary et al., 2018). The level of difficulty of these exercises, i.e., the lumbopelvic postural control challenge imposed on the participants, has been related to CS exercise intensity (Barbado et al., 2018) and generally modulated by manipulating different biomechanical constraints (i.e., lever arms, unsupported body mass, number and motion of elevated limbs, base of support, use of labile surfaces, etc.), (Mills et al., 2005; Parkhouse and Ball, 2011; García-Vaquero et al., 2012; Boucher et al., 2016; Vera-Garcia et al., 2020) according to the criteria of the people who select and prescribe the exercises.

Although training intensity is one of the main characteristics of the exercise programs, basic information on how to control and manage the intensity of CS exercises is lacking. For example, there are difficulties in establishing whether a prescribed exercise intensity is appropriate for the participant's level, as well as in determining, after a certain amount of workout sessions, if the given exercises are challenging enough for that participant or if it is necessary to progress toward other more intense exercises. In this sense, although randomized controlled trials on CS training programs usually report that CS exercises are prescribed based on participant's characteristics, the exercise intensity selection and its progression throughout the training program are normally conducted based on the experience and criteria of the professionals who develop the training programs, rather than on objective and quantifiable CS assessments (Cabanas-Valdés et al., 2016; Fox et al., 2016; Prieske et al., 2016; Doganay et al., 2020). Furthermore, the expert criteria used to individualize and modulate the intensity of these CS exercises are not normally specified in these studies (Areeudomwong and Buttagat, 2019; Kim and Yim, 2020). All these limitations hinder the replication of these interventions and do not allow the dose-response characterization of the CS exercise programs (Barbado et al., 2018).

Several biomechanical techniques have been used to objectively quantify the CS exercise intensity and to develop exercise progressions in different populations. In this sense, surface electromyography has traditionally been used to describe the trunk muscle activity intensity during many different CS exercises (García-Vaquero et al., 2012; Vera-Garcia et al., 2014; Calatayud et al., 2017), which supposes an internal index of CS exercise intensity. In addition, post-urographic techniques based on force platforms and smartphone accelerometers have been recently used to assess the participants' difficulty to control trunk posture during CS exercises (Barbado et al., 2018; Guillén-Rogel et al., 2019; Vera-Garcia et al., 2020), which represents an external index of CS exercise intensity. Despite the widespread use of electromyography and force platform post-urography in laboratory settings, their use outside the laboratory is limited due to the cost and complexity of these techniques. On the other hand, considering the low cost, portability, easy use and reliability of smartphone accelerometry (Barbado et al., 2018; Guillén-Rogel et al., 2019), it seems a useful and accessible technique to objectively quantify and control CS exercise intensity in many different contexts (e.g., clinical, athletic and research settings). However, despite the potential of this technique, to the best of our knowledge, no studies have analyzed which acceleration levels represent a sufficient or adequate exercise intensity to induce CS adaptations, nor how changes in the acceleration of CS exercises throughout a training program could be interpreted. Therefore, further research is needed to explore the smartphone accelerometer usefulness to objectively control and manage the intensity of CS exercise programs.

Considering that no literature exists with criteria to help to decide which are the best CS exercise intensity levels for each individual, observational screening guidelines targeting body alignment and postural sway were developed in this study to guide the decision-making process when establishing the intensity level of some of the most popular isometric CS exercises: front bridge, back bridge, side bridge and bird-dog. The main aims of this study were: (i) to analyze the degree of agreement between the evaluations performed by expert and non-expert raters (inter-and intra-rater agreement) using the observational screening guidelines; and (ii) to assess the relationships between the experts' observational assessments and the pelvic sway recorded with a smartphone accelerometer to ultimately try to establish pelvic acceleration thresholds representing the minimum level of exercise intensity for CS training.

MATERIALS AND METHODS

Participants

Ten healthy physically active individuals (males: $n = 7$; age $= 26.60 \pm 3.13$ years; height $= 179.14 \pm 6.04$ cm; mass $= 73.00 \pm 5.75$ kg; females $= 3$; age $= 26.33 \pm 1.15$ years; height $= 167.00 \pm 2.65$ cm; mass $= 65.30 \pm 6.75$ kg) voluntarily participated in this research. Participants were included in the study if they: (i) did not suffer a disease that contraindicated physical exercise practice (e.g., severe respiratory diseases, hypertension, heart disease, musculoskeletal injuries, etc.); (ii) did not suffer from urinary incontinence; (iii) did not suffer an inguinal hernia; (iv) were under 30 years old; and (v) were not pregnant. Participants were recreationally active, performing 2–5 sessions of 30–120 min of light to vigorous physical activity (jogging, resistance exercises, soccer, gymnastics, cycling, mountain bike, rugby, etc.) per week. None of them participated in a structured CS program at the time of the study, although all of them were familiar with the performance of bridging and bird-dog exercises. At study entry, participants signed an informed consent approved by the University Office for Research Ethics (DPS.FVG.02.14) according to the Declaration of Helsinki.

Data Collection

The participants completed a single testing session (90 min) in a biomechanics laboratory. They were asked to carry out the testing session barefoot and dressed in short tights and t-shirts. Firstly, the participants filled out a questionnaire about their injury history and their usual physical activity–sports practice. After collecting their anthropometric features (height with the height scale Seca 213®, Germany; mass with the weight scale Tanita BC-601®, Japan), the general characteristics of the CS exercises were explained to the participants and they were encouraged to maintain the spine and pelvis in a neutral position ("as still as possible") during the exercise execution. Prior to the testing, participants completed a warm-up, which consisted of 10 repetitions of the following exercises: lumbopelvic mobility (i.e., pelvic circles, pelvic anti-versions and retroversions, and cat-camels), twisting crunches, side crunches, trunk extensions and free-weight squats.

During the testing session, participants performed seven variations of the front bridge, side bridge, back bridge and bird-dog exercises, for a total of 28 variations: (i) for the *front and side bridge* exercises (**Figure 1**): (1) short bridging, (2) long bridging, (3) bridging with single leg support, (4) bridging with double leg support on a hemisphere ball (54 × 24 cm; Medusa T1, Elksport®, Spain), (5) bridging with single leg support on a hemisphere ball, (6) bridging with double leg support on a fit ball (diameter: 45 cm; Amaya Sport, Spain), and (7) bridging with single leg support on a fit ball; (ii) for the *back bridge* exercise (**Figure 1**): (1) short bridge, (2) bridging with single leg support, (3) bridging with double leg support on a hemisphere ball, (4) bridging with single leg support on a hemisphere ball, (5) bridging with double leg support on a fit ball, (6) bridging with single leg support on a fit ball, and (7) bridging with single leg support and with the upper-back on a fit ball; and (iii) for the *bird-dog* exercise (**Figure 1**): (1) three-point position with an elevated leg, (2) three-point position with an elevated leg and the contralateral knee on a hemisphere ball, (3) classic two-point bird-dog position with elevated contralateral leg and arm, (4) two-point bird-dog position with the forearm on a hemisphere ball, (5) two-point bird-dog position with the knee on a hemisphere ball, (6) two-point bird-dog position with the forearm on a hemisphere ball while drawing squares in the air with the elevated limbs, and (7) two-point bird-dog position with the knee on a hemisphere ball while drawing squares in the air with the elevated limbs. The variations of these CS exercises were executed following less-to-more intensity order based on the information provided by a recent post-urographic study on CS exercise progressions (Vera-Garcia et al., 2020). Participants performed all the variations on a single leg with their preferred limb support. In addition, during the bird-dog variations in which participants drew squares in the air, a metronome (60 beats/min) was used to control the pace of the elevated limb motion (participants drew one side of the square every second).

Participants were asked to maintain the appropriate posture for 15 s. This trial duration was selected to avoid participant postural sway changes throughout the exercise because of fatigue. In addition, in order to avoid the large postural oscillations that usually appear at the beginning of the task, the acceleration recording started once the researcher verified that the participant was in the appropriate posture. Each CS exercise variation was performed on a mat (52 × 183 cm; McKinley Trekker M1.3, USA), resting 1 min between trials. Two expert researchers participated in the testing session. One of them controlled the exercise execution and asked the participants to rectify their position when necessary, while the other conducted the post-urography testing.

Instrumentation and Recording

During the CS exercise performance, pelvic linear accelerations were recorded to evaluate the lumbopelvic postural control challenge imposed on the participants as an index of exercise intensity. Pelvic accelerations were recorded at 200 samples/s from a 3-axis accelerometer (model LIS3DH, STMicroelectronics, Switzerland) embedded in a smartphone (Motorola Moto G, 2013, USA; Chipset Qualcomm MSM 8,226 Snapdragon 400; CPU Quad-core 1.2 GHz Cortex-A7; 1 GB RAM) using a free mobile application (Accelerometer Analyzer, Mobile Tools, Poland). To reduce accelerometer motions caused by muscle contractions, the smartphone was placed between the iliac crest and the great trochanter of the participants' dominant side (the support leg in the single-leg exercises) held in an elastic belt. To control the smartphone remotely and not interfere in the exercise execution, a free remote-control application installed on the smartphone and a laptop (TeamViewer Quick Support, TeamViewer, Germany) was employed.

Two video cameras (Sony Handycam HDR–XR260, Japan and Panasonic FZ200, Japan) were used to record a lateral and an oblique view of the participants performing each exercise variation. The cameras were set up on a tripod at a height of 150 cm above the ground for both views and separated 150 and 200 cm from the exercise mat for the lateral and the oblique view respectively.

Data Processing

The time series of acceleration data obtained from the accelerometer were filtered using a Butterworth digital filter (4th order, zero-phase lag, low-pass cut-off frequency of 10 Hz). The first second of each trial was discarded, selecting the following 12 s as signal window for the subsequent analyses. Pelvic linear acceleration was analyzed through the mean acceleration, which was calculated as the vector average magnitude in the three axes (Duarte et al., 2014). The acceleration data processing was carried out using a software specifically designed "*ad hoc*" by our research group in Lab View 9.0 environment (v9.0, National Instruments, Austin, Texas, USA).

The recorded videos were edited in a single 15 s long capture that combined the lateral and oblique view using the video editor Camtasia® (version 2020, Tech Smith Corporation, Okemos, Michigan, USA).

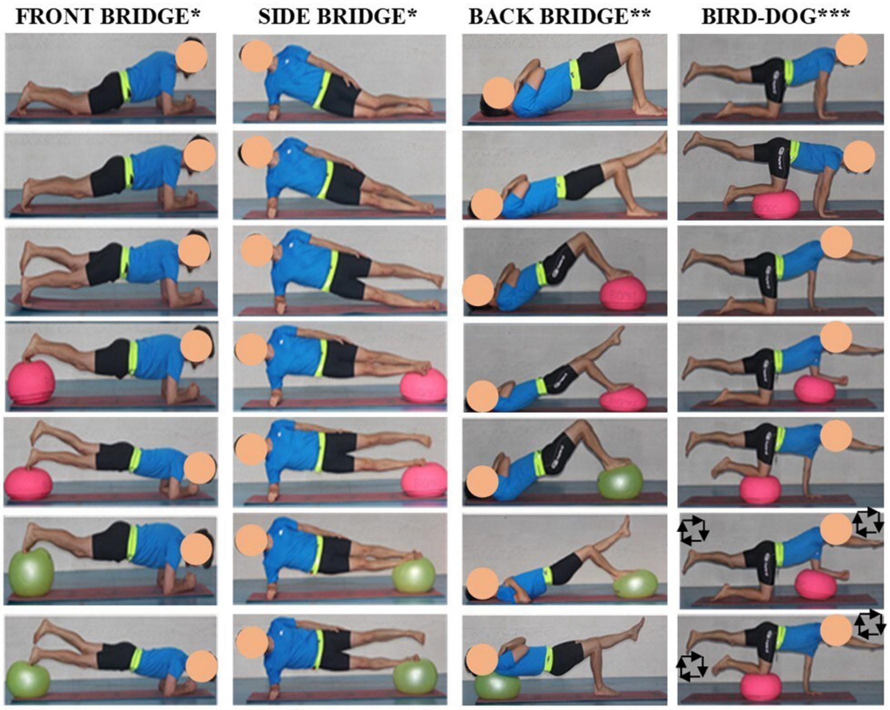

FIGURE 1 | Core stability exercises. *Variations of the *front* and *side bridge* exercises: (1) short front/side bridge; (2) long front/side bridge; (3) front/side bridge with single leg support; (4) front/side bridge with double leg support on a hemisphere ball; (5) front/side bridge with single leg support on a hemisphere ball; (6) front/side bridge with double leg support on a fit ball; (7) front/side bridge with single leg support on a fit ball; **Variations of the *back bridge* exercise: (1) short back bridge; (2) back bridge with single leg support; (3) back bridge with double leg support on a hemisphere ball; (4) back bridge with single leg support on a hemisphere ball; (5) back bridge with double leg support on a fit ball; (6) back bridge with single leg support on a fit ball; (7) back bridge with single leg support and with the upper-back on a fit ball; ***Variations of the *bird-dog* exercise: (1) three-point position with an elevated leg; (2) three-point position with an elevated leg and the contralateral knee on a hemisphere ball; (3) classic two-point bird-dog position with elevated contralateral leg and arm; (4) two-point bird-dog position with the forearm on a hemisphere ball; (5) two-point bird-dog position with the knee on a hemisphere ball; (6) two-point bird-dog position with the forearm on a hemisphere ball while drawing squares in the air with the elevated limbs; (7) two-point bird-dog position with the knee on a hemisphere ball while drawing squares in the air with the elevated limbs.

Rating Protocol

Once the videos were edited, they were jointly watched by two CS exercise experts (professors in Biomechanics at bachelor and post-graduate degrees) with more than 10 years of experience in designing, conducting and researching on CS exercise programs. The experts developed a set of observational screening guidelines based on their experience to decide for each exercise variation if it constituted an adequate training intensity level for the participant (*YES-Training level*) or not (*NO-Training level*). For an exercise variation to be rated as YES-Training level, the experts had to consider that *it clearly challenged the participant' CS*, and therefore one of the following criteria were met: (1) the participant showed some difficulty to maintain the head, trunk and limbs aligned and continuously lost and restored the aligned position; (2) the participant showed some difficulty to limit trunk movement (rotation, vibration, tremor, etc.) showing a moderate to high and continuous trunk oscillation around the position. On the other hand, for an exercise variation to be rated as NO-Training level, the experts had to consider that *it did not clearly challenge the participant's ability to maintain the lumbopelvic neutral position*, so both of the following criteria had to be met: (1) the participant maintained the head, trunk and limbs aligned with little or no difficulty; (2) the participant limited trunk movement (rotation, vibration, tremor, etc.) with little or no difficulty while maintaining the body posture. In addition, it was also considered as NO-training level when *the participant was not able to maintain the required position during the exercise variation* (i.e., it was too difficult). Experts were allowed multiple viewings, even pausing or rewinding each exercise progression. They could share their decisions and any discrepancies were discussed until an agreement was reached.

Subsequently, four non-expert raters (Ph.D. students) with 1–3 years of experience in CS training and researching (especially in CS exercises, but not in assessing the intensity of these exercises) attended a training session given by the expert raters. In this session, the non-expert raters watched several video examples of CS exercise progressions while they received feedback from the expert raters on how to decide if the exercise variations showed in the videos represented an adequate challenge/intensity level or not based on the abovementioned criteria. Considering that the lack of rater training standardization may reduce the rating reliability (Eastlack et al., 1991), all the non-expert raters received the same training, and were given an ample opportunity to practice with the exercise variations presented in the videos and ask questions to the expert raters. The non-expert raters were encouraged to rate as YES-Training level those exercise variations in which the *participants clearly showed difficulty maintaining the required posture*, watching the videos of all variations of a given exercise progression before deciding which ones represented an adequate training intensity level and which ones didn't. Besides, the non-expert raters were instructed to rate as YES-Training level all exercise variations that met the established criteria, regardless of the number of variations rated as YES-Training level.

After the training session, the non-expert raters watched the 10 participants' videos and used the experts' criteria to assess the participants' performance and to decide for each exercise variation if it constituted an adequate training intensity level for the participant or not. The same as the expert raters, non-expert raters were allowed multiple viewings, even to pause or rewind the exercise progressions, but they watched the videos and made the decisions alone. To evaluate the intra-rater agreement, the four non-expert raters reassessed the same videos 6 months later to reduce the likelihood that they remembered their previous evaluations.

Statistical Analysis

In order to analyze the inter-and intra-rater agreement, the standard and multirater Kappa (κ) coefficient, maximum Kappa and observer agreement (Po) and maximum Po were used. To avoid the bias when a higher prevalence of a category existed in the Kappa coefficient, the prevalence-adjusted bias-adjusted Kappa (PABAK) with its confidence limits (CL) was also calculated. The variations for each CS exercise (i.e., front bridge, back bridge, side bridge and bird-dog exercises) were analyzed as individual cases for inter-and intra-rater agreement calculations. Therefore, a total of 70 cases for each CS exercise (10 participants $\times$ 7 variations) were included in the analysis. The Kappa and PABAK coefficients were interpreted as: slight agreement (0.0–0.20), fair agreement (0.21–0.40), moderate agreement (0.41–0.60), substantial agreement (0.61–0.80), and almost perfect agreement (0.81–1.00), (Landis and Koch, 1977).

Regarding the acceleration data, the mean and standard deviation of the average pelvic accelerations were calculated for each participant and exercise variation in which she/he was able to maintain the required posture during the whole exercise. Subsequently, the Kolmogorov-Smirnov normality test with the Lilliefors correction was used to verify the normality of the data. Then, to analyze the differences in pelvic acceleration between the exercise variations rated by the experts as YES-Training level and NO-Training level, a one-way ANOVA was performed, being *training level* the between-subject factor (2 levels: YES-Training level and NO-Training level). Besides, to analyze the practical significance of the differences between exercise variations rated as YES-Training level and NO-Training level, the effect size was calculated using the statistical g of Hedge. The effect sizes were interpreted as: large ($\geq$0.8), moderate ($<$0.8–$\geq$0.5), small ($<$0.5–$\geq$0.2), and trivial ($<$0.2), (Cohen, 1992).

Finally, in order to explore if the smartphone accelerometry allowed to classify CS exercise variations as YES-Training level or NO-Training level, a receiver operating characteristic (ROC) curve was calculated for those exercise variations that showed differences between training levels (*YES-Training level* $\neq$ *NO-Training level*), linking the expert ratings with the average acceleration values obtained by the participants in each CS exercise variation (except those variations in which the participant was not able to maintain the required posture). The area under the ROC curve (AUC) was calculated by comparing it with the non-discrimination value (0.50). For the purposes of our study, acceleration cut-off points were chosen based on two methods. The first method aimed to maximize both sensitivity (Se) and 1-specificity (1-Sp) indexes (i.e., Youden Index) for each exercise variation with the condition that 1-Sp should be $<$16.7% (i.e., equivalent to one standard deviation). This method was used to reduce the bias caused by the inherent subjectivity of the different raters judging if a CS exercise is challenging or not for the participant. The second method aimed to minimize 1-Sp to remove all the false positives (i.e., exercise variations with acceleration scores over the selected threshold that were categorized as NO-Training level). This more restrictive method was used to ensure that all the exercise variations with acceleration scores over the selected threshold are considered as a sufficient training stimulus based on the experts' criteria, no matter which rater assessed the CS exercise performance. Considering that all bridging exercises (back, side and front bridges) showed similar acceleration scores, the ROC analysis was also applied for all of them together to obtain global pelvic acceleration thresholds.

All statistical analyses were carried out using the Statistical Package for Social Sciences package (SPSS, version 22.0, SPSS Inc., Chicago, IL, USA), establishing a significance level of $p < 0.05$.

RESULTS

As the inter-and intra-rater agreement results for each CS exercise were very similar, a composite value of all of exercises is presented in **Tables 1**, **2**. **Table 1** shows the inter-rater agreement values for the CS exercise variations rated as YES-Training level or NO-Training level based on the screening guidelines. The expert raters rated 61 CS exercise variations as YES-Training level and 219 CS exercise variations as NO-Training level. The observed agreement (Po) was high with values $\geq$81% in all cases and a value of 0.84 for multiraters*experts (Maximum Po = 0.98). The Kappa index ranged between 0.41 and 0.59 among the four non-expert raters and the experts with a multiraters*experts' value of 0.53 (Maximum Kappa = 0.93). The PABAK index was $\geq$0.62 among the four non-expert raters and experts and 0.69 (95% CL = 0.60–0.77) for multiraters*experts, which implies a "substantial" agreement.

Regarding the intra-rater agreement (**Table 2**), the observed agreement (Po) was high with values $>$80% for the four non-expert raters and a value of 0.87 for multiraters*experts (Maximum Po = 0.99). The four non-expert raters obtained a Kappa index $\geq$0.44 (Maximum Kappa $\geq$0.70) with a value of 0.61 for multiraters*experts. The PABAK index ranged between 0.59 and 0.80 for the four non-expert raters with a multiraters*experts'

TABLE 1 | Inter-rater agreement for the training level screening criteria between the 4 non-expert raters and the experts.

	Po	Max po	Kappa	Max kappa	PABAK (95% CL)
Rater 1/experts	0.85	0.98	0.56	0.92	0.71 (0.62–0.79)
Rater 2/experts	0.86	1.00	0.59	0.99	0.72 (0.64–0.80)
Rater 3/experts	0.81	0.96	0.41	0.88	0.62 (0.53–0.71)
Rater 4/experts	0.85	0.97	0.58	0.92	0.70 (0.62–0.78)
Multi-raters/experts	0.84	0.98	0.53	0.93	0.69 (0.60–0.77)

Po, Observed agreement; Max, Maximum; CL, Confidence limits; PABAK, Prevalence-adjusted bias-adjusted Kappa.

TABLE 2 | Intra-rater agreement for the training level screening criteria for the four non-expert raters.

	Po	Max po	Kappa	Max kappa	PABAK (95% CL)
Rater 1	0.90	1.00	0.68	1.00	0.80 (0.73–0.87)
Rater 2	0.90	0.99	0.67	0.70	0.79 (0.72–0.86)
Rater 3	0.90	0.96	0.61	0.85	0.79 (0.72–0.86)
Rater 4	0.80	1.00	0.44	0.99	0.59 (0.50–0.69)
Multi-rater	0.87	0.99	0.61	0.95	0.74 (0.67–0.82)

Po, Observed agreement; Max, Maximum; CL, Confidence limits; PABAK, Prevalence-adjusted bias-adjusted Kappa.

value of 0.74 (95% CL = 0.67–0.82), which also implies a "substantial" agreement.

The ANOVA showed differences between the exercise variations rated as YES-Training level and those rated as NO-Training level for all the CS exercises ($p \leq 0.001$, with a Hedge's $1.2 < g < 2.5$ effect size). The mean pelvic accelerations for the CS exercise variations rated as YES-Training level ranged from 0.32 to 0.48 m/s^2 (**Figure 2**), while the mean pelvic accelerations for the CS exercises rated as NO-Training level ranged from 0.17 to 0.26 m/s^2 (**Figure 2**). Regarding the ROC curve analysis using the Youden Index, the cut-off points for the four CS exercises were (**Figures 2**, **3**): bird-dog = 0.24 m/s^2 (AUC: 0.923; Sensitivity: 0.90; 1-Sp: 0.16); front bridge = 0.35 m/s^2 (AUC: 0.946; Se: 0.94; 1-Sp: 0.15); back bridge = 0.37 m/s^2 (AUC: 0.921; Se: 0.82; 1-Sp: 0.10); and side bridge = 0.35 m/s^2 (AUC: 0.931; Se: 0.93; 1-Sp: 0.15). In addition, the global cut-off point for the three bridging exercises was 0.35 m/s^2 (AUC: 0.912; Se: 0.87; 1-Sp: 0.12). On the other hand, the cut-off points for the four CS exercises when minimizing 1-Sp were (**Figures 2**, **3**): bird-dog = 0.32 m/s^2 (Se: 0.40; 1-Sp: 0.00); front bridge = 0.48 m/s^2 (Se: 0.44; 1-Sp: 0.00); back bridge = 0.50 m/s^2 (Se: 0.55; 1-Sp: 0.00); and side bridge = 0.49 m/s^2 (Se: 0.46; 1-Sp: 0.00). Furthermore, the global cut-off point for the three bridging exercises was 0.50 m/s^2 (Se: 0.31; 1-Sp: 0.00).

DISCUSSION

Considering the fact that the lack of training load control, mainly exercise intensity, is one of the main limitations of the CS training programs found in both, the scientific literature and the practical settings, this study: (i) provides new observational screening guidelines to decide if a CS exercise variation represents an adequate training intensity level for a given participant; (ii) analyzes the inter-and intra-rater agreement when using the screening guidelines; and (iii) establishes pelvic acceleration thresholds based on the relationships between expert raters' assessments and pelvic accelerations recorded with a smartphone accelerometer.

To the best of our knowledge this is the first study which has developed screening guidelines to try to establish the intensity of some of the most popular CS exercises. The expert and non-expert raters used these guidelines to select those CS exercise variations which in their opinion represented an intensity level that clearly challenged CS. The inter-rater analysis showed a substantial-high level of agreement between the expert and non-expert raters (PABAK = 0.69; 95% CL = 0.60–0.77), with a high percentage of observed agreement (Po = 84%; Maximum Po = 0.98). This high level of agreement indicates that, with a single training session

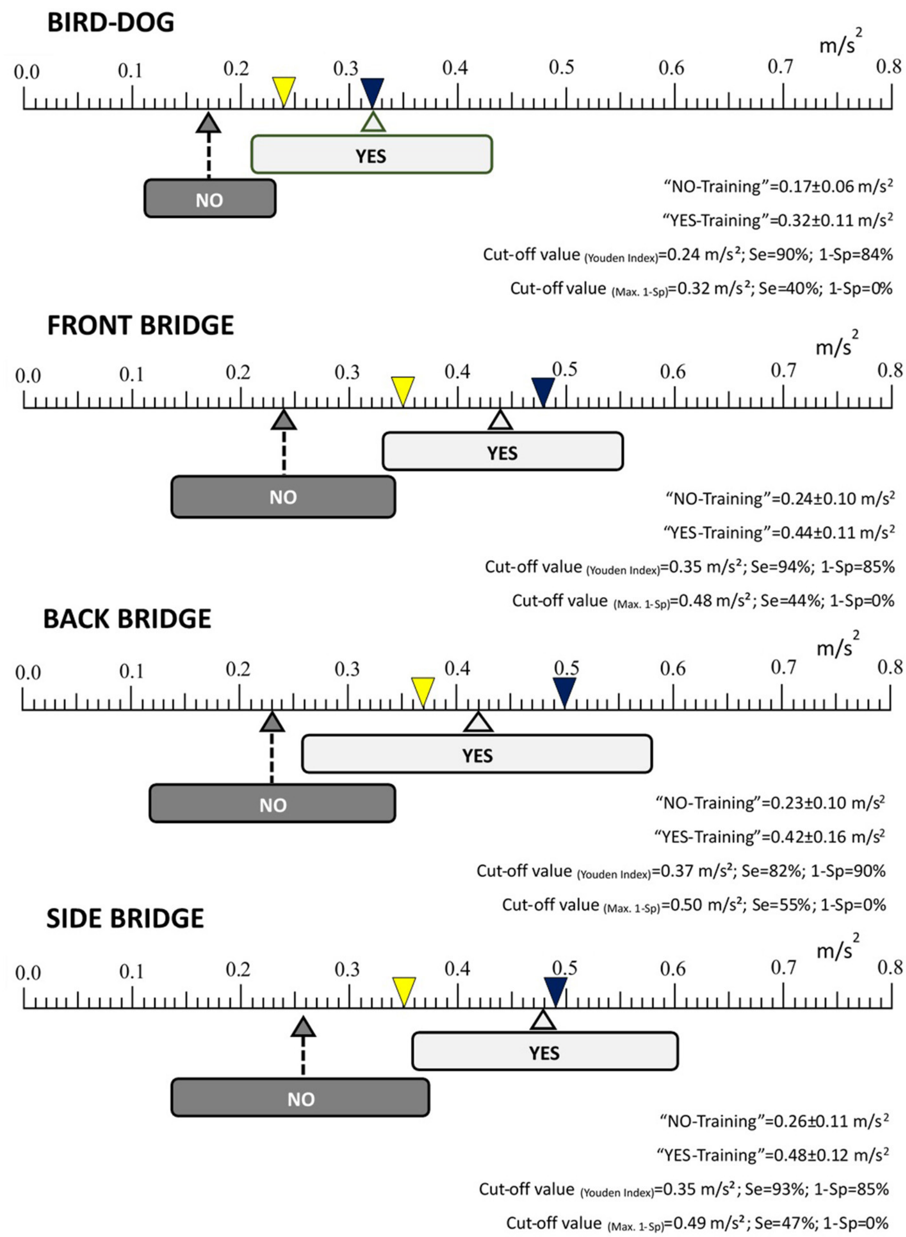

FIGURE 2 | Mean accelerations (±standard deviation) and cut-off values of the core stability exercise variations rated as "YES-Training level" and "NO-Training level". Se, Sensitivity; 1-Sp, Specificity. Each arrow points (on an acceleration/intensity scale ranging between 0 to 0.8 m/s^2) the average pelvic acceleration value of all the exercise variations rated by the experts as "YES-Training level" (light gray) or as "NO-Training level" (dark gray). The width of each light gray and dark gray rectangle represents the standard deviation. The yellow inverted triangle indicates the cut point using Youden Index while the blue inverted triangle indicates the cut point minimizing 1-Sp.

using the observational screening guidelines, the non-expert raters were able to make similar decisions to those of the expert raters about which exercise variations meant an adequate challenge/intensity level for a given participant. In addition, the data also showed a substantial-high level of intra-rater agreement (Po = 87%; Maximum Po = 0.98; PABAK = 0.74; 95% CL = 0.67–0.82), indicating that, after a period of 6 months, the non-expert raters still retained the rating skills developed in the training session at the beginning of the study.

The high level of inter-and intra-rater agreement and the fact that the observational screening guidelines are few and target different aspects of the CS stability exercise performance, i.e., body alignment and postural sway, lead us to believe that they can be easily applied by sport and health professionals. An important factor in enhancing the inter-and intra-rater

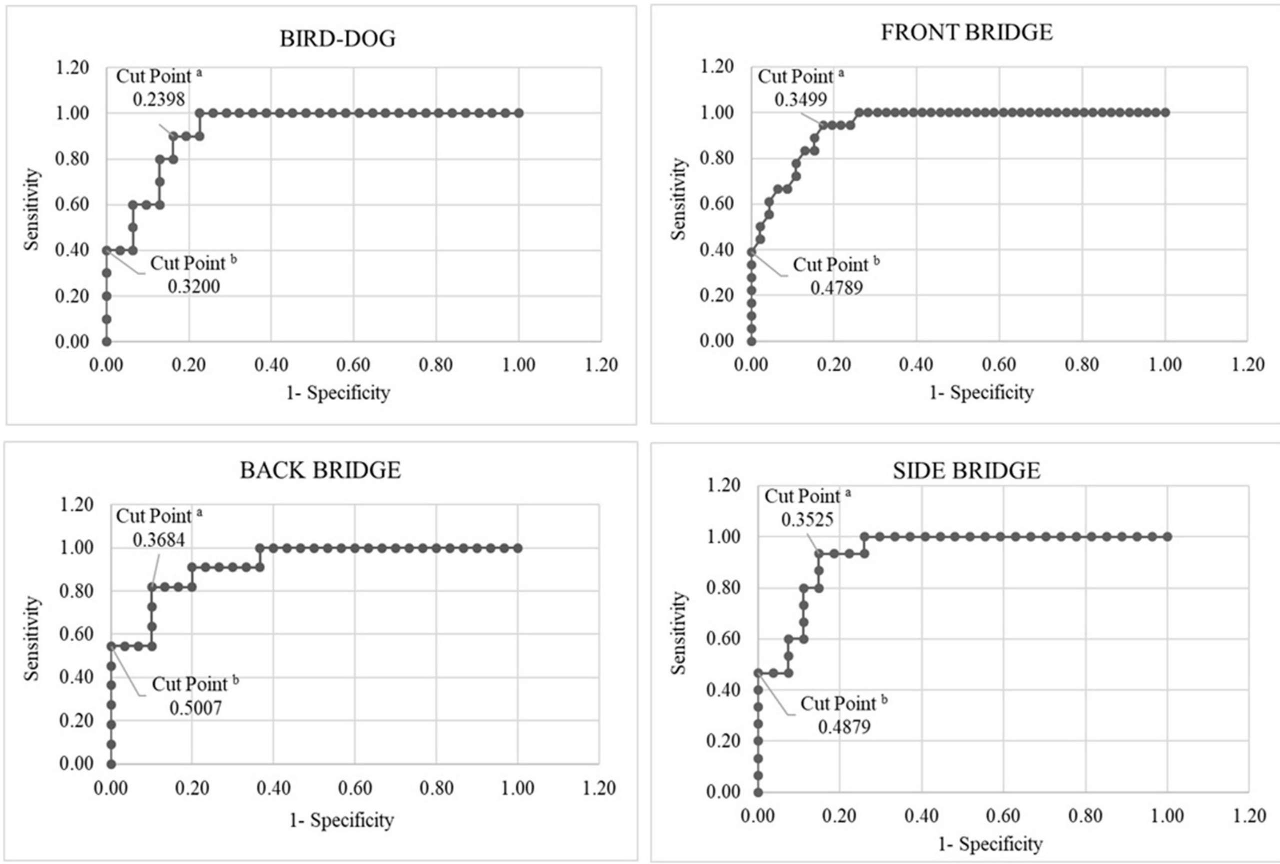

FIGURE 3 | Receiver operating characteristic curves linking the expert ratings with the pelvic accelerations obtained in each of the core stability exercises. [a]Cut point using Youden Index; [b]Cut point minimizing 1-Sp.

agreement were the characteristics of the methodology used in the training session. In this sense, the use of videos of CS exercise progressions with which to practice the selection of the most challenging exercise variations and the feedback from the expert raters on how they assess the participant's performance based on the screening guidelines was very useful. Four videos of a person performing CS exercise progressions with expert raters' feedback on CS exercise performance (**Supplementary Material 1**) and a table related to these videos with the expert rating for each exercise variation (as YES-Training level or NO-Training level), (**Supplementary Material 2**) are presented in the **Supplementary Material** to help those sport and health professionals interested in CS exercise design and prescription to use the criteria properly.

Although the screening guidelines provided in the current study could help people with little experience in CS exercise programs to make reliable assessments about the level of CS exercise intensity/difficulty, the correct performance of these assessments always depends on personal decisions. In order to increase the objectivity of these decisions, the use of smartphone accelerometers placed on the pelvis has been proposed to reliably quantify and control the CS exercise intensity (Barbado et al., 2018). However, as there is no information in the literature to interpret the pelvic accelerations during the CS exercises properly, ROC curves linking the expert ratings with the participants' pelvic accelerations were calculated in this study using two methods. As **Figures 2**, **3** show, the acceleration cut-off points of the back, side and front bridge exercises (which share analogous characteristics) based on the most conservative method, the Youden Index, were very similar (0.37, 0.35 and 0.35 m/s^2, respectively), with a global cut-off value of 0.35 m/s^2 and high values of Se (≥82%) and 1-Sp (≥15%)." On the other hand, the bird-dog exercise showed a lower cut-off point (0.21 m/s^2) with a 90% of Se and a 16% of 1-Sp. This lower cut-off point could be due to the fact that in the bird-dog exercise variations the pelvis has one or two points of support right below (i.e., the pelvis is supported by the lower limbs), while in the bridging exercises the support points are far from the pelvis, leaving the pelvis suspended in the air and leading to a higher oscillation. In addition, having an arm and a leg elevated during several bird-dog exercise variations might have made the body movements more easily noticeable, which could have influenced the expert's decisions. It must be pointed out that the acceleration thresholds based on the Youden index show high Se and 1-Sp values. Specifically, the high Se values observed for the bridging and the bird-dog exercises (≥82%) means that more than 82% of the selected exercise variations (those rated by the experts as YES-Training level) had a mean pelvic acceleration over the cut-off points. Besides, the high 1-Sp values of the CS exercises ($\leq$16%) imply that more than 84% of the exercise variations with a mean pelvic acceleration below the cut-off points were rated by the expert as NO-Training level.

Regarding the cut-off points based on minimizing the 1-Sp index, the acceleration cut-off values shown for the back, side and front bridge exercises were also very similar (0.50, 0.49 and 0.48 m/s^2, respectively), with a global cut-off value of 0.50 m/s^2. As occurred when using the Youden Index, the bird-dog exercise showed a lower cut-off point (0.32 m/s^2). These cut-off values, higher and more restrictive than those obtained using the Youden Index, mean that all the CS exercise variations that were rated as NO-Training level by the expert raters had acceleration values below them. Therefore, CS exercise variations showing acceleration scores above 0.50 m/s^2 could be considered as proper training stimulus according to the experts' ratings. The problem of choosing cut-off acceleration points minimizing 1-Sp is that the Se is low ($0.31 \leq Se \leq 0.55$), and thus, few exercise variations would be available to be used during a CS exercise program.

Based on the ROC curve results, the pelvic acceleration cut off points mentioned above may represent reference thresholds that could help select adequate training intensity levels for young, healthy and physically active individuals. From the authors' point of view, choosing cut-off acceleration points based on the Youden index or minimizing 1-Sp presents interesting practical implications. Although it has been proven that the conventional bridge and bird-dog variations do not impose high mechanical stress on the lumbar spine (Axler and McGill, 1997; Kavcic et al., 2004), acceleration thresholds based on the Youden index would be recommended when a training stimulus must be applied with the minimum possible level of mechanical stress (i.e., people without experience in CS exercises, with low levels of physical condition, with history of low back pain, etc.). Conversely, acceleration thresholds based on minimizing 1-Sp would be recommended when it is mandatory to ensure that a CS exercise imposes a sufficient training stimulus and the level of mechanical stress tolerance is high (i.e., athletes, people with experience in CS training, etc.).

Although further research is needed to explore the validity of these acceleration thresholds in the current and other populations, the objective data provided by the smartphone accelerometer could be used together with the observational screening guidelines to improve the decision-making process when establishing the intensity of bridging and bird-dog exercises. In relation to this, **Supplementary Material 2** shows the pelvic acceleration and the expert rating for each exercise variation presented in the example videos (**Supplementary Material 1**). Considering the mean acceleration values obtained in each exercise variation and the acceleration thresholds based on the Youden Index established in this study, some exercise variations that were not rated by the experts as YES-Training level (only based on the observational screening guidelines) could have been rated as YES-Training level if they had known the pelvic acceleration values. In this sense, smartphone accelerometry could be especially useful when the raters have doubts rating a CS exercise based on the screening guidelines, especially if they are not expert raters.

The main limitations of this study are the small sample size and the limited generalization of our results as our participants were young and relatively physically fit. Further research should include participants with different ages, spinal conditions, levels of training, etc. Nevertheless, the characteristics of the physical activities carried out by our participants (type, frequency, intensity, volume, etc.) were very heterogeneous, so the interpretation of our results could be applied to young people with different levels of physical fitness. Another limitation of the current study is that each exercise variation lasted only 15 s, so longer durations could have resulted in different acceleration cut-off points, as pelvic accelerations could change due to neuromuscular fatigue. As aforementioned, an exercise duration of 15 s was established because longer durations may have more impact on muscular endurance than on CS. This study also presents a technical limitation related to the generalization of our acceleration results to other devices. In this sense, although it is expected that the biological variations have a far more significant impact on the pelvic acceleration scores than the device noise, it is not clear how using other smartphones (and thus, other accelerometers) could affect the accuracy of the cut-off acceleration thresholds presented in this study. Finally, the acceleration thresholds were established based on the expert ratings rather than on data from experimental studies and therefore they should be interpreted with caution. Future randomized controlled trials should explore the effect of performing CS exercises at different intensity levels based on the acceleration cut-off points established in this study, which will allow to know the usefulness of these acceleration values to induce CS adaptations. In addition, performing CS interventions with different exercise intensities (i.e., pelvic accelerations) in combination with other training variables (e.g., sets, repetitions, exercise durations, etc.) could help to improve the dose-response characterization of the CS exercise programs.

CONCLUSIONS

To our knowledge, this is the first study that has developed observational screening guidelines to establish the intensity of bridging and bird-dog exercises, finding a substantial-high level of intra-and inter-rater agreement when using these criteria. In addition, ROC curves were performed with the aim of linking the CS exercise ratings based on the screening guidelines and the pelvic accelerations recorded with a smartphone accelerometer. The ROC curves showed global acceleration cut-off values which may represent the minimum training intensity levels for these exercises to produce CS adaptations in young physically active individuals, depending on whether a more restrictive (minimizing 1-Sp) or conservative criteria (Youden Index) is used. Therefore, this study provides new observational screening guidelines (targeting body alignment and postural sway while performing CS exercises) and acceleration thresholds based on smartphone accelerometry to facilitate the decision-making process when setting the intensity of bridging and bird-dog exercises in this population.

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 Miguel Hernández University Office for Research Ethics (DPS.FVG.02.14) according to the Declaration of Helsinki. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

All authors listed have made a substantial, direct and intellectual contribution to the work, and approved it for publication.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2021.751569/full#supplementary-material

Supplementary Table 1 | Didactic material of experts' assessment and mean pelvic acceleration for each variation of the core stability exercises showed in the videos presented in the **Supplementary Material 1**.

Supplementary Video 1 | Participant performing a progression of seven bird-dog exercise variations with experts' comments on the participant's performance.

Supplementary Video 2 | Participant performing a progression of seven front bridge exercise variations with experts' comments on the participant's performance.

Supplementary Video 3 | Participant performing a progression of seven side bridge exercise variations with experts' comments on the participant's performance.

Supplementary Video 4 | Participant performing a progression of seven back bridge exercise variations with experts' comments on the participant's performance.

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BRIEF RESEARCH REPORT
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Acute Exposure to Normobaric Hypoxia Impairs Balance Performance in Sub-elite but Not Elite Basketball Players

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Although high and simulated altitude training has become an increasingly popular training method, no study has investigated the influence of acute hypoxic exposure on balance in team-sport athletes. Therefore, the purpose of this study was to investigate whether acute exposure to normobaric hypoxia is detrimental to balance performance in highly-trained basketball players. Nine elite and nine sub-elite male basketball players participated in a randomized, single-blinded, cross-over study. Subjects performed repeated trials of a single-leg balance test (SLBT) in an altitude chamber in normoxia (NOR; approximately sea level) with FiO_2 20.9% and PiO_2 ranging from 146.7 to 150.4 mmHg and in normobaric hypoxia (HYP; ∼3,800 m above sea level) with FiO_2 13.0% and PiO_2 ranging from 90.9 to 94.6 mmHg. The SLBT was performed three times: 15 min after entering the environmental chamber in NOR or HYP, then two times more interspersed by 3-min rest. Peripheral oxygen saturation (SpO_2) and heart rate (HR) were recorded at four time points: after the initial 15-min rest inside the chamber and immediately after each SLBT. Across the cohort, the balance performance was 7.1% better during NOR than HYP ($P < 0.01$, $\eta_p^2 = 0.58$). However, the performance of the elite group was not impaired by HYP, whereas the sub-elite group performed worse in the HYP condition on both legs (DL: $P = 0.02$, $d = 1.23$; NDL: $P = 0.01$, $d = 1.43$). SpO_2 was lower in HYP than NOR ($P < 0.001$, $\eta_p^2 = 0.99$) with a significant decline over time during HYP. HR was higher in HYP than NOR ($P = 0.04$, $\eta_p^2 = 0.25$) with a significant increase over time. Acute exposure to normobaric hypoxia detrimentally affected the balance performance in sub-elite but not elite basketball players.

Keywords: postural control, high altitude training, oxygen saturation, team sports, single-leg balance test

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INTRODUCTION

Field- and court-based sports require players to rapidly accelerate, decelerate, and change direction during matches, in order to perform repeated sprints, shuffles, and jumps (Abdelkrim et al., 2010; Pojskic et al., 2018). The resulting physiological and mechanical loading results in fatigue that detrimentally affects postural and motor control, which in turn leads to poorer balance, impaired athletic performance, and increased injury risk (Myklebust et al., 1998; Paillard, 2012; Ruzic et al., 2014). One could, therefore, suggest that balance is a critical determinant of athlete performance in field-based (e.g., football and rugby) and court-based (e.g., basketball and handball) team sports.

In addition to fatigue, several additional factors may negatively influence balance, including sensory alternation, dehydration, and exposure to acute hypoxia (Hoshikawa et al., 2010; Kraemer et al., 2011; Paillard, 2012; Ruzic et al., 2014). It has been well-documented that exposure to moderate and high altitude detrimentally affects various aspects of physical performance in team sports, including repeated sprint ability, running endurance, distance covered, and number of high-intensity sprints during matches (Hamlin et al., 2008; Valenzuela et al., 2019). Moreover, it is well-known that postural control and balance are altered during hypoxic exposure (Degache et al., 2020).

However, the majority of studies exploring the effect of hypoxia on balance have been conducted on aircraft crew members, mountaineers, or healthy active subjects in field-based and laboratory-based settings (Nordahl et al., 1998; Degache et al., 2020). Negative effects of hypobaric hypoxia on balance ability and postural control have been demonstrated in these populations at high altitudes (>3,500 m) (Stadelmann et al., 2015; Bruyneel et al., 2017) and in normobaric hypoxia (Nordahl et al., 1998; Cymerman et al., 2001; Wagner et al., 2011; Drum et al., 2016). Given that high-altitude training has become an increasingly popular training method for team sports (Girard et al., 2013) and that some national, professional, and University teams in ball and court sports (e.g., basketball, baseball, soccer, and American football) may train at moderate (2,300–3,500 m) and high (3,500–5,500 m) altitudes (Kraemer et al., 2011; Gore et al., 2013), improving understanding of the extent to which hypoxic exposure affects balance in team sports is pertinent.

Although it is well-known that balance is more altered in hypobaric than normobaric hypoxia (∼1,700–3,000 m above sea level) (Degache et al., 2020), it is also of practical interest to investigate whether exposure to normobaric hypoxia at a simulated altitude of ∼3,800 m above sea level would negatively affect balance in well-trained team-sport athletes. In other words, it might reveal a potential adaptation stimulus that could be included in balance training and facilitated by high-altitude chambers. Furthermore, evidence is lacking regarding the influence of player performance level, which has been shown to modulate balance performance both in normoxia (Paillard et al., 2011; Pojskic et al., 2020) and in mountain climbers at high altitudes (3,200 m; Bruyneel et al., 2017).

Moreover, several studies examining single-leg balance performance in normoxia using an omni-axial balancing board have reported good relative and absolute reliability of measurements (e.g., ICC > 0.70 and CV% < 10%; Wojtyczek et al., 2014; Hildebrandt et al., 2015; Pojskic et al., 2020). However, there is a lack of studies that have reported reliability data when testing balance in hypoxia using an omni-axial balancing board. Given that an appropriate measurement reliability is important to detect any changes in test performance (Hopkins, 2004; Pojskic et al., 2020), it is important that reliability is also determined in hypoxic conditions.

To the best of our knowledge, no study has investigated the influence of acute hypoxic exposure on balance ability in highly trained athletes active in team sports demanding high levels of dynamic balance and agility. Therefore, the purpose of this study was to investigate whether an acute exposure to normobaric hypoxia would be detrimental to balance performance in elite and sub-elite basketball players. Secondary aims were to investigate whether player performance level modulated balance performance in hypoxia and whether the variability of repeated single-leg balance tests (SLBTs) was greater in hypoxia than normoxia. We expected that an acute exposure to normobaric hypoxia would be less detrimental to balance in the elite than the sub-elite group (Paillard et al., 2011; Bruyneel et al., 2017; Pojskic et al., 2020). Furthermore, it was hypothesized that the balance test would be a valid, reliable, and useful testing tool in assessing balance in basketball players in both conditions (Hildebrandt et al., 2015; Pojskic et al., 2020).

METHODS

Subjects

The study included 18 highly trained, national-level, male basketball players: nine players who competed at the national senior level (elite group) and nine players (sub-elite group) who competed in U18 national competitions (**Table 1**). No participants had reported a history of neuromuscular disease or injuries in the previous 6 months, and participants had not undertaken specific balance or high-altitude training in the previous 6 months. Two days prior to the experimental visits, subjects were asked to avoid sleep deprivation, to refrain from high-intensity training, and to avoid tobacco, alcohol, and caffeine. To maintain adequate hydration, players were allowed to drink water *ad libitum* in each experimental condition. The study was approved by the Regional Ethical Review Board (2016-456-31). All participants were informed of the purpose, benefits, and risks of the investigation before providing written informed consent to participate. Parents or guardians provided informed consent for participants under 18 years of age.

Study Design

The study was conducted with a randomized, single-blind, cross-over design. To assess the influence of performance level on balance performance in hypoxia, participants were dichotomized into two sub-groups (elite vs. sub-elite). Testing took place over three sessions: a familiarization session and two experimental conditions during which participants performed a test protocol consisting of several attempts of an SLBT on each leg, in a 50 m^3 altitude chamber (Hypoxico, New York, USA) in normoxia [NOR, ∼ sea level] and normobaric hypoxia (HYP: ∼3,800 m above sea level, a.s.l, **Figure 1**). The chamber delivered 1,500 L/min dehumidified air with FiO_2 13.0% and PiO_2 ranging from 90.9 to 94.6 mmHg in HYP and FiO_2 20.9% and PiO_2 ranging from 146.7 to 150.4 mmHg in NOR. In HYP, FiO_2 was checked at the beginning and end of each trial; a tolerance of 0.3% FiO_2 was deemed acceptable performance. Ambient pressure was

TABLE 1 | Descriptive characteristics of the elite (*N* = 9) and sub-elite (*N* = 9) basketball players.

Variables	Total	Elite	Sub-elite		
	Mean ± SD	Mean ± SD	Mean ± SD	ES (Cohen's *d*)	CI 95% of Cohen's *d*
Age (years)	19.5 ± 2.9	21.8 ± 2.5	17.2 ± 0.4*	2.54	1.25 to 3.80
Playing experience (years)	6.9 ± 2.1	8.7 ± 1.6	5.2 ± 0.4*	2.83	1.47 to 4.16
Body weight (kg)	83.9 ± 9.2	84.5 ± 10.2	83.4 ± 8.6	0.11	−0.81 to 1.03
Body height (m)	1.89 ± 0.08	1.89 ± 0.09	1.89 ± 0.08	0.00	−0.05 to 0.05
Body mass index (kg/m^2)	23.3 ± 1.5	23.4 ± 1.5	23.2 ± 1.6	0.15	−0.77 to 1.07
Systolic blood pressure (mmHg)	131 ± 10	129 ± 11	132 ± 8	0.32	−0.61 to 1.24
Diastolic blood pressure (mmHg)	78 ± 7	80 ± 7	76 ± 8	0.56	−0.39 to 0.49

ES, effect size; CI 95%, 95% confidence interval; T1, collected immediately before entering the chamber; T2, collected after exposure inside the chamber; T3, collected after 3-min rest outside the chamber.
**Significantly different from the high-level group. $P < 0.05$.*

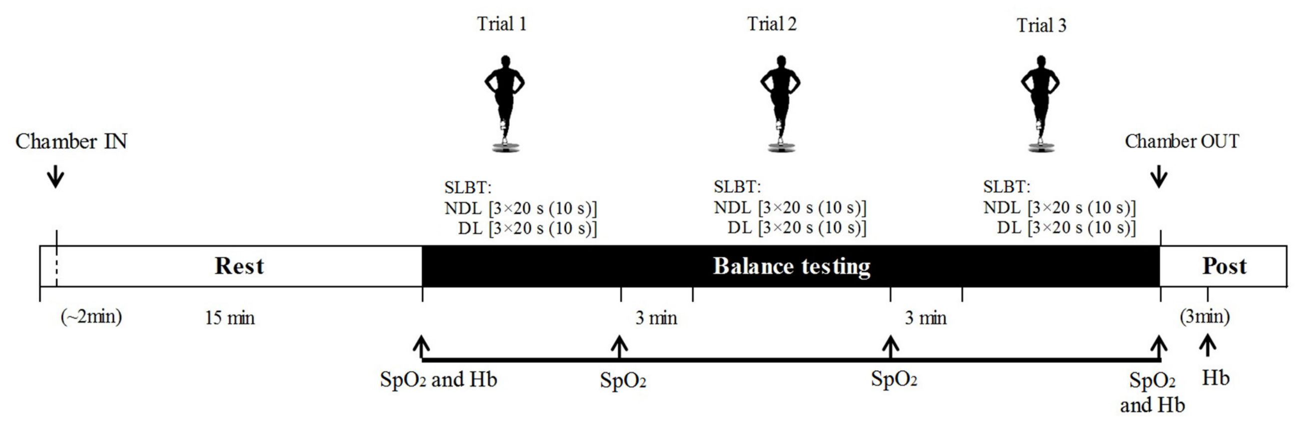

FIGURE 1 | Testing protocol indicating single-leg balance tests, oxygen saturation (SpO_2), and heart rate (HR) assessment in both conditions: normobaric hypoxia (HYP) and normoxia (NOR). Hemoglobin (Hb) was assessed only in HYP. For balance trials, number of repetitions, repetition duration, and resting periods are represented as follows: leg [repetitions × repetition duration (inter-set recovery)]. DL, dominant leg; NDL, non-dominant leg.

not significantly different between HYP and NOR trials ($P = 0.21$). The temperature controller was set to 20°C for all trials. Good relative reliability (ICC > 0.70) has been established for SLBTs performed on omni-axial balance boards in normoxic conditions (Hildebrandt et al., 2015; Pojskic et al., 2020). Given that this study employed normobaric hypoxia, the reliability of the test equipment would not be expected to differ in the hypoxic condition, although the reliability of execution of the test by participants may be affected, and was therefore investigated as a secondary aim.

Experimental visits were separated by at least 48 h and took place during the competitive season. To avoid anticipatory effects, subjects were blinded to the environmental conditions in the chamber (i.e., all digital panels were concealed from view) and the purpose of measuring various physiological responses. We chose the simulated high altitude to be over 3,000 m a.s.l, as it is the altitude threshold at which reductions in SpO_2 start to detrimentally affect postural control and perceptual-motor performance (Fowler et al., 1987). This was to ensure that the hypoxic stimulus would be sufficient to affect balance performance in basketball players, presumed to have well-developed balance abilities (Curtolo et al., 2017).

Testing Protocol

On the familiarization day, body weight, height, blood pressure, and resting heart rate (HR) of subjects were measured (**Table 1**). After that, the subjects were familiarized to the SLBT. On the second and third sessions, subjects performed a standardized balance test protocol 15 min after entering the environmental chamber either in the NOR or HYP condition. The test protocol consisted of three trials interspersed by 3-min rest. Each trial comprised three 20 s attempts of the SLBT with 10 s recovery, performed first on the dominant (DL) followed by the non-dominant (NDL) leg (**Figure 1**). Special attention was paid to ensure an accurate foot placement on the balance platform and maintain a stable body posture and hand positioning. The participants were instructed to refrain from any other breathing methods such as hyperventilation.

Single-Leg Balance Test

The SLBT employed a single-leg upright stance on an omni-axial balance board and was chosen to provide unstable, dynamic conditions to test balance ability requiring complex rearrangement of postural control (Valle et al., 2015; Petró et al., 2017; Pojskic et al., 2020). We assumed that this complexity would ensure discriminative power under different conditions (i.e., NOR vs. HYP) in balance-proficient athletes playing at different levels. The SLBT was executed using an MFT challenge disk system (TST Trendsport, Grosshöflein, Austria), which assesses the ability of an individual to maintain stability on a multiaxial tilting platform and produces a "balance index" ranging from 1 to 5, where a lower index represents less deviation from the horizontal plane and thus better balance. The procedure for the SLBT has been reported in detail elsewhere (Pojskic et al., 2020). Briefly, the DL of subjects was determined based on which leg a subject uses to kick a ball (van Melick et al., 2017). Then, without shoes, subjects stood on one leg in the middle of the balance plate with a slightly flexed knee, kept their hands on their waist, and were instructed to keep the platform in a horizontal position. The SLBT was performed for 20 s, but restarted if subjects made two "mistakes" (such as touch the floor or balance plate with the free foot). In addition, if only one mistake was made within a 20-s attempt, then the performance was penalized by adding 0.2 of the balance index to the produced result. The mean scores of the three attempts for each of the three trials were used to calculate test–retest reliability coefficients. To analyze the differences between experimental conditions, the mean score of the three trials was used.

Pulse Oximetry, Heart Rate, and Hemoglobin Monitoring

During the two experimental conditions, SpO_2 and HR were assessed using wireless pulse oximeters (Wristox 3100; Nonin, Plymouth, MN, USA) placed on the middle finger and recorded manually at four time points: after the initial 15 min rest inside the chamber and then immediately after each SLBT trial. The display on the oximeters was covered to assist with blinding. Hb samples were collected only in HYP at three time points: (a) immediately before entering the chamber; (b) after exposure inside the chamber; and (c) after 3 min rest outside the chamber. Capillary blood was sampled using a 21G safety lancet (Sarstedt, Nuembrecht, Germany) collected into microcuvettes, and Hb was analyzed in triplicate using a Hb 201 Analyzer (H31216, Hemocue, Ängelholm, Sweden).

Statistical Analyses

The sample size was estimated *a priori* using previously published SLBT index means and standard deviations (SDs; Hildebrandt et al., 2015; Pojskic et al., 2020). Using G-Power software (version 3.1.9.2; Heinrich Heine University Dusseldorf, Dusseldorf, Germany), we estimated that nine subjects (df = 8) would provide an appropriate sample size for paired-samples differences ($P \leq 0.05$, power = 0.90).

Data are presented as means and SDs. Normality was assessed using the Shapiro-Wilk test. A four-factor mixed ANOVA with three within-subject factors [condition (NOR vs. HYP), leg used (DL vs. NDL), and SLBT trials (1–3)] and one between-subject factor (elite vs. sub-elite) was used to analyze effects on balance. A three-factor ANOVA was used to evaluate the effect of group, condition, and trial on SpO_2 and HR. Bonferroni *post-hoc* tests were used for pairwise comparisons. Partial eta squared (η_p^2) was calculated for the ANOVA main effects, with effect sizes interpreted as follows: >0.02 (small), >0.13 (medium), and >0.26 (large). To investigate differences in balance between playing levels and between NOR and HYP, *t*-tests for independent and dependent samples were used. Cohen's *d* effect sizes were also calculated and interpreted as follows: <0.2 (trivial), 0.2–0.6 (small), 0.6–1.2 (moderate), 1.2–2.0 (large), and >2.0 (very large) differences (Cohen, 1988; Hopkins, 2000). Analysis of relative reliability was performed for all balance tests by calculating the intraclass correlation coefficient (ICC, model 3.1). The absolute reliability (within-subject variation) was established using the coefficient of variation (CV%) expressed in percentage (Hopkins, 2000). Usefulness was computed by comparing typical error (TE) and the smallest worthwhile change (SWC) of the balance index (Hopkins, 2000). SWC was derived from between-subject SD multiplied by 0.2 ($SWC_{0.2}$; small effect) or 0.5 ($SWC_{0.5}$; moderate effect) (Cohen, 1988). A TE below SWC indicates test usefulness as "good," and TE similar to SWC is rated "acceptable." If TE is higher than SWC, it is deemed to have "marginal" usefulness (Hopkins, 2004). Statistical significance for all tests was set at $P \leq 0.05$. Statistical analyses were performed using SPSS® 24.0 (IBM, New York, USA).

RESULTS

Reliability of Balance Tests

The relative reliability for the SLBT was high in each condition (DL and NDL, NOR and HYP), with ICC ranging from 0.73 to 0.90. The absolute reliability expressed as CV% for the SLBTs was 6.95% and 7.58% in NOR and 8.99% and 6.59% in HYP

TABLE 2 | Balance index scores for the high and low playing level groups in NOR and HYP condition.

Condition	Total	Elite	Sub-elite	Between-subjects differences		Within-subjects differences	
						Elite	Sub-elite
	Mean ± SD	Mean ± SD	Mean ± SD	ES (Cohen's *d*)	CI 95% of Cohen's *d*	ES (Cohen's *d*)	ES (Cohen's *d*)
NOR-DL	3.34 ± 0.36	3.22 ± 0.39	3.47 ± 0.30†	0.71	−0.25 to 1.66	0.71	1.65
HYP-DL	3.65 ± 0.38	3.45 ± 0.35	3.86 ± 0.31*	1.23	0.20 to 2.23		
NOR-NDL	3.45 ± 0.51	3.29 ± 0.46	3.61 ± 0.52†	0.64	−0.31 to 1.58	0.12	0.80
HYP-NDL	3.59 ± 0.48	3.31 ± 0.47	3.88 ± 0.29*	1.44	0.37 to 2.47		

SLBT, single-leg balance test; ES, effect size; CI 95%, 95% confidence interval. Mean ± SD (range) of balance index (performance) across the three trials in both normoxia (NOR) and normobaric hypoxia (HYP) for the dominant (DL) and non-dominant (NDL) legs.

**Significantly different from the elite group.*

†Significantly different from HYP condition. P < 0.05.

condition, for DL and NDL, respectively. In both the NOR and HYP conditions, the TE exceeded $SWC_{(0.2)}$, whereas it was below the $SWC_{(0.5)}$ for all balance tests (see **Supplementary Table 1**). No significant differences in performance between legs or within trials were detected in any condition for any group (**Table 2**).

Effects of Hypoxia on Balance Performance

Balance performance was 7.1% better during NOR (3.40 ± 0.43) than HYP (3.62 ± 0.44) ($F = 22.3$, $P = 0.00$, $\eta_p^2 = 0.58$; **Figure 2**). The balance performance was also 10.4% better in the elite (3.32 ± 0.41) than the sub-elite group (3.70 ± 0.37) ($F = 4.85$, $P = 0.04$, $\eta_p^2 = 0.23$; **Figure 2**). Further exploration revealed that only the elite group performed better than the sub-elite group in the HYP condition, on both DL [$t_{(16)} = 2.62$, $P = 0.019$, $d = 1.23$] and NDL [$t_{(16)} = 2.78$, $P = 0.01$, $d = 1.44$] (**Table 2**).

In the elite group, no differences in balance were detected between NOR and HYP, for DL [$t_{(8)} = 2.15$, $P = 0.06$, $d = 0.71$] or NDL [$t_{(8)} = 0.37$, $P = 0.72$, $d = 0.12$]. Conversely, the sub-elite group had poorer balance on DL [$t_{(8)} = 4.96$, $P < 0.00$, $d = 1.65$] and NDL [$t_{(8)} = 2.40$, $P = 0.04$, $d = 0.80$] in HYP than NOR (**Table 2**).

Pulse Oximetry, Heart Rate, and Hemoglobin Concentration

The SpO_2 was lower in HYP than NOR ($F = 14.9$, $P < 0.00$, $\eta_p^2 = 0.99$; **Supplementary Figure 1A**) with a mean difference of ∼15% [95% CI (14.0–15.7)]. However, the group × condition interaction revealed lower SpO_2 in elite compared with sub-elite players in HYP ($F = 7.26$, $P = 0.02$, $\eta_p^2 = 0.31$). Regardless of condition and group affiliation, SpO_2 decreased across the sampling time points ($F = 3.98$, $P = 0.01$, $\eta_p^2 = 0.20$, **Supplementary Figure 1A**), but without significant pairwise differences between them. The HR response was higher in HYP than NOR ($F = 4.85$, $P = 0.04$, $\eta_p^2 = 0.25$, **Supplementary Figure 1B**) with a mean difference of ∼6 beats/min [95% CI (0.2–11.4)]. Significant differences in HR were found 15 min after entering the chamber compared with the first, second, and third trials in both conditions (**Supplementary Figure 1B**). There were no differences in HR response between the groups ($P = 0.64$). No significant changes were found in Hb concentration between sampling time points in HYP (pre: 146 ± 9 g × L^{-1}, after: 148 ± 9 g × L^{-1}, and 3 min post-recovery: 147 ± 10 g × L^{-1}; $F = 1.45$, $P = 0.26$, $\eta_p^2 = 0.16$). No significant interaction between the group and condition was found for Hb ($F = 1.98$, $P = 0.17$, $\eta_p^2 = 0.21$).

DISCUSSION

This is the first study that has investigated the effects of acute exposure to high altitude on single-leg balance performance in highly trained basketball players. The main findings of this study are that: (a) acute exposure to normobaric hypoxia equivalent to ∼3,800 m a.s.l. impaired balance performance in sub-elite but not elite basketball players and (b) the SLBT showed good reliability in normobaric hypoxia that was comparable with the reliability in normoxia.

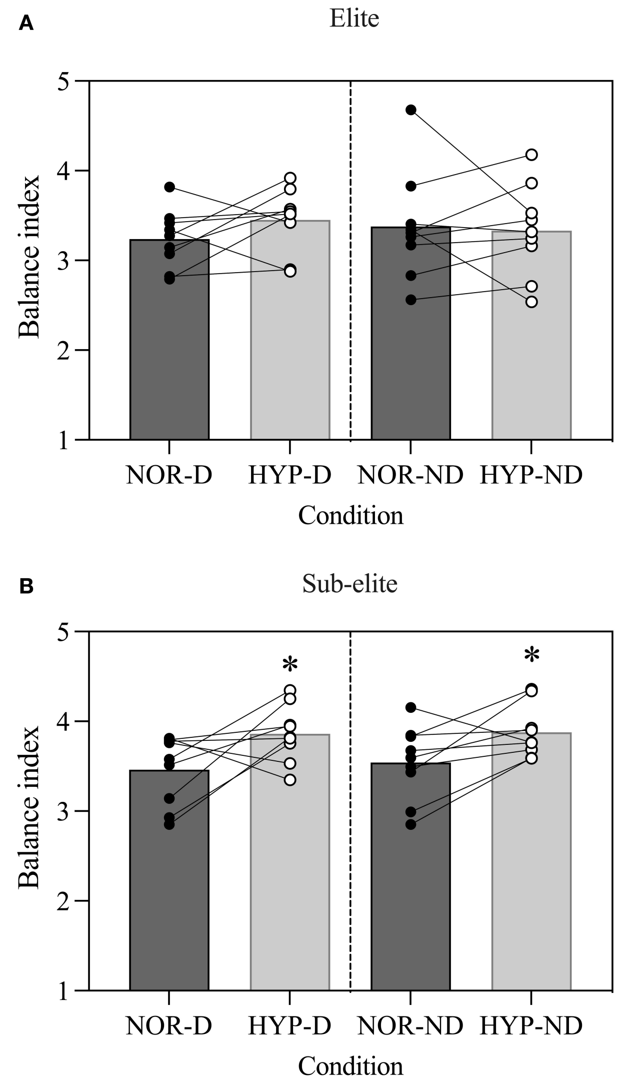

FIGURE 2 | Acute effects of normobaric hypoxia on a single leg balance test performance in **(A)** elite and **(B)** sub-elite basketball players. NOR, normoxic condition; HYP, normobaric hypoxic condition; D, dominant leg; ND, non-dominant leg. *Values significantly different from those obtained in NOR; $P < 0.05$.

Effects of the Exposure on Balance Performance

In this study, normobaric hypoxia detrimentally affected the balance (∼7%), when subjects were exposed to a simulated altitude of ∼3,800 m. This is in line with previous studies that have demonstrated elements of impaired balance, such as postural stability (Cymerman et al., 2001; Degache et al., 2020) and body sway (Nordahl et al., 1998; Wagner et al., 2011) in simulated altitude (2,400 m to 5,500 m) vs. normoxia, with exposure durations of a few minutes to 24 h in non-athletic populations. However, this study corroborates and extends these findings by revealing detrimental effects of HYP on balance even in highly trained basketball players who engage in activities demanding high levels of balance. It is well-known that hypoxia can negatively affect the sensory (e.g., the vestibular, proprioceptive, and visual) and central nervous systems, which are eminently sensitive to decrements in tissue oxygenation (Pickard and Gradwell, 2008) leading to impaired neuromuscular coordination, postural stability, and balance (Degache et al., 2020).

Interestingly, this study showed that the detrimental effect of HYP on balance was modulated by the playing level of subjects, with those playing at a higher level having better balance in hypoxia. Previous studies conducted in other sports (e.g., curling, football, and surfing) in normoxia showed advanced balance in athletes competing at higher playing levels, suggesting that balance may be sensitive to training status, playing experience, and/or sport- and movement-specific adaptations (Paillard et al., 2011; Pojskic et al., 2020). In this study, given that there was no difference in balance between HYP and NOR in the elite group, we could assume that both longer and advanced basketball training and competition might have induced advanced adaptation of the neuromuscular system, which in turn was less susceptible to the adverse effects of the HYP condition (Bruyneel et al., 2017). Although there was a significant age difference between the elite (21.8 ± 2.5 years) and sub-elite (17.2 ± 0.4 years) groups, we cannot consider a difference in the maturation of the sensory systems as a crucial factor contributing to advanced balance performance in the higher level group. We reason this based on two arguments: firstly, because there was no difference between groups in SLBT performance in NOR condition, and secondly, because the visual and vestibular afferent systems, that are responsible for providing accurate sensory inputs and maintenance of postural control, reach adult level at 15–16 years of age (Hirabayashi and Iwasaki, 1995; Steindl et al., 2006).

We can speculate that the elite group may have been better able to compensate for the additional demands on balance performance in hypoxia, potentially by reducing spinal reflex activity and the excitability of the spinal α-motoneurons, that are increased by reduced oxygen delivery to target tissues (Lundby et al., 2009). As a result, reduced activation of the muscles encompassing the joints (e.g., the ankles and knees) may inherently prevent unwanted and uncontrollable joint oscillations when performing the SLBT (Keller et al., 2012). Additionally, considering that vision is one of the first senses to be affected by acute hypoxia (Nordahl et al., 1998; Degache et al., 2020) and that the SLBT was performed with open eyes, the elite group may have been able to compensate for reduced vision capacity through greater contributions from vestibular and proprioceptive sensory information (Paillard et al., 2002, 2011; Pojskic et al., 2020). In brief, it is well-documented that the role of the visual system on balance performance decreases with higher playing level in athletes (Paillard and Noé, 2006; Paillard et al., 2006). Given that successful basketball game play requires players to control the ball and move down the court without watching it while observing the situation in the game (e.g., movement of teammates and opponent), it means that elite players, as a result of long-term adaptation to basketball training, became less dependent on vision and more on their proprioception when maintaining balance (Paillard, 2019).

In contrast, less experienced athletes rely more on vision for maintaining balance (Paillard et al., 2002, 2011; Paillard and Noé, 2006), and consequently, their balance may have been more negatively affected by reduced SpO_2 (Bruyneel et al., 2017). This can be supported by the fact that sub-elite players experienced a bigger deterioration in balance performance despite displaying milder reductions in SpO_2 than their elite counterparts. Moreover, sub-elite players had higher HR across all trials which might indicate a higher cardio-ventilatory response and subsequent negative effects on postural control. It is known that higher hyperventilation increases the frequency of respiratory movements which in turn increases postural sway and decreases balance performance (Hodges et al., 2002; Malakhov et al., 2014). Collectively, we can assume that elite players had advanced adaptation that enabled them to efficiently integrate inputs from the visual, vestibular, and somatosensory systems required for an adequate motor response to maintain balance and postural control (Wagner et al., 2011).

SLBT Validity, Reliability, and Usefulness

In this study, the SLBT showed good relative reliability both in normoxia and normobaric hypoxia. These results were comparable with the reliability in previous studies that have used an omni-axial balancing board and single-leg dynamic balance measurements in healthy and sporting populations (ICC range 0.76–0.97; Hildebrandt et al., 2015; Pojskic et al., 2020). To establish stable measurement conditions in hypoxia as a prerequisite for high measurement reliability, subjects performed the SLBT after 15' acclimation in the chamber. Moreover, good relative reliability could be also explained by high between-subjects variability in SLBT with those with higher playing level and longer playing experience having better balance than their sub-elite counterparts (**Table 2**; Pojskic et al., 2020). Furthermore, the absolute reliability (e.g., within-subjects variability) obtained in both HYP and NOR was in line with previous data for similar tests (CV% = 7–8%; Pojskic et al., 2020). The "acceptable" values could be attributed to players having the required test-dependent motor proficiency that is developed in agility-based team sports (Curtolo et al., 2017). In addition, it seemed that using the omni-axial balancing board as a simple testing tool for testing balance in an upright stance, which is a natural position for basketball players, could reduce potential covariates of performance (i.e., measurement error) and consequently enable better test–retest reliability and thus the potential to attribute results to the condition (NOR vs. HYP), instead to the testing equipment or protocol (Sekulic et al., 2017; Pojskic et al., 2020). Given that the SLBT did not show significant systematic change over the three trials, it means in practice that as long as the familiarization is conducted as described, two SLBT trials would be sufficient to obtain reliability data (Pojskic et al., 2020).

Moreover, the usefulness of the tests was estimated by a trial-to-trial change in balance performance and by comparing the TE and both the $SWC_{(0.2)}$ and $SWC_{(0.5)}$. For all tests, $SWC_{(0.2)}$

was shown to be "marginal" (i.e., TE > SWC) in both the NOR and HYP conditions. In contrast, in both conditions, $SWC_{(0.5)}$ exceeded TE, showing "good" usefulness. In summary, the SLBT can be utilized to detect moderate changes that exceed 0.5 × SD, in both the NOR and HYP conditions (Hopkins, 2004). Finally, SLBT performed only in HYP could be considered to have greater power to discriminate balance performance between playing levels. This is not totally unexpected, because the power of a test to discriminate playing levels is higher if the test protocol and conditions are sport specific and more demanding (Pojskic et al., 2020). The results are in agreement with Bruyneel et al. (2017) who recently reported differences in postural stability between two expertise groups of mountaineers at 3,200 m, but not at 1,500 m, tested immediately after exiting the cable car.

Limitations

This study has several limitations that must be acknowledged. First, the cross-sectional design limits the extent to which the group differences can be attributed only to the experimental conditions. Second, even though the subjects were blinded to the conditions being exposed to, using a double-blind design could make testing results less likely to be biased by the experimenter. Third, the technological design of the balance plate did not provide information on balance performance in specific mediolateral or anteroposterior directions, only a composite multidirectional balance index. Moreover, the balance plate used does not measure balance performance in basketball-specific movement patterns. Also, the study did not incorporate hypobaric hypoxia, which is most ecologically relevant for high altitude training and competition venues. This study did not evaluate physiological responses related to balance, such as the spinal reflex, brain activity, and muscles activity, which could help to identify mechanisms responsible for the observed differences between the conditions. In addition, even including highly trained players and calculating the minimum number of included subjects before, the sample size was low which limits the potential of the study to generalize the findings to the basketball population. Moreover, the age difference between groups was significant which might affect the results. Finally, the present results are specific to the basketball players chosen for the experiment, so caution should be exercised in generalizing the effects across other groups of athletes competing at high altitudes.

CONCLUSION

This study was the first to demonstrate that acute hypoxic exposure detrimentally affects balance performance in well-trained athletes active in team sports demanding high levels of dynamic balance and agility. Furthermore, acute effects of normobaric hypoxia on balance performance were modulated by the playing level with elite players having better resistance to its adverse effects than their sub-elite counterparts. The SLBT showed good reliability and usefulness in both normoxic and normobaric hypoxic conditions. However, the SLBT showed sufficient sensitivity to discriminate playing levels only in normobaric hypoxia equivalent to high altitudes (∼3,800 m).

Basketball coaches and practitioners should be aware that acute exposure to hypoxia equivalent to high altitudes (∼3,800 m) might negatively affect balance in sub-elite players and consequently execution of all activities that require it, such as acceleration and deceleration movements and sport-specific motor tasks (e.g., shooting and passing accuracy, and dribbling speed). Moreover, as acute normobaric hypoxia impairs balance performance, future studies can be designed to explore whether hypoxic balance training has the potential to facilitate faster and more advantageous training adaptation and provide a positive transfer to balance performance in hypobaric hypoxia (i.e., high altitude).

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the corresponding author, without undue reservation, to any qualified researcher.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the Regional Ethical Review Board in Umeå (2016/456-31). Written informed consent to participate in this study was provided by the participants. Legal guardian/next of kin provided informed consent for participants under 18 years of age.

AUTHOR CONTRIBUTIONS

HP contributed to the research concept and study design, literature review, data analysis and interpretation, statistical analyses, manuscript writing, and manuscript editing. HH contributed to the data analysis and interpretation, manuscript writing, and manuscript reviewing/editing. LR-Z contributed to the literature review, data collection, data analysis and interpretation, and manuscript writing. T-HT contributed to the data collection, manuscript writing, and manuscript reviewing/editing. All authors contributed to the article and approved the submitted version.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2021.748153/full#supplementary-material

Supplementary Figure 1 | Acute effects of normobaric hypoxia on oxygen saturation (SpO_2) and heart rate response (HR). T_{1-3}, trials 1–3 of the single-leg balance tests; NOR, normoxic condition; HYP, normobaric hypoxia condition. *Indicates significant difference between elite and sub-elite players, $P < 0.05$. †Indicates significant difference from values obtained at first measurement (after the initial 15 min of resting inside the chamber), $P < 0.05$. #Indicates a significant difference between HYP and NOR, $P < 0.05$. ###$P < 0.001$.

Supplementary Table 1 | Reliability parameters for the SLBT at DL and NDL in NOR and HYP.

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Inter-Individual Variability in Postural Control During External Center of Mass Stabilization

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Understanding the relation between the motion of the center of mass (COM) and the center of pressure (COP) is important to understand the underlying mechanisms of maintaining body equilibrium. One way to investigate this is to stabilize COM by fixing the joints of the human and looking at the corresponding COP reactions. However, this approach constrains the natural motion of the human. To avoid this shortcoming, we stabilized COM without constraining the joint movements by using an external stabilization method based on inverted cart-pendulum system. Interestingly, this method only stabilized COM of a subgroup of participants and had a destabilizing effect for others which implies significant variability in inter-individual postural control. The aim of this work was to investigate the underlying causes of inter-individual variability by studying the postural parameters of quiet standing before the external stabilization. Eighteen volunteers took part in the experiment where they were standing on an actuated cart for 335 s. In the middle of this period we stabilized their COM in anteroposterior direction for 105 s. To stabilize the COM, we controlled the position of the cart using a double proportional–integral–derivative controller. We recorded COM position throughout the experiment, calculated its velocity, amplitude, and frequency during the quiet standing before the stabilization, and used these parameters as features in hierarchical clustering method. Clustering solution revealed that postural parameters of quiet standing before the stabilization cannot explain the inter-individual variability of postural responses during the external COM stabilization. COM was successfully stabilized for a group of participants but had a destabilizing effect on the others, showing a variability in individual postural control which cannot be explained by postural parameters of quiet-stance.

Keywords: postural control, inverted pendulum, external stabilization, hierarchical clustering, postural variability

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INTRODUCTION

Maintaining postural equilibrium is fundamental for standing upright. This is achieved by coordinating motor commands and responses based on multiple sensory inputs and biomechanical constraints (Nashner, 1997). Measures of body sway as a movement of center of mass (COM) or center of pressure (COP) are commonly used to evaluate the performance of standing posture (Palmieri et al., 2002).

Investigating underlying mechanisms of postural control requires understanding of the relationship between COM and COP. Traditionally, COP variations are assumed to correct the unstable COM position back to the equilibrium (Johansson et al., 1988; Peterka, 2000). In contrast to traditional theories, other studies found that an additional purpose of the COP oscillations is to increase the sensory information flow from the environment (van Emmerik and van Wegen, 2002; Mochizuki et al., 2006; Stergiou et al., 2006; Carpenter et al., 2010).

One way to investigate the link between COM and COP is to stabilize the COM and observe parameters of COP oscillations, which was already done by Carpenter et al. (2010). However, their approach to COM stabilization was based on constraining the participant's joint movement by bracing them to a fixed board which did not allow for an ecological standing posture. Although this makes human body mechanically comparable to the inverted pendulum (Winter et al., 1998; Chagdes et al., 2013), it substantially affects postural control (Gage et al., 2004). Based on this, Gorjan et al. (2021) developed a method for stabilization of COM without mechanically constraining the natural movement of subjects. The method is based on a pulling system attached around the hips that stabilizes the motion of the COM by applying feedback forces. Even though this method does not constrain the joint motion of the subjects, it does have a mechanical effect by applying forces on the human body and this could have an effect on postural control.

To fully avoid constraining the body during the COM stabilization, we designed a novel methodological approach to stabilize the COM by moving the cart on which the participants can stand. This way, no mechanical forces are applied at the human body, except through the ground reaction forces. The cart stabilization method was designed based on the inverted pendulum model and the fact that the inverted pendulum can be stabilized by putting it on a moving cart (Lozano et al., 2000; Muskinja and Tovornik, 2006). Preliminary experiments using this stabilization method showed that COM of only a subgroup of participants was stabilized while it had a destabilizing effect for the others. Since the movement of the cart was controlled based on the movement of COM, the participant's reactions to the stabilization cannot be investigated by classical postural analysis based on COM motion. Nevertheless, according to Hsiao-Wecksler et al. (2003), it is possible to predict an individual's dynamic response to a mild perturbation only by analyzing quiet-stance data.

The aim of this study was to perform an experiment where a group of participants is subjected to external COM stabilization and to examine the possible relationship between the postural parameters of quiet standing and the susceptibility of an individual to external COM stabilization. We hypothesized that there should be a specific pattern of postural, kinematic and frequential parameters of COM during quiet standing associated with the individuals that are stabilized and those who are destabilized by the external COM stabilization.

MATERIALS AND METHODS

Participants

Eighteen healthy young adults (seven females) participated in the study. Their average age was 23.2 years ($SD = 2.1$ years), height 174.7 cm ($SD = 11.1$ cm), and body mass 70.2 kg ($SD = 12.8$ kg). The participants' individual characteristics are presented in **Table 1**. The age (18 to 30 years) was the inclusion criteria, and the exclusion criteria were history of neurological or musculoskeletal disorders or recent injury. The participants gave their written informed consent before participating in this study which was approved by the National Medical Ethics Committee of the Republic of Slovenia (No. 339/2017/7). Participants were divided into two groups based on the effect that stabilization method had on them. Seven participants were stabilized by the stabilization system and on the remaining 11 it had a destabilization effect in form of losing their balance (large oscillations or movement of the cart).

Study Design

The setup consisted of a cart with an integrated force plate (Kistler Instrumente AG, Winterthur, Switzerland) on which the participants stood. The cart was mounted on the rails that constrained the motion of the cart in anteroposterior direction. The cart was actuated by two motors located at both sides of the rail that pulled the cart with a steel wire. Location of the participant's COM was approximated by an infra-red marker placed under the L5 vertebra and recorded in real time by Optotrak motion capture system (3D Investigator, Northern Digital Inc., Waterloo, Canada) with 1,000 Hz sampling frequency. The motors were controlled by a double proportional–integral–derivative (PID) controller based on the location of the participant's COM. In effect, our setup allowed us to arbitrarily move the participant's COM in anteroposterior direction (**Figure 1**). We set the parameters of the PID controller by

TABLE 1 | Characteristics of participants with subject numbers (m = male, f = female).

Subject number	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18
Gender	f	m	f	m	m	m	f	m	m	f	f	m	m	f	m	m	m	f
Age	28	26	25	22	26	22	26	23	23	20	24	22	22	22	24	21	25	24
Height (cm)	165	184	155	184	172	186	167	180	182	158	165	191	177	160	175	183	190	172
Weight (kg)	65	76	51	90	77	69	58	85	74	54	59	94	67	58	60	82	82	63

first rough-tuning them to stabilize an aluminum inverted pendulum with a single rotational joint at the bottom and then fine-tuning them for the size and weight of the human body. The parameters were the same for all participants.

Participants were instructed to quietly stand on the cart with their feet hip-width apart. To exclude the effects of arm motion on quiet standing, we asked the participants to hold their arms crossed over the chest (Milosevic, et al., 2011). During the first 115 s, the cart was fixed and did not move, then it stabilized the participant's COM for 105 s and went back to the fixed mode for another 115 s. To avoid possible anticipatory behavior, we did not educate the participants about the time when the cart will perform the stabilization. In total, the participants were standing on the cart for 335 s.

Data Processing

To remove the noise, we first filtered the COM data obtained by the motion capture system using the second-order Butterworth low-pass filter with a 5 Hz cutoff frequency. We then calculated the RMS amplitude of COM and the RMS velocity of COM. Further, we calculated the power spectral density with fast Fourier transformation. Finally, we divided the spectrum to low frequency range (LF: 0.02–0.1 Hz), medium frequency range (MF: 0.1–1 Hz), and high frequency range (HF: 1–10 Hz) and calculated the area under the curve (AUC) for each range (Fujimoto et al. 2014).

Clustering and Statistical Analysis

There are several techniques available to investigate the correlations between different biomechanical parameters and to understand the movement variability of human subjects. Using supervised methods, correlations between the group of subjects can be performed using discriminant analysis paradigm (Tibarewala and Ganguli, 1982) where the groups of subjects need to be predefined. Moreover, several unsupervised methods were proven suitable to get insight into gait analysis (Holzreiter and Köhle, 1993; Begg et al., 2005; Horst et al., 2019). However, most of this research was predominantly oriented into classification of biomechanical parameters into predefined subject groups, ignoring the problem of unbiased discovering the underlying biomechanical parameters that lead to the changes of movement variability. Powerful methods to investigate inter-individual variability in human motion patterns without manual predefinition of groups are the clustering techniques. They were used to investigate variability in complex movements, such as walking and running (Mulroy et al., 2003; Bartlett et al., 2014). Moreover, hierarchical clustering algorithms have been used for mining gait patterns based on stride length and step frequency (Xu et al., 2006) and to investigate universal and individual characteristics of postural sway during quiet standing (Yamamoto et al., 2015). Multivariate clustering techniques were used for discovering human balance patterns and finding the association between COP parameters and different demographic and health characteristics of the participants (Malik and Lai, 2017).

Based on these previous methodological approaches, we used Ward hierarchical clustering technique which selects a pair of clusters to merge them at each step based on minimal error sum of squares (Mojena, 1977). The cluster solutions were generated using the anteroposterior COM displacement data from the 115 s period of quiet standing before the cart stabilized the participants. We selected RMS amplitude of COM, RMS velocity of COM, and AUC of HF COM motion as features for clustering, since postural, frequential, and kinematic parameters together thoroughly describe the movement of the human (Stins et al., 2011; Luca, 2016). We used *Z*-score standardization method to have equal influence of all included parameters (Mohamad and Usman, 2013).

To investigate possible effects of age, height, and weight of the participants on the stabilization and on the categorization results of the clustering, we compared the means of these

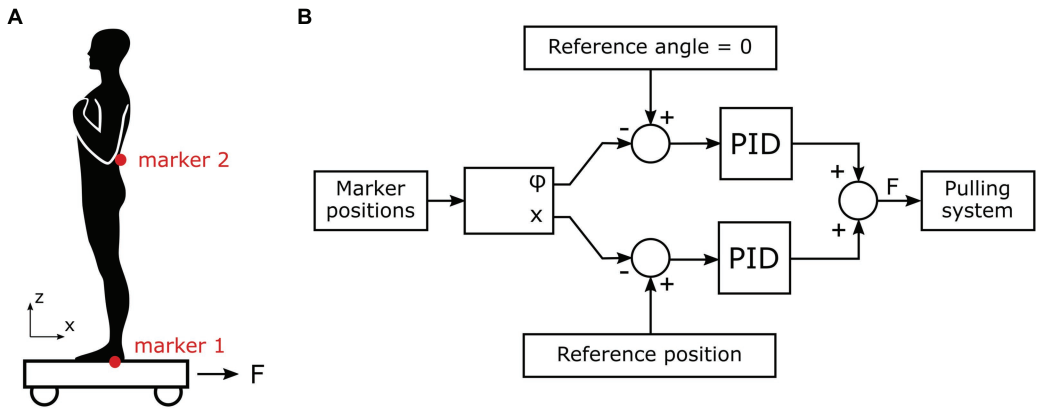

FIGURE 1 | (A) Human participant standing on the cart and the location of the motion capture markers. Marker 1 is placed on the cart below the ankle joint and marker 2 is placed on the L5 vertebra as an approximation of the center of mass (COM) position. **(B)** Scheme of cart controller to stabilize the participant's COM. φ is the angle between vertical line and the line that connects both markers, x is the horizontal position of marker 1, and F is the resulting force applied on the cart.

parameters using Welch t-test. All statistical analyses were performed using R, version 4.0.2.

RESULTS

Participants stabilized by our stabilization method were 1, 2, 9, 11, 12, 14, 16 (pink circles on **Figure 2A**) and the ones who were destabilized were 3, 4, 5, 6, 7, 8, 10, 15, 17, 18 (blue circles on **Figure 2A**). On the other hand, clustering based on the RMS amplitude of COM, RMS velocity of COM, and AUC of HF COM motion separated the participants into Group 1 with participants 2, 3, 4, 5, 6, 11, 14 and into Group 2 with participants 1, 7, 8, 9, 10, 12, 13, 15, 16, 17, 18. This indicates that the clustering analysis based on the selected parameters of quiet standing did not separate the participants into those that were stabilized by our method and those that were destabilized. Oppositely, the clustering analysis separated participants into two groups regardless of stabilization or destabilization effect of our method on them.

Moreover, RMS amplitude, RMS velocity, and AUC of HF COM motion of the participants grouped in Group 1 are higher compared to the participants grouped in Group 2 (**Figure 2B**). The two groups are different in terms of included parameters, but that difference does not explain the response of participants to the external stabilization.

Means and standard deviations of the clustering features for participants that were stabilized by our stabilization method and for those that were destabilized are presented on **Figure 3**.

Neither the participants' age, their height, nor their weight had an effect neither on stabilization nor on the categorization of participants using the clustering method. The statistical results of all relevant comparisons are presented in **Table 2**.

DISCUSSION

The cart stabilization method stabilized 7 out of 18 participants and had a destabilizing effect for the rest. Even though the method was the same for all participants, their response to the stabilization was different. Our results show that neither amplitude, velocity, nor the frequency parameters of COM during quiet standing cannot explain the inter-individual variability of postural responses during the external COM stabilization.

Clustering techniques were already used for analyzing postural data in similar studies (Xu et al., 2006; Malik and Lai, 2017). We selected COM frequency, amplitude, and velocity as features for the clustering algorithm, since these three parameters thoroughly describe the quiet standing movement. However, we also explored other subgroups of measured parameters (knee/hip/ankle joint angles, COM/COP velocity, amplitude, frequency), but no better clustering solutions were obtained in terms of equality of cluster sizes and according to the Ward criterion function (analyses not included in this paper). There is still room for further investigation of parameters that would better explain variability of postural sway. For example use of nonlinear parameters could improve the characterization of sway dynamics (Sabatini, 2000; Ghomashchi et al., 2011).

A viable possibility would be that the effect of stabilization could be explained by the differences of participants' age, height, or weight. The effects of anthropometric characteristics on standing balance were previously studied (Kejonen, et al., 2003;

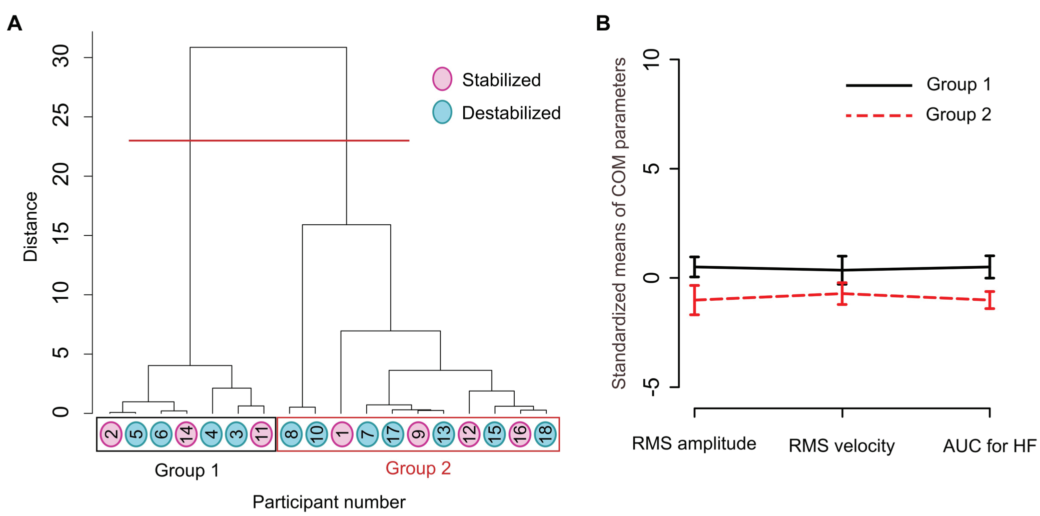

FIGURE 2 | (A) Dendrogram of the process of merging the units into groups with Ward's hierarchical clustering method based on the amplitude, velocity, and frequency parameter of COM as features. Our clustering solution separated participants into 2 groups. Participants stabilized by the cart stabilization method are marked with pink and the destabilized participants are marked in blue. **(B)** Means and standard deviations of standardized (subtracted mean and multiplied by 1,000) COM parameters for two groups grouped by Ward's hierarchical clustering method.

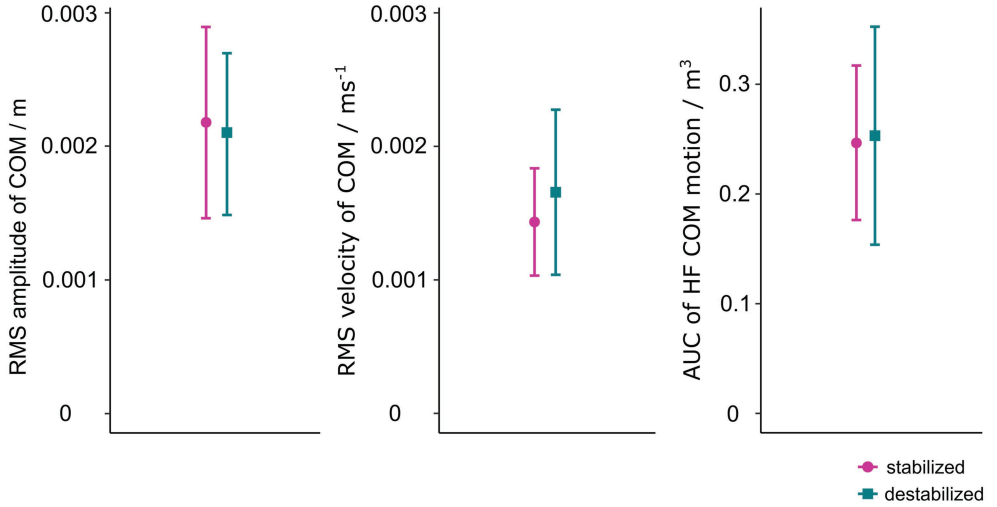

FIGURE 3 | Means and standard deviations of COM parameters of quiet standing for participants that were stabilized by cart stabilization method and for those that were not.

TABLE 2 | Statistical results of comparing participants' age, height, and weight based on stabilization and clustering categorization.

	Stabilised vs. Distabilised			Group 1 vs. Group 2		
	t	*df*	*p*	*t*	*df*	*p*
Age	−0.15	10.46	0.88	0.41	14.86	0.69
Height	−0.27	12.06	0.79	−0.71	10.95	0.49
Weight	−0.61	13.01	0.55	−0.42	12.47	0.68

Alonso, et al., 2015). They found a small effect of anthropometric characteristics on balance and that the height has the most influence on it. Even though the effects of anthropometry on balance are small, it could be enough to explain the differences of the COM stabilization used in our study. However, neither the age, height, nor the weight had an effect neither on stabilization nor on the categorization of the participants. We can therefore conclude that, in our case, these parameters were not determinant of the stabilization effects. We can also conclude that neither the age, height, nor the weight had an effect on the differences of COM amplitude, velocity, and frequency parameters found between Group 1 and Group 2. Nevertheless, there are several other biomechanical parameters (e.g., strength, physical ability levels, and ankle mobility) that affect postural control and whose possible effects on external COM stabilization should be explored in the future studies.

An important question is whether we can investigate responses to the cart stabilization method based on the quiet-stance data. Hsiao-Wecksler et al. (2003) found that it is possible to predict dynamic response of individuals to a mild perturbation only by analyzing quiet-stance data. Nevertheless, we could not predict the responses to stabilization based on the quiet-stance data, possibly because Hsiao-Wecksler et al. (2003) used a discrete impulse disturbance applied as a pull to the waist while we used a continuous, movement-dependent COM stabilization.

The analysis of the immediate postural reaction on the stabilization method has the potential to explain the inter-individual postural control variability (Moore et al., 1988). However, the nature of the cart stabilization method where COM was driving the cart movement, prevents us to separate the cause and the consequence of the COM movement. To analyze the immediate reaction to the stabilization, an additional experiment with the same participants investigating reaction to the discrete perturbation should be carried out. Moreover, the effect of the subjects knowing when the stabilization is initiated might have an important effect on their postural responses. This would allow us to elucidate if a possible difference in immediate reaction to the perturbation correlates with the response to the external stabilization.

Another possible cause for the different responses could be the differences in the sensitivity to the threat (Johnson et al., 2019). Since the participants were not informed about the stabilization, they could consider it as a postural threat. To investigate the difference in perception of postural threat, an additional experiment should be held where participants would be exposed to different levels of postural threat while we would measure the electro-dermal activity, which would allow us to compare the level of stress caused by the threat (Sibley et al., 2009, 2010).

There are many different algorithms for the stabilization of the inverted pendulum (Lozano et al., 2000; Muskinja and Tovornik, 2006) and we used one of the simplest version based on a pair of PID controllers with fixed parameters for all participants. It is important to note that the individual responses to stabilization did not correlate neither with the height nor with the weight of the participants. Furthermore, individualized tuning of the PID parameters would bias the experiment since the participants would have to be involved in the tuning procedure and would hence experience the external stabilization before the actual experiment.

Different responses to the external COM stabilization imply inter-individual variability in postural control. Even though general principles of postural control have been studied for decades, differences between individuals still cannot be fully explained (Foisy and Kapoula, 2016; Coste et al., 2021). For instance, it is widely accepted that the postural sway increases when eliminating the visual sensory information; however, there is a large group of people who sway less with their eyes closed (Lacour et al., 1997; Chiari et al., 2000). In conclusion, our study shows that the variability in individual postural control cannot be explained by postural parameters of quiet-stance. Further experiments are needed to understand the roots of postural variability and, among others, suggest how to improve the stabilization method to be applicable for a larger population.

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 National Medical Ethics Committee of the Republic of Slovenia (No. 339/2017/7). The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

All authors have made substantial contributions to the conception and design of the study, acquisition of data, analysis, interpretation of data, drafting the manuscript and revising it critically for important intellectual content.

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frontiers in Physiology

ORIGINAL RESEARCH
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Neuromuscular Fatigue Affects Calf Muscle Activation Strategies, but Not Dynamic Postural Balance Control in Healthy Young Adults

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Neuromuscular fatigue could negatively affect postural balance, but its effects on dynamic postural regulation are still debated. This study aimed to investigate whether a fatigue protocol on calf muscle could affect muscle activation strategies and dynamic balance performance. Seventeen male adults (age 24.1 ± 4.6 years; height 183.9 ± 7.2 cm; weight 80.2 ± 7.2 kg) volunteered in the study. They performed a dynamic test on an instrumented platform, which provided anterior-posterior oscillations on the sagittal plane, before and after a localized fatigue protocol. High-density surface electromyographical (EMG) signals were recorded bilaterally from the soleus and the medial gastrocnemius muscles. The fatigue protocol, consisting of two quasi-isometric tiptoe standing exercise to failure with a fixed load, did not affect the global dynamic balance performance. Conversely, the frequency value corresponding to 95% of the total power spectrum density of the angular displacement signal increased after fatigue (from 1.03 ± 0.42 to 1.31 ± 0.42 Hz; $p < 0.05$). The EMG analysis showed a significant difference in the PRE/POST fatigue ratio of the root-mean-square (RMS) between the soleus and the gastrocnemius medialis muscles. No differences were detected for the coefficient of variation and the barycenter coordinates of the RMS EMG values between muscles and sides. The variations in the frequency content of the angular displacement and EMG activity across muscles may be related to an increase in the calf muscles stiffness after fatigue. The role of neuromechanical calf muscle properties seems to be relevant in maintaining the dynamic postural performance after a quasi-isometric fatigue protocol until failure.

Keywords: muscle fatigue, high density EMG, dynamic balance, balance control, exercise

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INTRODUCTION

Postural balance control is fundamental both in everyday life and sports activities. If posture represents the position of the different body segments (Paillard, 2017), balance is defined as the capacity to keep the center of pressure (COP) within the base of support to avoid falling (Winter et al., 1996). The scientific literature differentiates static from dynamic postural balance control: the first is described as the ability of a person to maintain the balance in an unperturbed environment (e.g., quiet standing on a firm surface) (Macpherson and Horak, 2013); the second represents the ability of a person to cope with sudden changes of postural conditions (e.g., displacement of the base of support) or external mechanical perturbations (e.g., forces applied to large body segments) (Paillard, 2019). An efficient postural balance control can reduce the fall risk and contribute to the efficiency of motor/sports performance (Paillard, 2017). On this topic, muscle fatigue has been demonstrated to affect neuromuscular control negatively, and thus postural balance, since fatigue can alter the somatosensory inputs (e.g., the threshold of muscle spindle discharge) (Gribble and Hertel, 2004), motor neuron discharge statistics (Enoka, 2012) and muscle stiffness (Granata et al., 2004). The effect of fatigue on postural balance has been investigated both after a total body exercise as uphill walking, cycling, or running (Nardone et al., 1997, 1998) and following exercises involving single joints (i.e., ankle, knee, or hip joints) (Gribble and Hertel, 2004; Salavati et al., 2007; Gimmon et al., 2011). Indeed, it has been shown that strenuous treadmill exercise negatively affects the COP sway measured on a force platform during quiet standing and that the effect vanishes within a few minutes (Nardone et al., 1998). Moreover, COP sway is little influenced when exercises are executed below the anaerobic threshold (Nardone et al., 1997). Conversely, COP is primarily affected when exercises envisage strong intensity muscle contractions providing a significant amount of proprioceptive stimulation (e.g., running vs. cycling) (Nardone et al., 1997; Zemková, 2014). In addition to this, localized muscle fatigue seems to significantly influence postural balance control when proximal lower limb muscle groups are involved (Gribble and Hertel, 2004; Salavati et al., 2007). Furthermore, plantar flexor muscles are crucial for maintaining the upright posture (Loram et al., 2005) due to the anterior position of the COP concerning the ankle joints (Winter, 1995). Indeed, when ankle plantar flexors are fatigued, the postural balance impairment during static posturography mainly occurs in the sagittal plane (Gimmon et al., 2011). To date, only a few studies reported at the same time the effect of fatigue both on postural balance performance and plantar flexor muscle activity (Ritzmann et al., 2016; Watanabe et al., 2018). In particular, the decrement of static balance performance after fatigue has been explained by a change of the neuromuscular control that caused an increase of muscle co-contractions and, as a consequence, of joint stiffness (Ritzmann et al., 2016). Watanabe et al. (2018) demonstrated with bipolar surface electromyography that the worsening of balance performance was accompanied by a reduction of the low-frequency common input to the plantar flexor muscles.

Although it seems conceivable to perform dynamic besides static tests because they provide more discriminant information on postural balance performance (Petró et al., 2017), the literature on the effect of fatigue on dynamic postural control is still scarce. Miller and Bird (1976) have shown that localized fatigue only on some specific muscle groups can affect the dynamic postural control performance. More recent studies highlighted contradicting results on the effect fatigue has on dynamic balance control (Salavati et al., 2007; Marcolin et al., 2016). Regardless of the results, these studies demonstrate that compensatory mechanisms may occur at the neural or peripheral levels of specific muscle groups to overcome the adverse effects of muscle fatigue on dynamic postural control.

Nevertheless, none of the previous studies investigated the electromyographical (EMG) activity of the muscles fatigued while executing the dynamic balance tests.

Therefore, the present study aimed to increase the body of knowledge on the interplay between neural and mechanical components of the dynamic postural balance control in healthy subjects following a standardized fatigue protocol designed to involve the ankle dorsiflexor muscles. Specifically, we aimed to investigate if the fatigue protocol could influence the calf muscle activation strategies during the performance of a dynamic postural balance test. We hypothesized that compensation mechanisms in the calf muscle activation strategies would be engaged to minimize the effects of the fatigue protocol on the dynamic postural balance performance.

MATERIALS AND METHODS

Participants

A total of 17 young male adults (age 24.1 ± 4.6 years; height 183.9 ± 7.2 cm; weight 80.2 ± 7.2 kg) participated in the study. Exclusion criteria were the presence of pathologies to the vestibular system, the presence of non-corrected visual refractive errors, and the presence of acute or chronic pathologies affecting muscles, tendons, and joints of the lower limbs. Each participant was instructed on the experimental protocol before giving his written informed consent to the participation. The study was approved by the ethical committee of the Department of Biomedical Sciences, University of Padova.

Experimental Procedure

Dynamic Postural Balance

The dynamic postural balance of the participants was assessed barefoot on a custom-made marine plywood platform (length: 50 cm; width: 50 cm; height: 8.5 cm) which rotated along a single axis providing anterior-posterior perturbations on the sagittal plane. A marine plywood semicylinder allows the platform to rotate 16 deg, both anteriorly and posteriorly. The surface of the platform was covered with an anti-slip film. Measurements were taken before and after a fatigue protocol. More in detail, each participant had to maintain an upright posture with arms along their sides. The midpoint of each foot, identified as half the distance between the medial malleolus and the head of the first metatarsal head, was marked and positioned over the line which divided into two equal parts the surface of the platform. Feet had to be kept parallel with the step width equal to the pelvis width. The protocol consisted of two trials of 40 s with 3-min rest in between. During the trials, participants were asked to maintain the platform parallel to the ground while straight gazing at a target positioned in front of them at 2.5 meters. After that, the fatigue protocol occurred. It consisted of two quasi-isometric tiptoe standing calf exercises to failure gripping a 10-kg disk on each hand. The participants had to stand on their tiptoes until failure: the first standing calf ended when the heels touched the platform, which had been preventively locked, avoiding oscillations. After recovery of 1 min, participants performed the second quasi-isometric standing calf. The time duration of each quasi-isometric standing calf to failure was recorded. Indeed, inducing an incapacity to continue a particular effort has been described as one of the three techniques generating fatigue (Paillard, 2012). During the fatigue protocol, the operators verbally encouraged participants to do their best. Immediately after the participants' heels touched the platform in the second standing calf to failure, the platform was unlocked, and participants performed again a 40-s trial with the same indications followed in the two previous trials. The representation of the experimental setup is shown in **Figure 1**.

The anterior and posterior oscillations of the platform were recorded by an optoelectronic system (V120 Trio, OptiTrack, United States) that collected at a sampling frequency of 120 Hz the three-dimensional position of two reflective markers applied on the edge of the platform.

High-Density Surface Electromyography

High-density surface EMG signals were recorded from the soleus and the medial gastrocnemius muscles of each leg. The recordings were made in monopolar derivation with two-dimensional adhesive grids (Spes Medica, Salerno, Italy) of 13 (rows) × 5 [columns electrodes 8/8/spaced by 8 mm (1 mm diameter)] on each muscle. The high-density surface EMG signals were acquired using an EMG-USB2 + amplifier (256 channels plus 16 auxiliary channels; OT Bioelettronica, Turin, Italy). After the skin was shaved and cleaned with abrasive paste and water, the electrode grids were covered with a double-sided foam and the electrode cavities were filled with conductive paste (Spes Medica, Salerno, Italy). The matrices were positioned following the guidelines according to Barbero et al. (2012). The surface EMG signals were amplified (EMG-USB2+, OT Bioelectronics, Italy), band-pass filtered (−3 dB, from 20 to 500 Hz), and sampled at 2,048 Hz with the OT Biolab software (OT Bioelettronica, Turin, Italy). Surface EMG signals were amplified with variable gain (500–2,000 V/V). Although the tibialis anterior was not involved during the fatigue protocol, bipolar surface EMG signal was recorded by two self-adhesive pre-gelled electrodes (diameter 1 cm), placed at an interelectrode distance of 20 mm as control, allowing to exclude the instauration of compensatory activation patterns. The bipolar EMG signals were filtered with a 512 Hz low pass filter before being recorded. The bipolar surface EMG signals were collected by a16 channel surface electromyographic signal amplifier (OT Bioelettronica, Turin, Italy).

Data were analyzed offline using MATLAB 2019b (The MathWorks, Inc., Natick, MA, United States). The kinematic and electromyographic data were collected synchronously through a manual trigger.

Data Analysis

Kinematic Data Analysis

The platform's oscillation was calculated with a custom SmartAnalyzer program (BTS Bioengineering, Milan, Italy) starting from the three-dimensional coordinates of the two reflective markers. Thus, 0°corresponded to the platform parallel to the ground, positive angular values to posterior oscillation (i.e., ankle dorsiflexion), and negative angular values to anterior oscillation (i.e., ankle plantarflexion). The platform's whole range of motion was 32° (16° anteriorly and 16° posteriorly, respectively). Three parameters were considered for the assessment of the dynamic postural balance similar to a previous work (Sarto et al., 2020): the overall integral of the time-angle curve (Full Balance, FB); the time participants kept the platform between 4° of anterior and 4° of posterior oscillation (Fine balance, FiB); the time participants kept the platform between 8° of anterior and 8° of posterior oscillation (Gross balance, GB). Moreover, we calculated the power spectral density (PSD) as the square of the amplitude spectrum estimated by fast Fourier transformation of the time-angle input signal. PSD allowed highlighting which oscillation frequencies were predominant in the signal input, providing information not clearly visible in the time domain. Specifically, we adopted the in-built MATLAB periodogram function with a rectangular window of duration equal to the signal length (The MathWorks, Inc., MA, United States). Then, the integral of the PSD curve was calculated to extrapolate for each trial the frequency values corresponding to the 50% (PSD_50%) and 95% (PSD_95%) of the total PSD.

Electromyographical Data Analysis

The root mean square map amplitude (RMS_MAP) of the EMG signals was calculated during the dynamic tasks performed on the platform before and after the fatigue protocol. RMS_MAP was estimated from the average of all differential channels of the electrode grid (Martinez-Valdes et al., 2020), as this provides more reliable estimates of muscle activity (Vieira and Botter, 2021). The RMS_MAP values calculated on the entire high-density surface grid for each calf muscle were averaged over the 40-s signal duration. Additionally, the root mean square amplitude (RMS) derived from a bipolar derivation with larger electrodes was calculated. Specifically, the monopolar EMG signals from two sets of five electrodes in close proximity were averaged to derive the EMG signals detected from two large electrodes. The derived EMG signals were subtracted to estimate a bipolar EMG derivation with an interelectrode distance of 1.6 cm, similar to a previous study (Del Vecchio et al., 2017). The RMS values for each large bipolar configuration were extracted for each calf muscle and averaged over the 40-s signal duration. The RMS values were also calculated for the tibialis anterior bipolar EMG signals (RMS_TA) in the same segments. In our experimental protocol, we could not measure the maximal voluntary activation of the investigated muscles. Therefore, the variation of the RMS_MAP, RMS and RMS_TA values before and after the fatigue protocol (PRE/POST*100) was compared across muscles and sides. Additionally, the spatial distribution of the muscle activity was estimated by extracting the barycenter coordinates (BAR_X and BAR_Y, respectively) and the coefficient of variation (CoV) of the RMS maps for the soleus and the medial gastrocnemius muscles on both monopolar and differential signals. In order to match with the RMS analysis, both barycenter coordinates and CoV variables were normalized to the PRE condition.

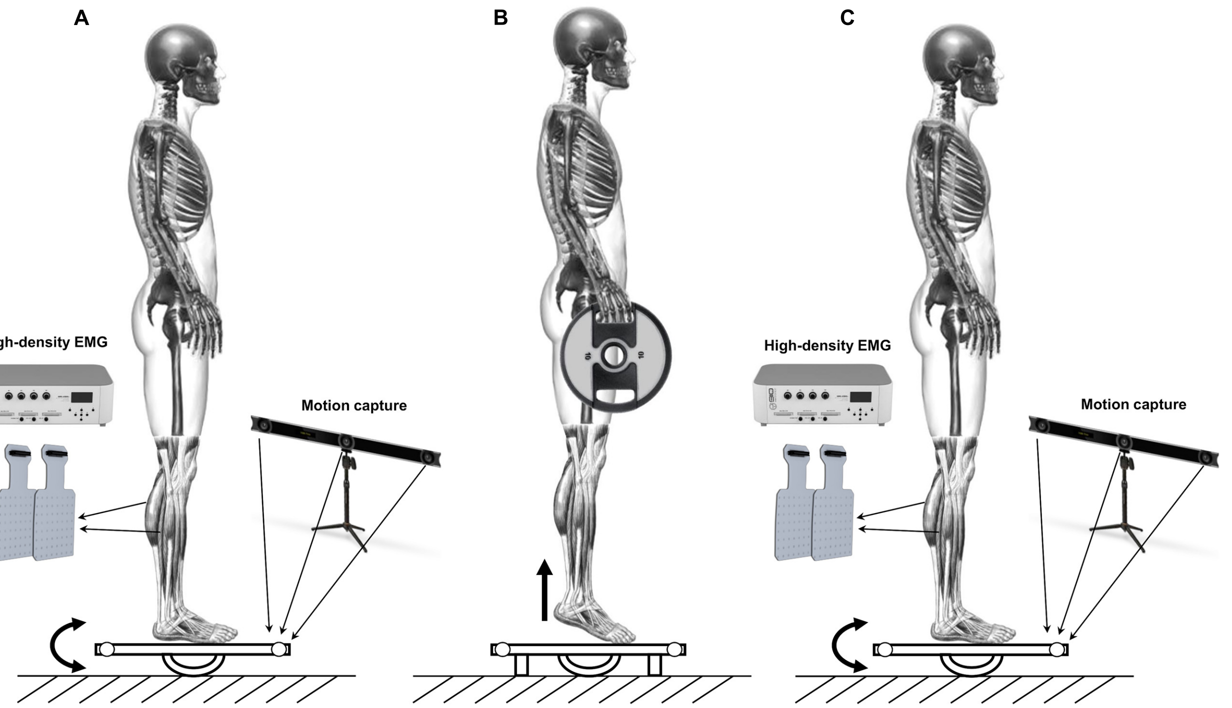

FIGURE 1 | Schematic representation of the experimental protocol. **(A)** 40 s trial on the oscillating platform. **(B)** Standing calf to failure gripping a 10 kg disk on each hand. **(C)** 40 s trial on the oscillating platform after the fatigue protocol.

TABLE 1 | Kinematics and power spectral density results before and after the fatigue protocol.

	PRE	POST	*p*-value	Cohen's *d*
Full balance (FB)	133.5 ± 44.16	134.3 ± 24.58	0.94	0.02
Gross balance (GB) [s]	36.78 ± 3.06	38.19 ± 1.69	0.08	0.53
Fine balance (FiB) [s]	27.92 ± 6.49	25.80 ± 4.40	0.22	0.37
PSD_50% [Hz]	0.15 ± 0.17	0.13 ± 0.10	0.66	0.16
PSD_95% [Hz]	1.03 ± 0.42	1.31 ± 0.42	0.02	0.67

Data are expressed as means and standard deviations.

Statistical Analysis

A two-tailed paired t-test was carried out to compare the kinematic parameters before and after the fatigue protocol and compare the duration of the two fatigue trials. This analysis was performed with GraphPad Prism version 4.00 (GraphPad Software, San Diego, CA, United States). Cohen's d effect size (d) was calculated with G*Power 3.1 (Faul et al., 2007) and was interpreted as follow: 0.00–0.19: trivial; 0.20–0.59: small; 0.60–1.19: moderate; 1.20–1.99: large and >2.00: very large (Hopkins et al., 2009). A two-way ANOVA for repeated measures was employed to compare the ratios of the EMG variables (RMS, RMS_MAP, COV, BAR_X, and BAR_Y) with factors of side (right and left) and muscle (soleus and gastrocnemius muscle). Additionally, a paired t-test was performed for both muscles to compare the EMG activity before and after the fatigue protocol. To compare the RMS_TA percentage values, a One-Sample t-test was performed for left and right legs independently. The significant level was set at $p < 0.05$ for all the statistical tests performed. All EMG statistical analyses were performed with IBM SPSS Statistics 26 (IBM, Armonk, NY, United States).

RESULTS

The time duration of the second fatigue trial (196.7 ± 80.10 s) was shorter than the first one (279.8 ± 97.49 s) and the difference was statistically significant ($p < 0.01$; $d = 0.92$), highlighting the effectiveness of the fatigue protocol adopted. Nonetheless, the fatigue protocol did not affect the dynamic balance performance on the oscillating platform. In fact, for both FB, FiB, and GB, no statistically significant worsening or improvement was detected. Conversely, the frequency value corresponding to 95% of the total PSD (PSD_95%) showed a statistically significant increment ($p < 0.05$; $d = 0.67$), underlying a shift toward higher frequencies of oscillation in the dynamic postural balance test after the fatigue protocol. An example of the angular displacement signal before and after the fatigue protocol is reported in **Figure 2A** together with the integral of the PSD time-angle curve (**Figure 2C**). Numerical data of both kinematics and power spectral density results are presented in **Table 1**.

Figure 2B shows the unnormalized RMS activity maps of each muscle and side before and after the fatigue protocol for a representative subject. Previous studies have well described the representation through color maps that characterize the electromyographic activity (Orizio et al., 2017;

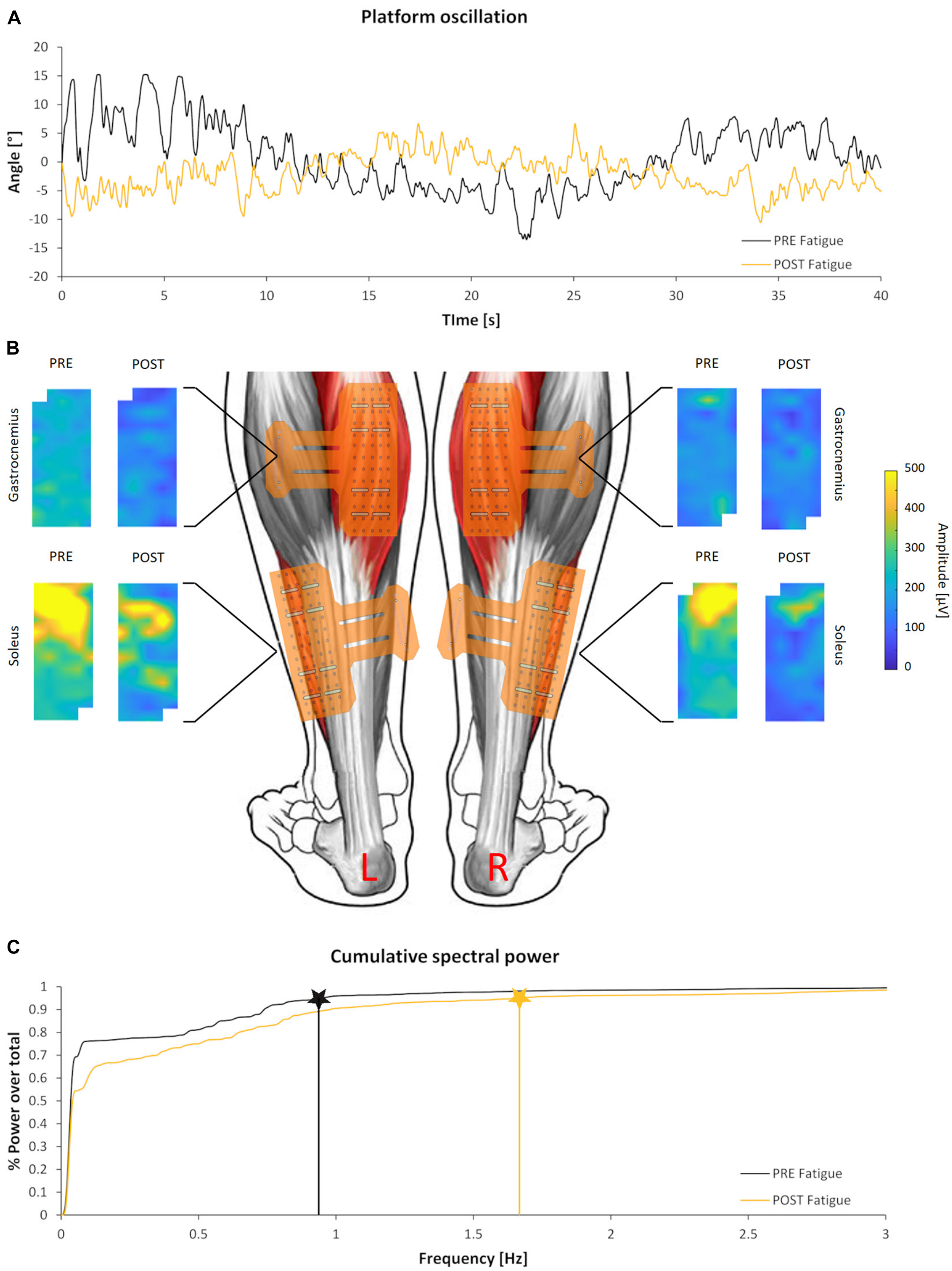

FIGURE 2 | Kinematics and electromyographical (EMG) outcomes of a representative subject. **(A)** Angular displacement signal of the oscillating platform before (black) and after (yellow) the fatigue protocol. **(B)** Colored maps representing the spatial distribution of EMG activity before and after the fatigue protocol of the calf muscles analyzed. Maps are scaled to the same amplitude to highlight the differences between and within muscles. **(C)** Integral of the PSD time-angle curve and correspondent PSD_95% values before (black) and after (yellow) the fatigue protocol.

Falla and Gallina, 2020; Pradhan et al., 2020). The parameters extracted from the EMG analysis showed significant differences after the fatigue protocol. In particular, the normalized RMS_MAP values averaged over the full matrices for all subjects demonstrated an effect of muscle ($F = 7.4$, $p = 0.015$, $\eta p2 = 0.31$) but not of side ($F = 3.4$, $p = 0.083$, $\eta p2 = 0.176$, **Figure 3**). Similarly, results of the RMS values calculated from the large bipolar derived from the surface EMG matrices showed effect of muscle ($F = 8.4$, $p = 0.013$, $\eta p2 = 0.345$) but no effect of side ($F = 2.7$, $p = 0.116$, $\eta p2 = 0.148$), demonstrating a larger variation of normalized RMS values in the soleus muscle during the dynamic activity after the fatigue protocol. Notably, EMG

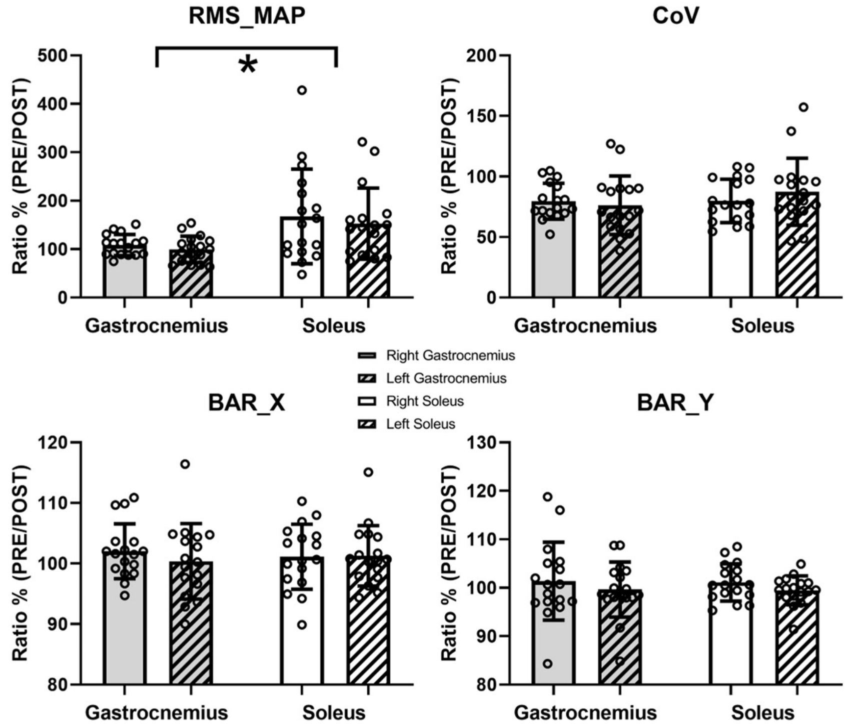

FIGURE 3 | Results of the EMG analysis. Data are presented as means and standard deviations. RMS_MAP, CoV, and barycenter coordinate (BAR_X and BAR_Y) are reported. * represents statistically significant differences between muscles ($p < 0.05$).

activity showed a statistically significant reduction for the soleus (left side: $p = 0.028$, $d = 0.52$; right side: $p = 0.022$, $d = 0.68$) and no variation for the gastrocnemius medialis (left side, $p = 0.496$; right side, $p = 0.440$) after the fatigue protocol.

Conversely, the spatial distribution of the RMS maps over the muscles during the dynamic activity did not change after the fatigue protocol. Specifically, the PRE/POST variations in the spatial variability (CoV for RMS values) did not show statistical difference, with no muscle ($F = 1.163$, $p = 0.297$, $\eta p2 = 0.068$) or side ($F = 0.306$, $p = 0.588$, $\eta p2 = 0.019$) effects (**Figure 3**). Neither the ratios of the barycentres of the RMS activity showed statistically significant effects considering BAR_X (muscle, $F = 0$, $p = 0.994$, $\eta p2 = 0$; side, $F = 0.414$, $p = 0.529$, $\eta p2 = 0.025$) and BAR_Y (muscle, $F = 0.06$, $p = 0.81$, $\eta p2 = 0.04$; side, $F = 1.031$, $p = 0.325$, $\eta p2 = 0.061$). No changes were found between the right ($p = 0.115$) and left ($p = 0.239$) tibialis anterior for the normalized RMS_TA values after the fatigue protocol.

DISCUSSION

The present study aimed to investigate the effect of a neuromuscular fatigue protocol on dynamic postural performance and calf muscle activation strategies. The main finding of the current study was that the reduction of the EMG activity of the soleus during the dynamic balance task after fatigue did not affect the global dynamic postural balance performance. We can speculate this behavior is part of calf muscle compensatory mechanisms acting to minimize the effect of fatigue on the dynamic balance performance.

Previous studies showed that changes in the dynamic postural balance performance depend on the characteristics of the fatigue protocol. For example, localized muscle fatigue has been shown to affect dynamic balance only when knee and hip flex-extensors, but not plantar flexors, were fatigued (Miller and Bird, 1976). Similarly, it has been demonstrated that exercise inducing fatigue in plantar flexors had no effects on postural control (Adlerton and Moritz, 1996; Caron, 2004; Rojhani-Shirazi et al., 2019). In particular, Alderton and colleagues suggested that compensatory mechanisms are put in place to counteract plantar flexors fatigue (i.e., increased reflex activity in muscle spindles or increased muscle stiffness due to fatigue) and maintain an unaltered postural control. Nonetheless, we are still far away from a clear consensus since in other similar studies, localized muscle fatigue of the plantar flexors has been demonstrated to decrease both static (Yaggie and McGregor, 2002; Vuillerme et al., 2006; Gimmon et al., 2011) and dynamic (Gribble and Hertel, 2004; Salavati et al., 2007) postural control. To this extent, our findings support the body of literature that found no differences in the dynamic balance performance after a localized plantar flexors fatigue.

Interestingly, when generalized muscle fatigue was induced, the dynamic postural behavior was similar to the one reported in the current study. In line with this, ultramarathon runners fatigued by more than 15-h race (Marcolin et al., 2016) or recreational runners after 25-min treadmill running

(Marcolin et al., 2019) showed a similar dynamic balance performance before and after fatigue. Therefore, although it appears to exist causation between muscle fatigue and postural control, this relation seems highly dependent on the type of the fatigue protocol (i.e., localized on specific muscles or generalized) and postural task (i.e., static or dynamic). The reasons for such discrepancies may be found in the influence of fatigue on the interaction between neural and mechanical components of the human postural control. Alternatively, we can speculate that an increase in the vigilance level after the fatiguing could occur (Paillard, 2012), improving the effectiveness of the descending drive for the postural muscle motor neurons activation and the integration of afferent information (Nardone et al., 1998).

Nonetheless, the first hypothesis seems the most robust since muscle fatigue is well recognized reducing the rate of force development (Allen et al., 2008), altering proprioceptive information from muscle afferents (Vuillerme and Boisgontier, 2008), inducing metabolic inhibition of the contractile process and excitation-contraction coupling failure, and increasing joint stiffness (Gajdosik, 2001; Zhang and Zev Rymer, 2001; Hoang et al., 2007).

Our findings observed an increase in the high-frequency content (i.e., an increment of PSD_95%) of the angular displacement signal (**Figure 2C** and **Table 1**) after the fatigue protocol. The frequency content change could have been indirectly related to a higher rigidity of the triceps surae muscle group that emphasized this aspect of the motor control mechanical outcome. Indeed, Baratta and Solomonow (1992) reported that the more rigid the system, the less the tension dynamics filter by its viscoelastic elements. Stiffness could also be reflected in a more effective force transmission (Bojsen-Møller et al., 2005; Cogliati et al., 2020). Moreover, previous studies suggested that increased joint stiffness (Winter et al., 1998; Caron, 2003; Corbeil et al., 2003) plays a determinant role in postural control management. Although recent researches underlined limitations in this interpretation for static postural control (Morasso and Schieppati, 1999; Loram et al., 2005, 2009), it is reasonable to assume a key role of muscle stiffness during dynamic postural tasks (Adlerton and Moritz, 1996), like the one performed in the current study. Future studies are warranted to investigate the possible relationship between the frequency components of the kinematic signal detected in our experiment and the motor unit neural control features retrievable by high-density EMG processing (Negro et al., 2009). According to the theory that increased stiffness of plantar flexors could counterbalance the effect of fatigue letting unaltered the dynamic balance performance on the platform, it was not surprising to observe a difference in the variation of the surface EMG amplitude between soleus and gastrocnemius muscles after the fatigue protocol (**Figure 3**). The unchanged EMG activity of gastrocnemius medialis may provide evidence of higher soleus muscle engagement after the fatigue protocol performing a dynamic balance test with a relatively small range of motion. Interestingly, we did not observe changes in the coefficient of variation and the barycenter coordinates of the RMS values across the EMG matrices, highlighting no modifications in the spatial activation strategies of the calf muscles performing the dynamic balance task after fatigue. In line with our work, previous studies have found minimal variations of the RMS of the barycenter during isometric or dynamic fatigue tasks in healthy individuals (Falla and Farina, 2007; Arvanitidis et al., 2021). This finding certainly deserves further investigation.

Last, it is relevant to mention a limitation of the present work. Although participants were instructed to maintain an upright posture with arms along their sides, we observed slight movements of the trunk relative to the pelvis that we did not control to ensure the participants performing the task at their best. Indeed, those tiny motions could have somehow affected the participants' center of mass location and consequently could have marginally influenced the calf muscle activation.

CONCLUSION

The present study hypothesized compensatory mechanisms in the triceps surae muscles before and after the fatigue protocol that allowed maintaining similar performance levels in the dynamic postural task on the oscillating platform. Specifically, we highlighted two compensation strategies consequent the fatigue protocol: (i) a difference between soleus and gastrocnemius activity PRE/POST fatigue; (ii) an expansion of the frequency content of the platform oscillations, likely due to an overall increase of the calf muscles stiffness. Although we hypothesized causation between muscle fatigue and neuromechanical compensation strategies, further studies need to deepen these compensatory mechanisms investigating the key role of muscle stiffness, the neuromuscular complexity of the physiological parameters through fluctuation and variability analysis (e.g., approximate entropy and detrended fluctuation analysis), and the neural control features of motor units retrievable by high-density EMG processing.

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 Ethical Committee of the Department of Biomedical Sciences, University of Padua. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

GM, MC, CO, and AP conceived and designed the experiments. GM, MC, AC, FN, and RT performed the experiments. GM, MC, AC, FN, CO, and RT analyzed the data. AP and CO contributed to the materials. GM, MC, AP, and FN wrote the manuscript. All authors approved the final version of the manuscript.

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Effects of Fatigue on Postural Sway and Electromyography Modulation in Young Expert Acrobatic Gymnasts and Healthy Non-trained Controls During Unipedal Stance

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This study investigated whether expert acrobatic gymnasts respond differentially than their non-trained counterparts during a single-legged stance task performed before and after a protocol designed to induce fatigue in the ankle plantarflexor muscles in terms of (a) postural steadiness and (b) electromyography (EMG) activation. We hypothesized that neuromuscular adaptation due to training would lead to different behavior of center of pressure (COP) and EMG quantifiers after fatigue. Twenty eight female volunteers (aged 11 to 24 years) formed two groups: expert acrobatic gymnastics athletes (GYN, $n = 14$) and age-matched non-gymnasts [control (CTRL), $n = 14$]. Fatigue of the ankle plantarflexors (dominant leg) was induced by a sustained posture (standing on the toes) until exhaustion. Traditional COP parameters (area, RMS, mean velocity, and power spectrum at low and high frequency ranges) were obtained with a force plate, and time and frequency-domain EMG parameters were obtained by surface electrodes positioned on the *tibialis anterior*, *soleus*, *lateral gastrocnemius*, *medial gastrocnemius*, *vastus lateralis*, *biceps femoris*, *spinal erector* and *rectus abdominis* muscles. The main results showed that fatigue induced a significant increase in postural oscillations in the ML axis (including RMS, velocity and frequency components of the power spectrum), with no significant effects in the AP axis. In terms of postural sway parameters (i.e., COP quantifiers), no superior balance stability was found for the GYN group as compared to CTRL, irrespective of the fatigue condition. On the other hand, the modulation of EMG parameters (in both time and frequency domains) indicated that expert acrobatic gymnastics athletes (as compared to healthy untrained matched controls) used different neuromuscular control strategies to keep their postures on single-legged quiet standing after the fatiguing protocol. The present results improve our knowledge of the mechanisms behind the interplay between fatigue and postural performance associated with the neuromuscular adaptations induced by sport practice.

The design of gymnastics training might consider strategies aimed at improving the performance of specific muscles (i.e., *tibialis anterior*, *soleus*, *biceps femoris*, *spinal erector*) for which particular activation patterns were used by the acrobatic gymnastics to control single-legged quiet standing.

Keywords: postural control, balance, training, steadiness, sport

INTRODUCTION

The control of body balance (i.e., quiet stance) is regulated by the postural control system, which needs to integrate information from visual, vestibular and somatosensory sources to generate adequate motor responses that keep the body still (Winter, 1995; Peterka, 2002). Assessments of balance ability are routinely performed with the use of a force plate that tracks the location of the center of pressure (COP), especially when postural deviations are small so that the body might be simplified as an inverted pendulum model, thereby reflecting the so-called ankle strategy (Peterka, 2003).

Some studies have pointed out the practice of sports such as gymnastics promotes efficient stimuli in generating neuromuscular adaptations, which seems to be associated with intense and regular practice of complex motor skills, including high demand for production and torque control around the ankle joint. Specifically, ankle plantarflexors are traditionally considered the most important muscle group engaged in the maintenance and control of postural balance (Rothwell, 1994; Winter, 1995), and hence it is plausible that increased performance of these muscles (e.g., due to long-term practicing) is accompanied by improvements in postural steadiness. In this direction, some studies have shown that trained gymnasts exhibit superior balance ability as compared to untrained participants or athletes involved in sports activities with different characteristics (Debu and Woollacott, 1988; Carrick et al., 2007; Andreeva et al., 2021). However, this issue remains controversial, as other studies showed no significant differences in balance ability between gymnasts and controls under certain conditions. Indeed, the literature indicates that such a "superior balance ability" is influenced by a variety of factors such as task specificity (Vuillerme et al., 2001a; Asseman et al., 2004, 2008; Isableu et al., 2017), attentional demand (Vuillerme and Nougier, 2004), level of expertise (Opala-Berdzik et al., 2021), sensory condition (Vuillerme et al., 2001a; Asseman et al., 2005, 2008; Garcia et al., 2011), anthropometric characteristics (Opala-Berdzik et al., 2018, 2021), age (Garcia et al., 2011), sex (Milosis and Siatras, 2012) and the parameter of postural sway being analyzed (Isableu et al., 2017; Opala-Berdzik et al., 2018). To the best of our knowledge, the majority of these studies involved artistic gymnasts, although rhythmic and acrobatic gymnastics have also been explored, with somewhat similar results (Calavalle et al., 2008; Opala-Berdzik et al., 2018, 2021).

The performance of acrobatic gymnastics focuses on static and dynamic routines which demand high levels of strength, stability, flexibility and agility, and hence training programs are designed to enhance these features thereby improving overall performance. Additionally, practicing also focuses on the presentation of specific routines in which the athletes work together in an attempt to have their performance well ranked by referees. Consequently, the duration, frequency, and intensity of the training sessions are high, and often begin in early childhood. Therefore, one of the main issues coaches have to deal with concerns the deleterious effect of fatigue (that occurs during long and intense training sessions) on the athletes' performance. In particular, the impact of fatigue on standing balance is of great importance as the overall stability during the static and dynamic routines are highly associated with the athletes' performance (i.e., observed by referees during competitions). In this vein, fatigue has been considered of particular importance to sports-related injury risk and performance, and the negative effect of neuromuscular fatigue on postural control has been extensively reported in a wide variety of conditions and for different populations (Paillard, 2012). Traditionally, many studies have shown increased postural oscillations (and hence decreased steadiness) during quiet standing after the performance of a fatiguing exercise, especially involving the ankle musculature (Gribble and Hertel, 2004; Boyas et al., 2011, 2013; Bisson et al., 2012).

Surface electromyography (sEMG) has been widely used to evaluate neuromuscular adaptations as a function of fatigue (Rainoldi et al., 2008; Contessa et al., 2009; Soo et al., 2009; Dideriksen et al., 2010), and also to better understand the mechanisms associated with muscle activation during postural control (Magalhaes and Goroso, 2009; Mello et al., 2013). The fatigue-related increase in time domain parameters of the sEMG signals (i.e., amplitude, RMS) during sustained contractions has been attributed to an increase in neuromuscular activation, promoting greater recruitment of motor units to compensate for the saturation of fibers that are already fatigued, avoiding the immediate failure of the system (De Luca, 1997). On the other hand, fatigue-related changes on sEMG frequency-domain parameters [i.e., increase in low frequency components and decrease in high frequency components (Merletti et al., 1991; Lowery et al., 2002)] have been attributed to two main mechanisms: (1) peripheral changes, especially the decrease in the conduction velocity of muscle fiber action potentials and; (2) central changes including increased synchronism in the firing of motor units along with the recruitment of new motor units (Mathur et al., 2005).

However, despite the large number of studies mentioned above, to the best of our knowledge, no study has addressed the putative effects of neuromuscular adaptations resulting from acrobatic gymnastics training on the behavior of the postural control system in the presence of muscle fatigue. Additionally, presumed neuromuscular mechanisms reflected by changes in time and frequency-domain parameters of the EMG signal recorded during postural tasks (and its fatigue-related effects) remain unexplored. Such an investigation has the potential to provide insights that might be useful for the design of gymnastics training strategies, besides improving our knowledge on the mechanisms behind the interplay between fatigue and postural performance associated with the neuromuscular adaptations induced by sport practice. More specifically, a deeper understanding of the long-term effects of gymnastics training on postural control, together with information on how the neuromuscular system of experienced athletes adapt to overcome the deleterious effect of fatigue on postural stability might help coaches to focus on specific training strategies (such as strength/endurance/control training of specific muscle groups) that can potentially benefit their athletes in improving performance and/or reducing injury risk.

Therefore, the objective of this study was to investigate the effects of fatigue and acrobatic gymnastic expertise on the neuromuscular and behavioral performance of a unipedal quiet stance task. So, we assessed whether expert acrobatic gymnasts respond differentially than their non-trained counterparts during a single-legged stance task (performed before and after a protocol designed to induce fatigue in the ankle plantarflexors muscles) in terms of (a) postural steadiness and (b) EMG activation. The single-legged postural task was chosen because it is a posture commonly adopted in the practice of acrobatic gymnastics (present in the punctuation code considered by referees in competitions). The hypothesis addressed in the present study was that neuromuscular adaptation due to gymnastics training would lead to different behavior of postural sway and EMG quantifiers (acquired during unipedal quiet stance) in trained gymnasts submitted to neuromuscular fatigue of the plantarflexor muscles as compared to healthy untrained controls. More specifically, we expected that fatigue would reduce postural stability (as assessed by COP parameters) and change EMG activation patterns (increasing time-domain and decreasing frequency-domain parameters) in both expert acrobatic gymnasts and their non-trained counterparts; and that these changes would be more pronounced in the non-trained group, as the gymnasts would putatively have developed training-induced adaptation mechanisms that would help them in dealing with the deleterious effects of fatigue on postural control.

MATERIALS AND METHODS

Participants

The experiments were carried out in 28 female volunteers, aged between 11 and 24 years. Two groups were formed: acrobatic gymnastics athletes (GYN, n = 14) and non-gymnasts [control (CTRL), *n* = 14]. Participants' age, height, and body mass are depicted in **Table 1**. The gymnasts were recruited from a highly reputed training center at Guarulhos, São Paulo. The CRTL group was composed of participants with age-matched to those in the GYN group. The inclusion criteria for the GYN group were being active gymnasts of acrobatic gymnastics currently training for participation in state championships, national league or higher (i.e., pan American and World class), with training routines of at least 15 h/week. The inclusion criteria for CTRL group were no involvement in competitive sports (they typically had their physical activity profile based on physical education classes at school or playing unprofessionally at leisure time). For both groups, exclusion criteria were obesity, musculoskeletal injuries, or vision/neurological disorders that might have affected their ability to maintain balance. At the time of data acquisition, the acrobatic gymnasts were involved in practicing routines 5–6 times per week for 3–5 h (data were acquired 2–3 weeks before the national championship, in which most of them were about to compete).All gymnasts had begun their training experience at least 3 years before the experiments. The experiments were approved by the Human Ethics Committee of the School of Arts, Sciences and Humanities of the University of São Paulo (CAAE69034717.6.0000.5390).

TABLE 1 | Characteristics (age, height and body mass) of the participants included in the Acrobatic Gymnastics group (GYN, *n* = 14) and age-matched non-gymnasts [controls (CTRL), *n* = 14].

	CTRL	GYN	$T_{(26)}$	P
Age (years)	16.3 ± 3.2 (12–24)	16.1 ± 3.1 (11–23)	0.121	0.905
Body height (cm)	161.4 ± 77.6 (146–174)	156.1 ± 85.6 (147–170)	1.734	0.095
Body mass (kg)	56.7 ± 10.3 (41–74)	53.0 ± 10.3 (30–66)	0.956	0.348

Electromyography Acquisition

Surface EMG signals were acquired with a DELSYS system, model Trigno-16. EMG signals were captured using surface electrodes positioned (at "preferred" side, corresponding to the support leg in the postural tests, see below) as recommended by SENIAM (Hermens et al., 2000) in *tibialis anterior* (TA), *soleus* (SO), *lateral gastrocnemius* (GL), *medial gastrocnemius* (GM), *vastus lateralis* (VL), and *biceps femoris* (BF) muscles. For the trunk muscles [*spinal erector* (ES) and *rectus abdominis* (RA)], the positioning of the electrodes replicated protocols commonly used in the literature. Thus, RA electrodes were positioned 3 cm laterally to the navel (Horak and Nashner, 1986; Runge et al., 1999; McCook et al., 2009); and in the ES muscle, the electrodes were positioned 3 cm laterally to the spinous process of the third lumbar vertebra (Hodges et al., 2002; McCook et al., 2009). Before electrodes positioning, proper skin preparation was performed, including trichotomy, abrasion and cleaning with alcohol.

Maximal Isometric Voluntary Contraction Assessments

Figure 1 shows a simplified representation of the experimental procedures. Initially, participants warmed-up on a cycle ergometer for 5 min at low intensity. Then, they performed three maximum isometric voluntary contractions (MIVCs) for each muscle under evaluation, consisting of sustained maximal contractions of approximately 5 s (with verbal encouragement), in which the highest amplitude value of surface EMG (aEMG, see calculation method below) was obtained in a 2-s window. The maximal aEMG observed among the three MIVCs was used to normalize the EMG intensity during the postural tasks

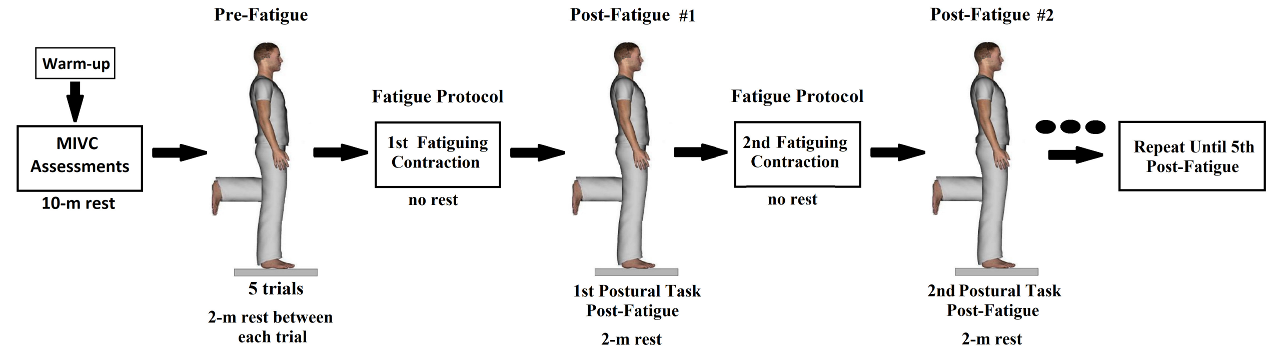

FIGURE 1 | Simplified representation of the experimental protocol.

(see below). Then, participants had a 10-min rest, seated in a comfortable armchair, before the initial postural assessments.

For MIVC assessments, participants laid (or seated) on a customized stretcher equipped with Velcro tapes and adjustable mechanical levers that prevented undesirable movements of the participants. MIVC of the SO, GL, and GM were obtained with the participants in the prone position, with feet free off the stretcher and ankle at 90°, with the area of the heads of the metatarsal bones in contact with a fixed restraint. Participants were asked to exert as much force as possible to push against the restraint with the foot sole using only leg muscles. A mechanical support was positioned against the participants' shoulders to prevent them to slip on the stretcher. TA MIVC was obtained with the participants in the supine position, the ankle at 90° and the dorsal part of the foot in contact with a fixed restraint. Participants were asked to pull the restraint with the dorsal part of the foot as hard as possible using only leg muscles. MIVCs of the VL and BF muscles were obtained with the participant seated on the edge of the stretcher with hip, knee, and ankle angles at 90°. A fixed restraint was positioned against the anterior and the posterior aspects of the legs, at ankle level. For the VL, participants were asked to push the anterior restraint forward, with maximum possible force using only thigh muscles, whereas for the BF they were asked to pull the posterior restraint backward. For the RA MIVC, the participants were in the supine position with a fixed restraint positioned over their chest. The participants were asked to exert as much force as possible with only the abdominal muscles in order to try to "lift the restraint." For the MIVC of the ES, participants were placed in the prone position with a fixed restraint placed over their back (at the level of shoulder blades). Participants were asked to exert the maximum force possible to try to "lift the restraint" using only back muscles. During the MIVCs verbal stimuli were used to motivate participants to exert the maximum possible force.

Postural Assessments (Pre-fatigue)

Participants were asked to maintain a single-leg stance (unipedal support), with the arms at the side, standing barefoot as quietly as possible over a force plate (OR6-7-1000, AMTI, Watertown, United States). The support leg was the participant's preferred leg, defined by asking which leg they would use to kick a ball. The knee of the stance leg was in an extended position (but not hyperextended), whereas the hip of the contralateral leg was in a neutral position and the knee was flexed at 90°. Participants looked at a fixed point (i.e., a black circle marked with adhesive tape) at the wall ∼4 m away from them, at eyes level. The position of the subject's feet on the platform was marked with adhesive tape to ensure that the same position was repeated on each trial. The postural tasks were performed with eyes opened. Each participant performed five trials, each lasting 60 s. A resting period of ∼2 min between trials was allowed (subject sat in a comfortable armchair placed next to the force plate).The single-legged postural task was chosen because it is a posture commonly adopted in the practice of acrobatic gymnastics (present in the punctuation code considered by referees in competitions).

Fatigue Protocol

Then, plantar flexor muscles of the dominant leg were subjected to muscle fatigue by methodology previously described in the literature (Vuillerme et al., 2001b; Boyas et al., 2013), which consisted of sustained submaximal isometric contractions. The participants were asked to remain in a single-legged posture, "on their toes," for as long as possible, without performing flexion-extension movements. In other words, the participants had to lift the heel, thereby supporting themselves "on the tip of the foot" with the greatest possible ankle extension by sustaining an isometric plantarflexion force. During the maintenance of this posture, the participants were allowed to lean on a structure positioned in front of them, whenever necessary (i.e., due to any type of imbalance). In addition, an experimenter stayed by the participant's side during the fatigue protocol, to ensure that no major imbalances occurred. The fatigue condition (i.e., each fatiguing contraction) was considered completed when the participant could no longer maintain this posture "on tiptoe" (i.e., failure, when participants' heel touched the ground). Verbal stimuli were used to motivate participants to stay until exhaustion. Time to fail was measured [time to failure (in seconds), from the beginning of the fatiguing contraction until failure]. The protocol (fatiguing contraction) was repeated before each trial of single-legged stance performed under the fatigue condition (see below), yielding 5 fatiguing contractions for each participant.

Postural Assessments (Post-fatigue)

Post-fatigue postural assessments were similar to those described initially (pre-fatigue). However, as a transient effect of fatigue-inducing protocols on postural control has been reported (Paillard, 2012; Kozinc et al., 2021), the fatigue protocol (submaximal contraction until exhaustion, as described above) was repeated before each trial to ensure the participants performed the postural task under the effects of muscle fatigue (Paillard, 2012). For each of the 5 repetitions, the postural task was initiated immediately after the fatiguing contraction was finished, thereby ensuring the presence of the fatigue effect in all post-fatigue trials.

Data Processing and Analyses

The signals from the EMG and the force plate were acquired by an A/D board (PCI-6015, National Instruments, United States) at 2000 samples/s. Data were analyzed off-line using custom-written programs in Matlab (Math Work Inc., United States).

The EMG signals were pre-amplified and filtered with cutoff frequencies of 20 and 450 Hz. In order to obtain the EMG parameters during the postural tasks, the raw EMG signals were filtered by a 4th-order digital Butterworth bandpass filter (with cutoff frequencies at 20 and 450 Hz). The parameters in the time (EMG amplitude, aEMG) and frequency (median frequency, Fmed) domains were then calculated, with Fmed computed in the frequency spectrum using the Discrete Fourier Transform. For aEMG, the RMS of the EMG signal was computed for the entire period over the postural tasks (60s), for every 250-ms epochs (time window), and averaged among the five trials performed for each condition (i.e., pre and post-fatigue) (Raez et al., 2006; Keenan and Valero-Cuevas, 2008). Then, aEMG values were normalized to the aEMG observed during the MVIC, and presented herein as a percentage of the MVIC, as suggested in the literature (Lehman and McGill, 1999). Muscle activity during MVIC was computed for every 250-ms epochs and averaged. This epoch (250-ms time window) was chosen in order to enhance EMG signal resolution and estimate more accurately aEMG during the postural tasks, which is in line to what has been used in the literature (Raez et al., 2006; Keenan and Valero-Cuevas, 2008). In addition, the same epoch was used in the MVIC to avoid discrepancies in the method of analyzing the data.

The forces and moments measured by the force plate were used to compute the two components of the center of pressure (COP): in the anterior-posterior axis (AP) and the media-lateral axis (ML), indicated as Copa and Comply, respectively. Before analysis, the COP data (acquired at 2 kHz) were low-pass filtered with a digital fourth-order Butterworth filter having an 8 Hz cutoff frequency and the mean was subtracted from each time series. The root-mean-square (RMS) and mean velocity (VM) of the COP data were computed for each axis (i.e., AP and ML). The area of the stabilogram (Area) was estimated from the COP data by fitting an ellipse to the AP x ML COP data that encompasses 95% of the data [using the method proposed by Oliveira et al. (1996)]. The COP velocity was calculated by dividing the total COP displacement (sum of the absolute values of the samples) by the total time interval. All time-domain parameters computed from the COP signals were normalized by participants' height (Cattagni et al., 2014), and hence are expressed here as a percentage of participants' height. The power spectral density (PSD) of the COP data (for both AP and ML directions) was estimated in each experimental condition (pre and post-fatigue). The power spectral density was estimated by the Welch periodogram of the detrended data with a resolution of 0.05 Hz. A Hann datawindow was used with subtraction of the best linear regression and a 50% overlap. The area under the PSD was calculated in order to represent the power at two frequency bands: "low frequencies" (LF, 0.05–0.5 Hz) and "high frequencies" (HF, 0.5–2.0 Hz) (Magalhaes and Kohn, 2011). These ranges were chosen as within the adopted low-frequency range (LF), the COP power spectrum approximates that of the center of gravity (Benda et al., 1994), whereas the upper limit of the high-frequency band (2Hz) delimits the frequency range that encompasses 99% of the total power of the COP signal during quiet stance (Mezzarane and Kohn, 2008).

Statistical Analysis

Participants' age, height, and body mass were compared between GYN and CTRL groups using *t*-tests for independent samples. COP and EMG parameters (aEMG, Fmed, Area, RMS, VM, HF and LF) were averaged across the 5 repetitions of the postural tasks (for both pre- and post-fatigue assessments). After applying a statistical test to verify the normality of the data (Shapiro-Wilk test), a two-way ANOVA was used to compare the duration of the fatiguing contractions between groups (GYN *vs* CTRL), and among the 5 repetitions (factor "time," representing the 1st to the 5th fatiguing contraction), with "time" considered as repeated measures. A two-way ANOVA was also used to compare EMG and COP parameters between groups (GYN *vs* CTRL), and between conditions (pre-fatigue *vs* post-fatigue), with this last factor considered as repeated measures. Eventual interactions between factors were obtained by the same test. All the analyses were performed with a significance level set at $p < 0.05$. Effect sizes are reported using partial eta-squared (η^2_p) indices (Tabachnick and Fidell, 2007), which indicate the percent of the variance in each of the effects (or interaction) that is accounted for by that effect (or interaction). Effect sizes are interpreted as small ($\eta^2 = 0.01$), medium ($\eta^2 = 0.06$), and large ($\eta^2 = 0.14$) effects.

RESULTS

Participants Characteristics

As expected, there were no significant differences between CTRL and GYN for age, as CTRL group were age-matched to GYN. Despite not being matched for body height and body mass, no significant differences between CTRL and GYN were observed for these parameters (see **Table 1**).

Time to Failure (Duration of the Fatiguing Contraction)

Figure 2 shows the average duration of each of the 5 fatiguing contractions performed by the participants immediately before

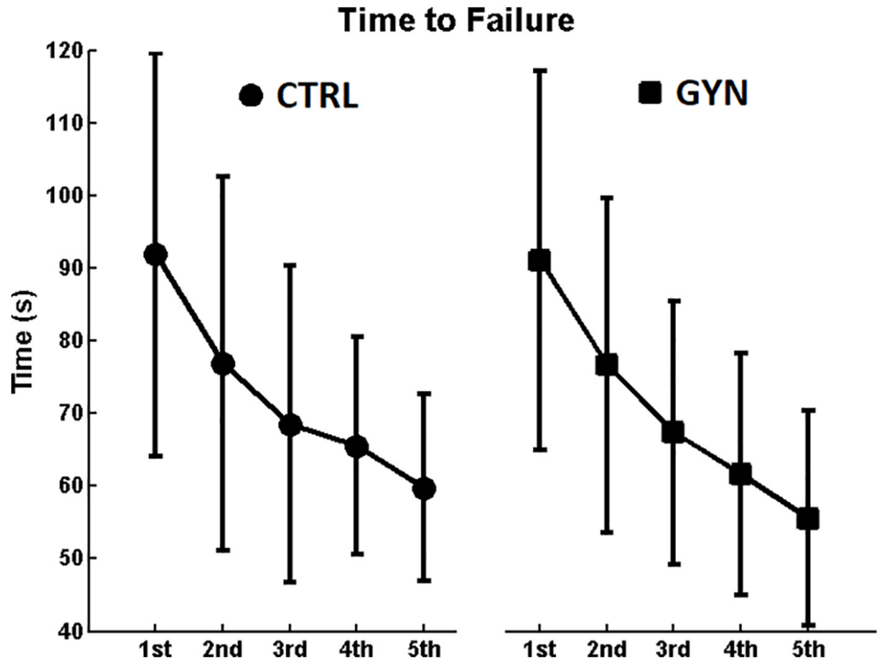

FIGURE 2 | Average duration (time to failure) of each of the 5 fatiguing contraction performed by the participants. Values associated with the controls (CTRL) and gymnasts (GYN) groups are represented by circles (left panel) and squares (right panel), respectively. Note the clear tendency to decrease time to failure (indicating a reduction in endurance performance) along the repetitions, yielding a significant main effect of time ($P < 0.001$). No significant main effect of group was observed neither significant interactions between group and time ($P > 0.05$), indicating that gymnasts showed similar performance during the fatiguing protocol as compared to the controls.

each of the post-fatigue postural tasks (see methods). A clear tendency to decrease time to failure (indicating a reduction in endurance performance) along the repetitions was observed, as assessed by a significant main effect of time [$F_{(1,26)} = 46.1191$; $P < 0.001$; $\eta^2_p = 0.26$]. No significant main effect of group was observed [$F_{(1,26)} = 0.0856$; $P = 0.772$; $\eta^2_p = 0.0027$], neither significant interactions between group and time [$F_{(1,26)} = 0.2284$; $P = 0.922$; $\eta^2_p = 0.0017$], which indicates that gymnasts showed similar performance during the fatiguing protocol (in terms of duration of the fatiguing contractions) as compared to the controls, along the 5 repetitions of the fatiguing contractions.

Center of Pressure Analyses (Time and Frequency Domains)

Figure 3 shows the average values of the COP quantifiers (i.e., COP parameters in time and frequency domains) for the CTRL and GYN groups, both before (NF, no fatigue) and after (PF, post fatigue) the fatigue protocol. A significant main effect of fatigue was found for Area, RMSml, VMml, LFml and HFml. These results indicate that the aforementioned parameters were increased after fatigue as compared to pre-fatigue, irrespective of the gymnastic expertise (CTRL *vs* GYN). Therefore, fatigue induced a significant increase in postural oscillations (including velocity and frequency components of the power spectrum), mainly in the ML axis. No significant main effect of fatigue was observed for the parameters that represent the COP exclusively in the AP axis. No significant group effects (CTRL *vs* GYN) were found for COP parameters, neither significant interactions between fatigue and group, which indicates that gymnasts showed similar postural sway as compared to the controls, during both pre and post-fatigue assessments. **Table 2** shows the statistical results (F, P, and η^2) obtained from the two-way ANOVA for the main effects and interactions observed for COP quantifiers.

Amplitude of the Electromyography Signal (Electromyography Amplitude)

Figure 4 shows the average values of aEMG for the CTRL and GYN groups, both before (NF, no fatigue) and after (PF, post fatigue) the fatigue protocol. A significant main effect of fatigue was found for the muscles *soleus* (SO), *lateral gastrocnemius* (GL), and *biceps femoris* (BF). These results indicate that the aEMG recorded from these muscles were significantly increased after fatigue as compared to pre-fatigue, irrespective of the gymnastic expertise (CTRL *vs* GYN). No significant main effect of fatigue was observed for the remaining muscles. A significant group effect (CTRL *vs* GYN) was found for the aEMG of the TA muscle, indicating that the GYN group showed lower TA EMG activation as compared to the CTRL group. No significant main effect of group (CTRL *vs* GYN) was observed for aEMG of the remaining muscles, neither significant interactions between fatigue and group. **Table 3** shows the statistical results (F, P and η^2) obtained from the two-way ANOVA for the main effects and interactions observed for aEMG.

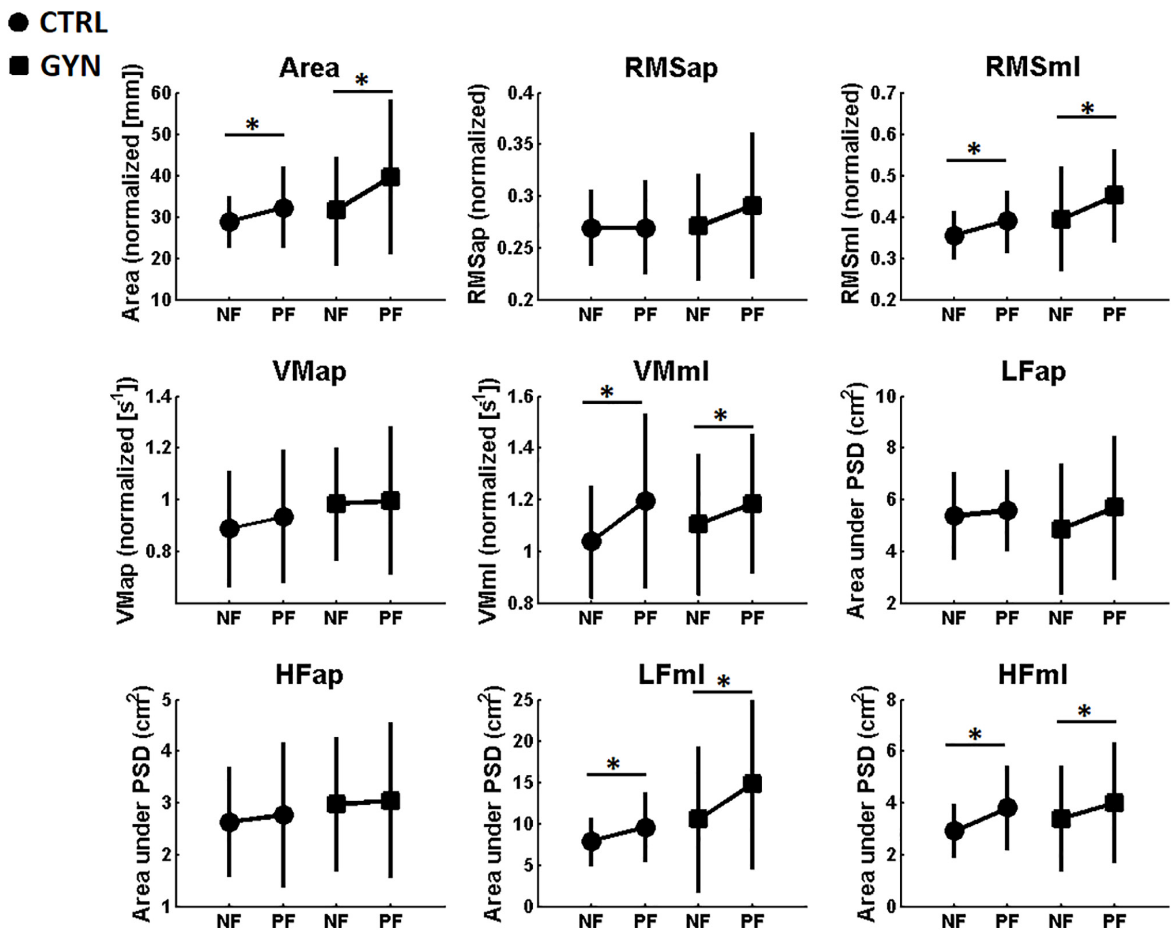

FIGURE 3 | Average center of pressure (COP) measures [associated with the AP (anterior posterior) and ML (medial-lateral) axis] computed before (NF, no fatigue) and after (PF, post-fatigue) the fatiguing protocol. Values associated with the CTRL and GYN groups are represented by circles and squares, respectively. * indicate a statistically significant effect of the fatigue factor, that is, significant differences between NF and PF ($p < 0.05$). Area: area of the area of the stabilogram, RMS: root mean square, VM: COP mean velocity, LF: low frequencies (LF, 0.05–0.5 Hz), HF: high frequencies.

TABLE 2 | Statistical results of the two-way analysis of variance (ANOVA) applied to the center of pressure (COP) parameters before and after the fatiguing protocol.

COP	Main effects and interactions								
	Group (CTRL *vs* GYN)			Fatigue (NF *vs* PF)			Fatigue *vs* Group		
	$F_{(1,26)}$	P	η^2_p	$F_{(1,26)}$	P	η^2_p	$F_{(1,26)}$	P	η^2_p
Area	1.3264	0.2599	0.042	10.4663	**0.0033**	0.052	1.6927	0.2047	0.0088
RMSap	0.3883	0.539	0.013	2.114	0.158	0.0099	2.0836	0.161	0.0097
RMSml	2.3033	0.1412	0.069	10.1359	**0.0037**	0.058	0.6227	0.437	0.038
VMap	0.7729	0.3874	0.028	1.9625	0.1731	0.0035	0.8307	0.3704	0.0015
VMml	0.0794	0.7804	0.0029	22.1234	**<0.001**	0.047	2.4681	0.1283	0.0055
LFap	0.070	0.793	0.0023	2.885	0.101	0.016	1.124	0.298	0.0062
LFml	2.423	0.131	0.079	15.287	**<0.001**	0.046	2.871	0.102	0.0089
HFap	0.447	0.509	0.015	0.343	0.563	0.0016	0.035	0.851	0.0002
HFml	0.261	0.613	0.0088	11.197	**0.002**	0.047	0.310	0.0014	0.5821

Significant P values ($P < 0.05$) are highlighted in bold, indicating significant main effects of fatigue (no fatigue, NF vs post-fatigue, PF). Area: area of the stabilogram, RMS: root mean square, VM: COP mean velocity, LF: low frequencies (LF, 0.05–0.5 Hz), HF: high frequencies.

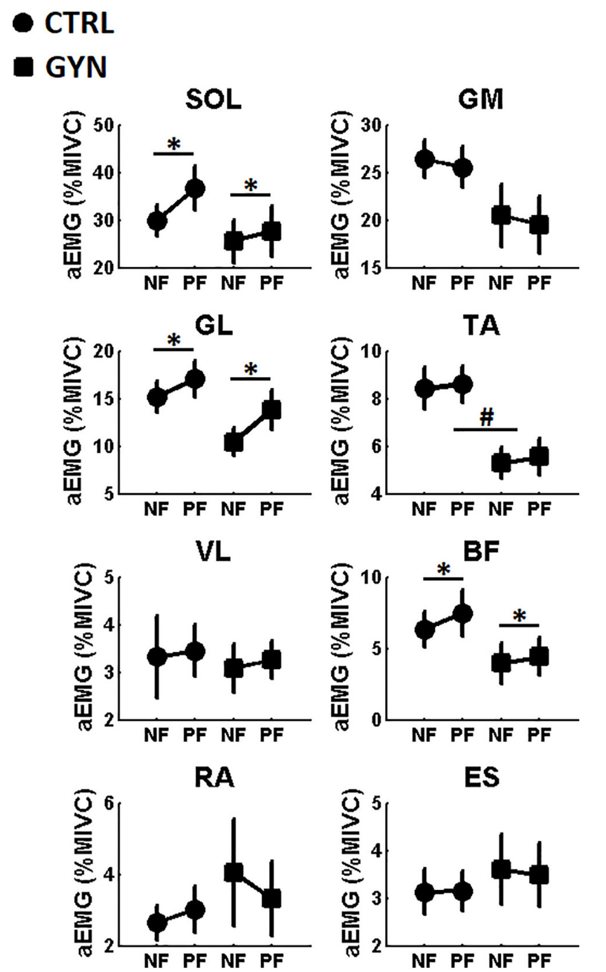

FIGURE 4 | Average amplitude of the EMG signal (aEMG, expressed in% of the Maximal Isometric Voluntary Contraction, MIVC) computed before (NF, no fatigue) and after (PF, post-fatigue) the fatiguing protocol. Values associated with the CTRL and GYN groups are represented by circles and squares, respectively. * indicate a statistically significant effect of the fatigue factor, that is, significant differences between NF and PF ($p < 0.05$). # indicate a statistically significant effect of the group factor, that is, significant differences between acrobatic gymnasts (GYN) and untrained controls (CTRL). TA: *tibialis anterior*, SOL: *soleus*, GL: *lateral gastrocnemius*, GM: *medial gastrocnemius*, VL: *vastus lateralis*, BF: and *biceps femoris,* ES*: spinal erector* and RA: *rectus abdominis*.

Electromyography Median Frequency

Figure 5 shows the average values of Fmed for the CTRL and GYN groups, both before (NF, no fatigue) and after (PF, post fatigue) the fatigue protocol. A significant main effect of fatigue was found for the muscles *lateral gastrocnemius* (GL), *vastus lateralis* (VL), *biceps femoris* (BF), *rectus abdominis* (RA) and *spinal erector* (ES). These results indicate that the Fmed of the EMG recorded from these muscles were decreased after fatigue as compared to pre-fatigue, irrespective of the gymnastic expertise (CTRL *vs* GYN). No significant main effect of fatigue was observed for the remaining muscles. A significant group effect (CTRL *vs* GYN) was found for the Fmed of the SOL, TA and BF muscles, indicating that the GYN group showed higher Fmed in SOL EMG signal, but lower Fmed in TA and BF EMG signals as compared to the CTRL group. No significant main effect of group (CTRL vs GYN) was observed for the remaining muscles. Interestingly, a significant interaction between fatigue and group was observed for Fmed of the ES muscle, indicating a less pronounced effect of fatigue (i.e., decreased Fmed) for the GYN group as compared to the CTRL group. No significant interactions between fatigue and group were found for Fmed of the remaining muscles. **Table 4** shows the statistical results (F, P, and η^2) obtained from the two-way ANOVA for the main effects and interactions observed for Fmed.

DISCUSSION

The present study compared the performance of single-legged stance and the associated modulation of time and frequency-domain EMG parameters between trained gymnasts and healthy untrained controls before and after a protocol designed to induce muscle fatigue in the ankle plantarflexor muscles. As hypothesized, fatigue-induced increased postural sway, while decreasing Fmed and increasing aEMG, irrespective of the gymnastics expertise. We had also hypothesized that fatigue-induced effects on COP and EMG parameters would be more pronounced in the non-trained group as compared to GYN, as the trained gymnasts would have counteracted (at least partially) the deleterious effects of fatigue by using neuromuscular adaptations developed as a result of their intense and regular practice. This last hypothesis was only partially confirmed (i.e., only confirmed in terms of EMG activation), as no superior stability was found for the GYN group as compared to CTRL in terms of postural sway parameters (i.e., COP quantifiers), irrespective of the fatigue condition (see **Figure 3**). On the other hand, expert acrobatic gymnastics used different neuromuscular control strategies to keep their postures on single-legged quiet standing [i.e., with lower aEMG of the TA muscle (see **Figure 4**), higher Fmed of the SO muscle, and lower Fmed of TA and BF muscles (see **Figure 5**)]. Notable, a more pronounced effect of fatigue on the ES muscles (i.e., decreased Fmed) was observed for the CTRL group as compared to the GYN group.

The fatigue protocol (submaximal contractions until exhaustion) was repeated before each trial associated with the post-fatigue condition to warrant that participants performed the postural assessments still under the effects of muscle fatigue (Paillard, 2012). The duration of the fatiguing contractions significantly decreased along with the repetitions (see **Figure 2**), irrespective of gymnastics expertise, indicating that the endurance performance of the ankle plantarflexor muscles tended to decrease with repetition, thereby suggesting the fatigue protocol was effective. Additionally, significant increases in aEMG values and significant decreases in Fmed values were observed after the fatigue protocol (see **Figures 4**, **5**) as compared to baseline assessments, which are in accordance with the most pertinent literature (Merletti et al., 1991; Lowery et al., 2002; Mathur et al., 2005), as mentioned in the Introduction section. Altogether, these results indicate that the fatigue protocol

TABLE 3 | Statistical results of the two-way ANOVAs applied to the measurements of the amplitude of the EMG signal (aEMG) before and after the fatiguing protocol.

aEMG	Main effects and interactions								
	Group (CTRL *vs* GYN)			Fatigue (NF *vs* PF)			Fatigue *vs* Group		
	$F_{(1,26)}$	P	η^2_p	$F_{(1,26)}$	P	η^2_p	$F_{(1,26)}$	P	η^2_p
SOL	1.220	0.279	0.043	9.588	**0.005**	0.019	2.504	0.126	0.005
GM	3.010	0.0946	0.090	0.408	0.528	0.0023	0.001	0.974	<0.0001
GL	3.053	0.0923	0.092	8.265	**0.008**	0.041	0.713	0.405	0.0037
TA	10.055	**0.004**	0.250	0.292	0.593	0.0016	0.013	0.907	<0.0001
VL	0.084	0.774	0.0026	0.164	0.688	0.0012	0.004	0.949	<0.0001
BF	2.003	0.168	0.069	4.365	**0.046**	0.0062	0.852	0.364	0.0012
RA	0.397	0.534	0.014	0.234	0.632	0.0007	2.230	0.147	0.0061
ES	0.260	0.613	0.0094	0.075	0.785	0.0001	0.143	0.707	0.0003

Significant P values (P < 0.05) are highlighted in bold, indicating significant main effects of fatigue (no fatigue, NF vs post-fatigue, PF) and/or group (control, CTRL vs acrobatic gymnasts, GYN). TA: tibialis anterior, SOL: soleus, GL: lateral gastrocnemius, GM: medial gastrocnemius, VL: vastus lateralis, BF: and biceps femoris, ES: spinal erector and RA: rectus abdominis.

used in the present study proved to be effective, representing an appropriate methodology for the comparisons between experimental conditions. Further analyses regarding time and frequency-domain EMG parameters recorded during the fatiguing protocol were previously published as a conference paper (Silva et al., 2021).

The duration of the fatiguing contractions was not significantly different between groups, and no significant interactions between group and time were observed, which indicate that gymnasts showed similar performance during the fatiguing protocol (in terms of duration of the fatiguing contractions) as compared to CTRL, along with the 5 repetitions of the fatiguing contractions. Therefore, all putative mechanisms associated with the differential effects of fatigue on COP and EMG measurements (discussed below) between GYN and CTRL are very probably associated with the gymnastics expertise itself, rather than a mere consequence of a different fatigue protocol.

Center of pressure (COP) analyses showed that fatigue induced a significant increase in postural oscillations in the ML axis, with no significant effects in the AP axis, whereas no superior stability was found for the GYN group as compared to CTRL in terms of postural sway parameters (i.e., COP quantifiers), irrespective of the fatigue condition (see **Figure 3**). These results are in line with previous studies that reported no significant differences in some postural sway parameters between gymnasts and non-gymnasts (Vuillerme et al., 2001a; Asseman et al., 2004, 2008; Vuillerme and Nougier, 2004; Gautier et al., 2008; Opala-Berdzik et al., 2021). As these studies used relatively non-challenging postural tasks (bipedal eyes-open posture), it has been suggested that the effect of gymnastics expertise on balance ability might be associated with specific postures for which the practice is specifically related (Asseman et al., 2008), as gymnasts' training routines (which involve difficult and specific postures) might have a slight or no effect on the performance of unspecific postures. Thus, in the present study, we chose the unipedal stance (performed with eyes open) as the postural task because it represents a more challenging and specific posture associated with acrobatic gymnastics training. However, unlike previous studies that reported significant differences between gymnasts and controls on some postural sway parameters for this condition (Asseman et al., 2008), our study did not find superior balance stability for the GYN. These different results might be associated with the different samples used in the studies, as Asseman et al. (2008) investigated male artistic gymnasts (and compared them with other sportsmen), whereas the present study investigated younger female expert acrobatic gymnastics as compared to non-trained counterparts. Indeed, expertise (Opala-Berdzik et al., 2021), sex (Milosis and Siatras, 2012) and training specificity (Vuillerme et al., 2001a; Asseman et al., 2004, 2008; Isableu et al., 2017) have been shown to significantly influence the development of postural control abilities.

As expected, fatigue induced a significant increase in postural oscillations, irrespective of the gymnastics expertise. Interestingly, such an effect was only observed for the postural parameters computed for the ML axis (as no significant main effect of fatigue was observed for the parameters that represent the COP exclusively in the AP axis). This might be related to the nature of the unipodal balance task used in the present study, in which the body oscillations occur mainly along the ML axis. Calavalle et al. (2008) showed superior balance stability during quiet bipedal posture in rhythmic gymnasts as compared to controls trained in other sports, which only occurred in the ML direction. However, these results cannot be directly compared as they relied in different samples and postures, and the effect of fatigue was not explored in the study of Calavalle et al. (2008).

A previous study (van Dieen et al., 2012) showed that fatigue (i.e., induced by exercising trunk muscles) negatively affected trunk stability in elite gymnasts (for both unperturbed balancing and recovery after balance perturbation), but no comparisons were made with other groups so that we cannot speculate whether gymnasts expertise might be associate with a differential ability in dealing with the detrimental effects of fatigue on trunk stability. To the best of our knowledge, the present study is the first to investigate the fatigue-related effects on unipedal postural control of expert acrobatic gymnasts as compared to untrained matched controls. Contrary to our initial hypothesis,

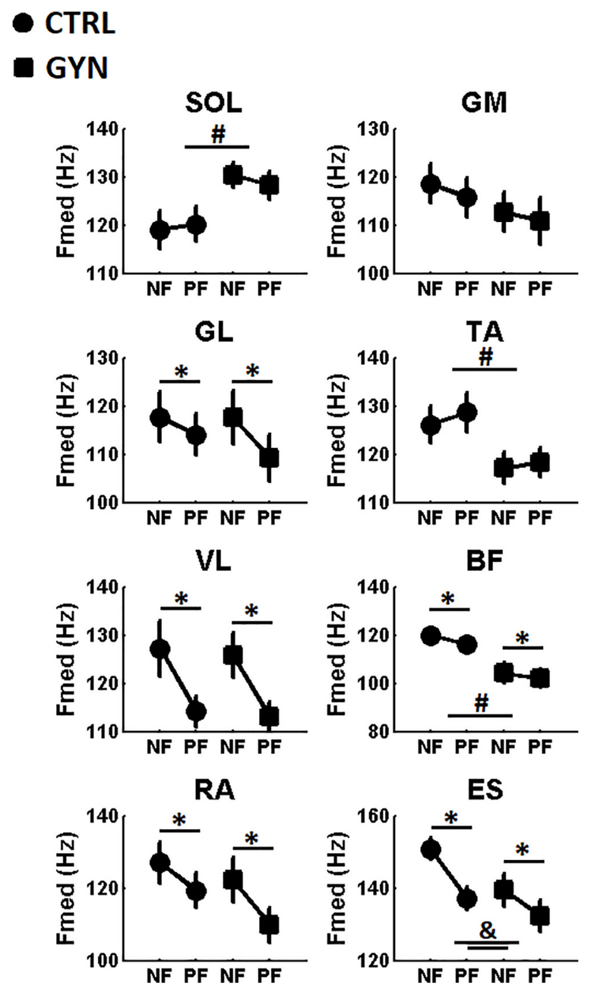

FIGURE 5 | Average Median Frequencies (Fmed, expressed in Hz) computed before (NF, no fatigue) and after (PF, post-fatigue) the fatiguing protocol. Values associated with the CTRL and GYN groups are represented by circles and squares, respectively. * indicate a statistically significant effect of the fatigue factor, that is, significant differences between NF and PF ($p < 0.05$). # indicate a statistically significant effect of the group factor, that is, significant differences between acrobatic gymnasts (GYN) and untrained controls (CTRL). & indicate significant interaction between "group" and "fatigue period" factor. & indicates a statically significant effect for "group" factor ($p < 0.05$). TA: *tibialis anterior*, SOL: *soleus*, GL: *lateral gastrocnemius*, GM: *medial gastrocnemius*, VL: *vastus lateralis*, BF: and *biceps femoris,* ES*: spinal erector* and RA: *rectus abdominis*.

the effects of fatigue in reducing balance steadiness during the single-legged stance task were comparable between GYN and CTRL (i.e., neuromuscular adaptations due to acrobatic gymnastics training was not able to significantly counteract for the detrimental effects of fatigue), at least regarding the COP-related postural sway measurements explored in the present study. Isableu et al. (2017) showed that non-linear dynamical analysis of COP trajectories might be useful to provide a deeper understanding of the mechanism behind sports-related adaptations on postural control mechanism, as the analysis of the COP regularity (i.e., sample entropy measurements) could distinguish between the bipedal quiet stance performance of gymnasts and non-gymnasts, which could not be evidenced by the traditional sway parameters. Therefore, future studies should focus on deeper analyses of postural sway parameters to explore whether gymnastics expertise might be associated with the differential ability to deal with muscle fatigue.

The comparisons between the EMG parameters recorded before and after the fatigue protocol indicate that the effect of fatigue was not restricted to the target muscles (plantar flexors), considering that, except for SOL, GM, and TA, all Fmed values decreased significantly after the fatigue protocol (which includes tight and trunk muscles) and aEMG of the BF muscle significantly increased after the fatigue protocol (besides SOL and GM). These results reinforce the notion that the fatigue effect associated with a specific contraction might be extended beyond the target musculature (McDonald et al., 2019; Goethel et al., 2020). This has important implications from a practical standpoint as it suggests that the performance of different gestures might be impaired after the execution of a task with little or no similarity to the one that induced fatigue. For instance, coaches, athletes, and referees must be aware that there might be a reduction in performance (and probably an increased injury risk) induced by a fatigue process generated by a previous exercise/routine, even when the task performed might seem fairly unrelated to the previous one.

Acrobatic gymnastics showed lower TA aEMG as well as lower TA and BF (but higher SOL) Fmed as compared to the CTRL group. Finally, it is worth noting that the Fmed of the ES muscle decreased more sharply after the fatigue protocol for the CTRL group (evidenced by the significant interaction between the factors "fatigue" and "group," see **Figure 5** and **Table 4**), indicating a greater effect of fatigue of the ES for the CTRL group as compared to the GYN group. These results suggest that trained acrobatic gymnasts develop specific neuromuscular strategies to deal with balance maintenance on single-legged stance (albeit these differential strategies seem to not influence their postural oscillations, at least in terms of the COP parameters analyzed herein). Higher levels of TAEMG activity during postural tasks have been traditionally associated with less stable situations, such as during the control of quiet bipedal stance performed by older adults (as compared to younger ones) (Kouzaki and Shinohara, 2010), which probably reflects more frequent corrections on sway trajectory to keep the body still. Therefore, the lower EMG activation (aEMG) of the TA muscle shown by the gymnasts as compared to CTRL suggests a training-induced adaptation that led the athletes to a reduced need to activate the anterior leg muscles during the control of single-legged stance (which also seems to be reflected by the lower TA Fmed). From a practical standpoint, the changes in the patterns of EMG activation shown by the acrobatic gymnasts as compared to their counterparts suggest that the design of training strategies might focus on increasing the strength, endurance, and control capacities of specific muscle groups that include the *tibialis anterior*, *soleus*, *biceps femoris,* and *spinal erector* muscles. The rationale relies on the preparation of specific muscle groups differentially used by the gymnasts to keep their postures on single-legged quiet standing before and after the fatiguing protocol. However, the

TABLE 4 | Statistical results of the two-way ANOVAs applied to the measurements of the median frequency of the EMG signal (Fmed) before and after the fatiguing protocol.

Fmed >	Main effects and interactions								
	Group (CTRL *vs* GYN)			Fatigue (NF *vs* PF)			Fatigue *vs* Group		
	$F_{(1,26)}$	P	η^2_p	$F_{(1,26)}$	P	η^2_p	$F_{(1,26)}$	P	η^2_p
SOL	4.681	**0.039**	0.150	0.195	0.661	0.0003	2.641	0.116	0.0051
GM	0.901	0.351	0.032	3.242	0.083	0.0064	0.127	0.724	0.0003
GL	0.132	0.718	0.005	12.869	**0.001**	0.0280	1.985	0.170	0.0044
TA	4.235	**0.049**	0.130	4.096	0.060	0.0053	0.643	0.429	0.0007
VL	0.0535	0.818	0.002	24.675	**<0.001**	0.150	0.0041	0.948	<0.0001
BF	11.103	**0.002**	0.290	8.983	**0.006**	0.015	0.584	0.451	0.001
RA	1.028	0.320	0.034	17.225	**<0.001**	0.066	1.010	0.324	0.0042
ES	2.573	0.120	0.083	46.855	**<0.001**	0.130	4.753	**0.038**	0.015

Significant P values (P < 0.05) are highlighted in bold, indicating significant main effects of fatigue (no fatigue, NF vs post-fatigue, PF) and/or group (control, CTRL vs acrobatic gymnasts, GYN) and/or significant interactions between fatigue and group. TA: tibialis anterior, SOL: soleus, GL: lateral gastrocnemius, GM: medial gastrocnemius, VL: vastus lateralis, BF: and biceps femoris, ES: spinal erector and RA: rectus abdominis.

potential benefit of such strategies in improving the athletes' performance and/or reducing injury risk is made here in a speculative way, so that appropriated controlled trials might be undertaken to test this hypothesis.

The present study included participants aged between 11 and 24 years. It is well known that the performance of the postural control system changes across the lifespan (Barela et al., 2003; Peterson et al., 2006; Kurz et al., 2018; Schedler et al., 2019), so that balance adaptability might be differentially developed in children, adolescents, and adults. Despite the majority of our sample comprised adolescents between 14 and 19 years old (only 1 participant in each group was 11–12 years old and other 23–24 years old), we cannot rule out the possibility that the wide range of participants' age has influenced the results somehow, as some of them might have had their postural performance associated with specific developmental changes in the postural control system.

Altogether, these results indicate that acrobatic gymnastics athletes used different neuromuscular control strategies (as evidenced by the EMG parameters in time and frequency domains), to keep a balance steadiness during the single-legged stance performed before and after the fatigue protocol. Further studies are warranted to investigate whether these mechanisms are accompanied by differences in the angular variation of specific joints (e.g., ankle, knee, hip) that may not have been reflected on the EMG parameters, as well as any differences in the coordination patterns of the associated joint and segments with respect to each other.

CONCLUSION

This study investigated whether expert acrobatic gymnasts respond differentially than their non-trained counterparts in controlling single-legged stance task performed before and after a protocol designed to induce fatigue in the ankle plantarflexors muscles (i.e., fatigue achieved by maintaining the posture "on the tip of the toes"). Fatigue induced a significant increase in postural oscillations in the ML axis, with no significant effects in the AP axis. In terms of postural sway parameters (i.e., COP quantifiers), no superior stability was found for the GYN group as compared to CTRL, irrespective of the fatigue condition. On the other hand, the modulation of EMG parameters (in both time and frequency domains) indicated that expert acrobatic gymnastics athletes (as compared to healthy untrained matched controls) used different neuromuscular control strategies to keep their postures on single-legged quiet standing before and after the fatiguing protocol. Future studies are needed to investigate whether such mechanisms are associated with different movement patterns or, alternatively, are manifested only in the electrical patterns of muscle activation. The efficacy of strategies aimed at improving the performance of specific muscles (i.e., *tibialis anterior*, *soleus*, *biceps femoris* and *spinal erector,* for which particular activation patterns were used by the acrobatic gymnastics to control single-legged quiet standing) on gymnasts' performance must be investigated by appropriate controlled trials.

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 Human Ethics Committee of the School of Arts, Sciences and Humanities of the University of São Paulo (CAAE69034717.6.0000.5390). Written informed consent to participate in this study was provided by the participants' legal guardian/next of kin.

AUTHOR CONTRIBUTIONS

FM and CS were equally involved in the conceptualization and design of the study. MS recruited participants. MS, FL, JG, and JL managed data collection. FM, FL, and CS completed data processing and analysis. FM drafted the first version of the manuscript and supervised data collection. All authors assisted with drafting, provided critical revision of the manuscript, and read and approved the final version of the manuscript.

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REVIEW
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Physiological Mechanisms of Exercise and Its Effects on Postural Sway: Does Sport Make a Difference?

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While the effect of a variety of exercises on postural balance control has been extensively studied, less attention has been paid to those requiring sport-specific skills. Therefore there is a need to analyze the literature and elucidate changes in postural balance control after exercises performed in conditions close to a particular sport. This scoping review aims (i) To map the literature that addresses postural sway aspects of acute responses to general and sport-specific exercises and their underlying physiological mechanisms, and (ii) To identify gaps in the existing literature and propose future research on this topic. The main literature search conducted on MEDLINE, Web of Science, Scopus, PubMed, and Cochrane Library databases was completed by SpringerLink, Elsevier, and Google Scholar. A total of 60 articles met the inclusion criteria. Findings identified that among a variety of studies evaluating the effects of exercise on postural balance control, only few of them were conducted under sport-specific conditions (i.e., while shooting in biathlon or pentathlon, and after simulated or match-induced protocols in combat and team sports). Therefore, more research is still needed to address this gap in the literature and aim research at investigation of postural sway response to sport-specific exercises. Further analysis of the literature showed that the type, intensity and duration of exercise play a key role in increased postural sway. Whole body and localized muscular fatigue of the trunk, neck and lower limbs is considered to be a main factor responsible for the magnitude of balance impairment in an initial phase of recovery and speed of its readjustment to a pre-exercise level. Other likely factors affecting postural stability are hyperventilation and deterioration of sensorimotor functions, though some contribution of muscle damage, dehydration, hyperthermia or dizziness cannot be excluded. A better understanding of the physiological mechanisms of balance impairment after exercises performed under simulated fatigue induced protocol, close to conditions specific to a particular sport, has implications for designing smart exercise programs tailored to individual needs to improve athlete performance with high demands on postural stability and/or decrease their risk of injuries.

Keywords: athletes, general exercise, physiological mechanisms, postural stability, sport-specific exercise

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INTRODUCTION

Balance is one of the limiting factors of performance in many sports. While static balance is important in shooting or archery, in free style sports, snowboarding, skateboarding, windsurfing or cycle acrobacy, dynamic balance plays an essential role in performance. In some sports, like gymnastics, ballet, aerobics, yoga, tai-chi or karate, athletes are required to maintain balance in various sport-specific positions. Others are sports where biomechanical stability for maintenance of balance is limited by a narrow area of support, such as figure-skating, ice-hockey, climbing or mountaineering. Other representative examples are canoeing, rowing, and equestrian sports where balance control in a sitting position is required. Control of body position during and after sport-specific exercises (e.g., body rotations) is necessary in dancing and ballet. Similarly, performance in gymnastics, figure-skating or rock and roll dancing is based on precise regulation of the center of mass movements. This ability also plays some role in weightlifting, powerlifting, golf and throwing events. This also includes most combat sports like boxing, fencing, karate, tae-kwon-do, wrestling and judo. The loss of balance in these injury-prone activities, including martial arts, may not only affect athlete performance but also increase the risk of injuries. Likewise, impairment of postural stability by high vertical forces produced during intensive bouncing exercises in acrobatic sports may cause back pain or lower limb injuries (e.g., ankle sprains). Loosing balance while performing side-to-side movements in individual and team sports, for example badminton, basketball, handball, field-hockey, soccer, softball, squash, table tennis, tennis or volleyball, may also contribute to lower limb injuries (e.g., anterior cruciate ligament sprain or tear). Last but not at least are long-term events such as biathlon, running, cycling, track and field and cross-country skiing, or sports requiring specific technical skills like hurdling and skiing, after which an increased postural sway may be observed when compared to baseline.

Such an exercise induced balance impairment does not only affect the outcome, but may also increase the risk of injuries (Zemková, 2014). Therefore, rapid readjustment of balance after sport-specific exercise to baseline is considered to be an important ability. Acute and chronic adaptations in postural control in response to different forms of exercise have been extensively researched and found to be beneficial for designing sport-specific and rehabilitation programs for balance improvements. Postural sway response to exercise was found to depend on its type, intensity, duration, intensity of proprioceptive stimulation, forms of muscle contraction and activation of muscle fibers (Zemková and Hamar, 2014). The review by Paillard (2012) revealed that short and intensive general exercise (involving the whole body) increases postural sway when the energy expenditure exceeds the lactate accumulation threshold. Hyperventilation rather than fatigue is responsible for increased postural sway after short-term intensive exercises (Zemková and Hamar, 2014). Exhaustive local exercise (involving a particular muscular group) affects postural control when it generates a strength loss at least 25–30% of maximal voluntary contraction (Paillard, 2012). Non-intensive general and local exercises can also disturb postural control when the exercise is prolonged (Paillard, 2012). Fatigue is usually considered as a main factor responsible for balance impairment after prolonged exercise (Zemková and Hamar, 2014). Both general and local exercises contribute to altering the effectiveness of sensory inputs and motor output of postural control (Paillard, 2012). Thus, impairments of sensorimotor functions very likely also play a role in increased post-exercise postural sway (Zemková and Hamar, 2014). This assumption may be corroborated by significant differences in balance impairment after exercises that induced the same ventilation but with a different intensity of muscle contractions eliciting a different level of proprioceptive stimulation, such as calf raises versus jumps and cycling versus running (Zemková and Hamar, 2014). Different compensatory postural strategies are triggered to counteract or limit the disturbance of postural control due to general and local muscle fatigue (Paillard, 2012).

Particularly highly skilled athletes are able to perform successfully in spite of increased postural sway. For instance, gymnasts and ice-hockey players while standing on a narrow area of support, mountaineers during a stance at a height of about 20 m above the ground, skiers and snowboarders on an unstable surface with fixed ankle joints, weightlifters and bodybuilders when performing barbell squats with an additional load, shooters during repeated shots or basketball players during repetitive free throw shots (Zemková, 2014).

Practicing any kind of sport is associated with improved postural stability (Andreeva et al., 2021). Postural adaptations occur in trained subjects because elite athletes exhibit better postural performance and different postural strategy than sub-elite athletes (Paillard, 2017). Adaptations are specific to the context in which the physical activity is practiced (Paillard, 2017). The most successful athletes in terms of sport competition level have the best postural performance both in ecological (specific postural conditions related to the sport practiced) and non-ecological (decontextualized postural conditions in relation to the sport practiced) postural conditions (Paillard, 2019). They also have more elaborate postural strategies compared with athletes at lower competition level (Paillard, 2019).

However, research to date has only marginally addressed the acute effect of sport-specific exercises on postural balance control and their underlying physiological mechanisms. Given the absence of information in a number of such studies, their design and findings, reviews in this field of balance research are warranted. There is a clear need for analysis of the existing literature to elucidate the effects of sport-specific exercises on postural balance control in athletes. A better understanding of physiological changes induced by exercises performed in specific conditions of a particular sport would provide the basis for designing smart balance training programs tailored to individual athletes. This scoping review aims (i) To map the literature that addresses postural sway aspects of acute responses to general and sport-specific exercises and their underlying physiological mechanisms and (ii) To identify gaps in the existing literature and propose future research on this topic.

METHODS

The paper was designed as a scoping review (Armstrong et al., 2011; Sucharew and Macaluso, 2019). Two specific questions were addressed in this review: (1) Does sport specialization play a role in postural sway response to exercise? and (2) What are physiological mechanisms underlying acute effects of exercise on postural sway?

An electronic literature search was provided to analyze existing studies dealing with acute changes in postural balance control induced by various exercises, particularly sport-specific exercises, and their underlying physiological mechanisms. Studies were searched on Web of Science, SCOPUS, PubMed, MEDLINE, and Cochrane Library databases. This search was completed on Google Scholar, SpringerLink, and Elsevier. The articles in peer-reviewed journals were considered for analysis. A manual search for references included in reviews was also conducted to identify further relevant studies. If multiple papers included overlapping data resulting from the same or similar studies, the one with the most recent publication date was analyzed. Articles or abstracts published in conference proceedings, theses, case studies and books were excluded. Articles were also excluded if they did not contain original research or were incomplete. The inclusion criteria involved research articles that specified participants, experimental protocols and measures relevant to this review. The literature search was limited to English language. Articles published after 1990 were preferred, however, earlier relevant studies were also included. Articles that failed to meet the eligibility criteria for this review were excluded.

The initial search was confined to research articles related to the main purpose of this review, i.e., those investigating the effects of sport-specific exercises on postural balance control. Thus, the key inclusion criterion was exercises performed under conditions specific to a particular sport. However, this approach revealed only a limited number of papers that met the eligibility criteria. The search was therefore widened to investigations dealing with acute changes in postural balance control after exercises of different type, intensity, and duration. In particular, physiological mechanisms underlying these changes were studied. These together helped to identify gaps in the existing literature and formulate recommendations for future research on this topic.

The search and appraisal of selected studies on the basis of exclusion and inclusion criteria was performed by the author of this review. Some concerns were related to sample representativeness, missing information about the control group and/or non-controlled compliance of experiments. The target population was athletes of team and individual sports where balance can play a role in their performance, as described in the first paragraph of this review. Proposed sports were combined with the following keywords.

A combination of these terms was included in the search strategy: "acute effect 'AND' post-exercise response 'AND' balance 'AND' postural control 'AND' athletes 'AND' sport-specific exercise 'AND' type of exercise 'AND' exercise intensity 'AND' duration of exercise 'AND' physiological mechanisms." Further searches were conducted using words from subheadings that for example, specified the form of exercise in a particular sport or postural sway variables used. Altogether 133 articles were identified through database searching. Following an initial screening and assessing for eligibility, articles that failed to meet the inclusion criteria were removed. 60 articles that investigated the effect of general and sport-specific exercises on postural balance control were included in this scoping review. Particular phases of the search process are shown in **Figure 1**.

RESULTS AND DISCUSSION

Acute Postural Sway Response to Exercises and Their Underlying Physiological Mechanisms

First of all, studies investigating postural sway response to sports-specific exercises were analyzed. The next step was to analyze studies that involved the effects of a variety of exercises on postural balance control while attention was paid to those discussing potential physiological mechanisms underlying these changes. This search included 60 studies (**Supplementary Table 1**).

Evidence based on the analysis of center of pressure (CoP) sway area and/or velocity velocity in 936 athletes of 13 groups of sports, such as (i) basketball, handball, (ii) biathlon, practical shooting, archery, trapshooting, (iii) boxing, karate, kickboxing, thai boxing, taekwondo, (iv) tennis, table tennis, (v) alpine skiing, snowboarding, (vi) figure skating, freestyle, (vii) football, (viii) rowing, canoeing, canoe slalom, (ix) freestyle wrestling, Greco-Roman wrestling, judo, sumo, (x) speed skating, curling, hockey, short track, (xi) cross-country skiing, (xii) sprint running, stayer running, orienteering, (xiii) artistic gymnastics, rhythmic gymnastics, cheerleading, trampoline tumbling, climbing, high jumping, sailing, skeleton luge, showed that their postural stability is better than in the non-athlete control group (Andreeva et al., 2021).

Good postural stability in shooters results from assiduous training aimed at improving balance (Aalto et al., 1990). They have significantly lower sway velocity compared to untrained subjects both with and without the competition clothing (Aalto et al., 1990). Higher Romberg quotient in shooters than in controls indicates that they use an increased amount of vestibular and proprioceptive cues to stabilize their posture (Aalto et al., 1990). Top-level rifle shooters can stabilize their posture during aiming before the shot better than naive shooters (Era et al., 1996). However, their worse posture stabilization during the last seconds preceding the shot seems not to be a reason for poor shooting performance (Era et al., 1996). Investigating shooting accuracy and precision for prone shooting in biathletes revealed that the exercise intensity has minimal influence on prone shooting performance, but does affect shooting in the standing position by altering the stability of the hold (Hoffman et al., 1992). Maximal physical effort performed on a ski ergometer until exhaustion influences both postural and rifle stability during aiming in biathletes (Sadowska et al., 2019a). Similarly, the running effort within the Laser Run affects the stability of the

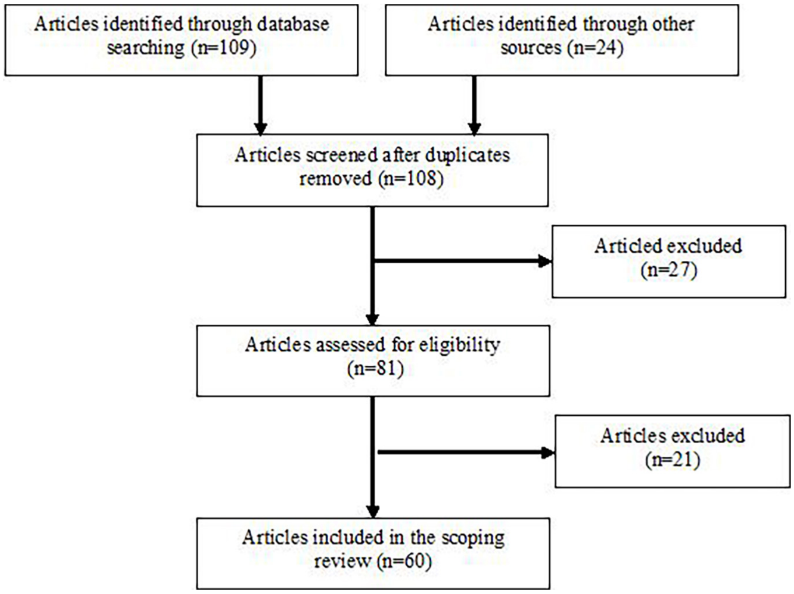

FIGURE 1 | Flow chart illustrating phases of the literature search and study selection.

shooting position in pentathletes (Sadowska et al., 2019b). The main factor hindering accurate and fast shooting is fatigue (Sadowska et al., 2019b). However, the fatigue level does not affect the magnitude of the disturbances of postural balance in the shooting position (Sadowska et al., 2019b).

Postural stability and its readjustment after sport-specific exercises is also important in acrobatic, combat and collision-based team sports. Training consisting of balance exercises usually contributes to better postural stability in highly skilled athletes compared to those at lower level of competition or untrained individuals. For instance, expert gymnasts and experts in other non-gymnastic sports demonstrate larger postural sway in the unipedal tasks when vision is removed; however, this effect is less accentuated for the gymnasts (Vuillerme et al., 2001). Gymnasts are able to use the remaining sensory modalities to compensate for the lack of vision in unstable postures (Vuillerme et al., 2001).

Further, linear and angular sway velocity are lower in a squat position but not in a bipedal stance in wrestlers compared to controls (Mel'nikov et al., 2012). Fatigue induced by PWC (170) test on the cycle ergometer increases postural sway in both bipedal stance and squat position (Mel'nikov et al., 2012). While the linear sway velocity in bipedal stance after PWC (170) increases to an equal extent in both groups, in a squat position it is lower in athletes (Mel'nikov et al., 2012).

In team sports, sensorimotor impairments resulting from previous injuries or muscular fatigue can be a factors contributing to an increased injury risk (Zech et al., 2012). For instance, higher sway area in basketball players than in controls can be related to their history of ankle trauma (Perrin et al., 1991). However, static postural sway measures may be insufficient to allow conclusive statements regarding sensorimotor control in the non-injured athletes (Zech et al., 2012). The CoP sway velocity during a single-leg stance on a force plate increases after both treadmill running and single-leg step-up exercises in handball players, whereas there is no fatigue effects for the star excursion balance test (Zech et al., 2012). Although fatigue affects static postural control, sensorimotor mechanisms responsible for regaining dynamic balance in healthy athletes seem to remain predominantly intact (Zech et al., 2012). On the other hand, soccer match-induced fatigue increases drop jump ground contact time concomitant with the impairment of dynamic balance and agility performance when moving short distances, whereas there are no significant changes in agility performance on longer movement distances, explosive power of lower limbs, static balance, speed of step initiation and the soccer kick (Zemková and Hamar, 2009b). Another study showed that the area of the CoP trajectory during a postural task is larger after a Canadian football G-Sim along with reductions in peak isometric knee extensor torque, peak power and take-off velocity during a countermovement jump (Clarke et al., 2015). These changes may be ascribed to alterations in excitation-contraction coupling due to structural damage and central activation failure (Clarke et al., 2015). The authors suggest that both neuromuscular and somatosensory alterations are induced by acute game-induced fatigue in collision-based team sports players.

As a consequence of match-induced fatigue may be an increased risk of injuries. There is a high correlation (0.92) between delays in peroneal muscle reaction time (onset of EMG following sudden ankle inversion) and increases in postural sway amplitude (Konradsen and Ravn, 1991). This delay in peroneal reaction time results in a delay of muscle force generation, which is similar to the electromechanical delay of muscle force generation seen with muscular fatigue (Häkkinen and Komi, 1983; Hortobágyi et al., 1991). Delay in muscle force generation leads to an increase in unilateral postural sway amplitude and may result in lower limb injury during long-term athletic activity (Tropp et al., 1984). Thus, if the forces required for the correction of an unstable placement of the foot are delayed due to fatigue, then mainly ankle joints would be at risk of injury. Ankle sprains and rupture of the anterior cruciate ligament are among the most serious injuries in players. These injuries occur mostly in the end of the match, which raises the possibility that muscular fatigue at the ankle and knee joints would place players at greater risk of injury.

Most of the studies considered fatigue as a main factor responsible for post-exercise balance impairment. The increase in postural sway was revealed after fatigue of ankle plantar-flexor muscles (Gimmon et al., 2011) induced by isokinetic contractions (Lundin et al., 1993; Yaggie and McGregor, 2002), isometric and isokinetic exercises (Bisson et al., 2012), repeated plantar-flexion of both legs (Corbeil et al., 2003), toe-lift until exhaustion (Pinsault and Vuillerme, 2008), single-leg, weight-bearing heel raises on an inclined platform (Springer and Pincivero, 2009), repeated standing heel raise exercise (Hlavackova and Vuillerme, 2012), and a calf raise exercise on top of a step (Barbieri et al., 2019). Also fatigue induced by voluntary muscular contraction or electrical stimulation applied to the dorsi-plantar flexor (Yu et al., 2014), the triceps surae (Magalhães and Kohn, 2012), and the quadriceps femoris (Paillard et al., 2010; Chaubet et al., 2012; Chaubet and Paillard, 2012) can impair postural control. Furthermore, it is lower extremity joint (knee, or ankle) and overall fatigue of the dominant leg (Dickin and Doan, 2008), fatigue induced by a high-intensity free-weight back-squat exercise (Thiele et al., 2015), lumbar extensor fatigue (Davidson et al., 2004; Pline et al., 2005) induced by a trunk extension-flexion exercise on a roman chair (Parreira et al., 2013), cervical muscular fatigue (Vuillerme et al., 2005), and neck musculature fatigue (Liang et al., 2014). Moreover, whole-body fatigue in a form of exercise on a rowing ergometer (Springer and Pincivero, 2009), maximal and submaximal cycle ergometry (Hill et al., 2014), cycle ergometry and treadmill walking (Hill et al., 2015), 25 min of moderate running on a treadmill (Marcolin et al., 2019), Wingate test on a cycle ergometer (Wojciechowska-Maszkowska et al., 2012), two times of a maximal voluntary pedaling for 10 s and at 50% of maximal aerobic power for 60 min at 60 rpm (Demura and Uchiyama, 2009), and maximum-effort sprints and yo-yo intermittent recovery test, level 1 exercise (Fox et al., 2008) plays a role in increased values of postural sway. The impairment of a single-leg dynamic balance is higher after aerobic (the Bruce protocol on a motorized treadmill) than anaerobic exercise (four maximal cycling efforts against a resistance equivalent to 0.075 kg/body mass for 30 s with 3-min rest intervals) (Güler et al., 2020). While the overall stability index and the anterior/posterior index increases significantly immediately following the fatiguing treadmill test, their values are not altered significantly after an incremental test on a cycle ergometer (Wright et al., 2013). There are no significant differences in the equilibrium and strategy scores after maximal exercise on the cycle ergometer as compared to baseline, neither with eyes open nor with eyes closed while standing on a stable platform, however, these values measured under dynamic conditions are significantly lower than prior to exercise with eyes closed as well as with sway-referenced vision (Zemková et al., 2007). Sensory analysis reveals that the vestibular system is more affected by exercise than the somatosensory system.

Findings indicate that the effect of exercise on balance depends mainly on its type, intensity and duration. While strenuous physical exercise (treadmill walking and cycle ergometer pedaling) increases body sway, it is little affected by exercise performed below the estimated anaerobic threshold (Nardone et al., 1997). Postural sway is also affected by prolonged fatiguing exercise in a form of treadmill walking for 25 min (Nardone et al., 1998). In both cases this effect is of moderate extent and short-lasting (Nardone et al., 1997, 1998). It seems that abrupt intensive exercise has a more profound but shorter detrimental effect on balance than prolonged exercise of moderate intensity. While the impairment of balance in the first case is mediated mainly by hyperventilation, fatigue is responsible for longer balance disorders in the second case.

Therefore both the magnitude of balance impairment in an initial phase of recovery as well as the speed of its readjustment should be taken into account when evaluating postural sway response to exercise (Zemková and Hamar, 2014). Since the adjustment of sway velocity at the onset of post-exercise recovery phase seems to be exponential, it may be characterized by the time required to decline to 50% of the difference between maximal sway velocity (achieved in an initial 5-s post-exercise phase) and steady state balance ($t_{1/2}$). This time is longer after maximal stepwise running and cycling than after abruptly instituted exercise. However, it may take few minutes to return postural sway to the pre-exercise level after prolonged or intensive exercises. One may identify three phases in the recovery period. In an initial phase of recovery there is a plateau or slight increase in the CoP velocity, which is most probably associated with a re-payment of ventilation. Second is a fast exponentional component in the decline of CoP velocity, with a half-time of about 1 min. This phase is most likely associated with a decline in ventilation, as it has been demonstrated by close correlation between the reduction of ventilation and CoP velocity in the recovery period (Zemková and Hamar, 2003). Then comes a more complex, slow component. A sustained increase in the sway velocity during the last phase may be attributed to the impairment of proprioceptive feedback mechanisms involved in balance control. However, the contribution of fatigue or the deterioration of other components of sensorimotor system cannot be excluded.

Overall, our previous research identified higher increase in postural sway, and in some cases also its slower return to a baseline, after short-term abruptly instituted intensive than longer stepwise exercise on the cycle ergometer, prolonged (45 min) than shorter (15 min) cycling at moderate intensity, rebound jumps than calf raises, treadmill running than cycling, upslope than level running on the treadmill, and cycling at higher (130/min) than lower (70/min) revolution rates (Zemková and Hamar, 2014).

On the other hand, some authors have reported no significant changes in postural sway variables after exercise. A recent study by Lyu et al. (2021) evidenced a general compensation in the central nervous system in response to the neuromuscular deficiencies induced by local fatiguing exercise and put forward the function of sensory recalibration in maintaining postural stability under fatigue conditions. For instance, repeated measurements for up to 10 min after a fatiguing calf-muscle exercise showed no increase of body sway, which indicates that postural control in quiet standing can be maintained by compensatory mechanisms activated during muscle fatigue (Adlerton and Moritz, 1996). The next study by Adlerton and Moritz (2001) also revealed that fatiguing exercise does not influence the CoP shift caused by vibration, thus indicating unchanged excitability of muscle spindles in fatigued muscles. Though calf-muscles fatigue does not impair postural control, it generates a change of the contribution of the proprioceptive information (myotatic loops), which is greater after voluntary muscular contractions than after electrical stimulation superimposed onto voluntary muscular contractions (Bizid et al., 2009). Similarly, ankle muscle fatigue does not affect postural variables under eyes open conditions, but with eyes closed sway area and antero-posterior velocity increases when both plantarflexors and dorsiflexors are fatigued simultaneously (Boyas et al., 2011). This may be ascribed to the impairment in the compensatory activity between agonist and antagonist muscles and/or a greater decrease in proprioception due to a greater number of fatigued muscles (Boyas et al., 2011). Fatiguing exercises consisting of sustaining plantarflexor isometric contractions at the intensity of 25, 50, and 75% of maximal isometric plantarflexor torque until task failure does not influence the extent of postural stability impairment, but does influence the type of fatigue induced and the neuromuscular function predictors explaining changes in postural variables (Boyas et al., 2013). Further, unilateral muscle fatigue induced on the hip's abductors of the dominant leg yields to larger CoP displacements under the non-fatigued leg only (Vuillerme et al., 2009). This indicates that supplementary somatosensory inputs to the central nervous system preserves/facilitates postural control in the condition of altered neuromuscular function of the dominant leg's hip abductors induced by the fatiguing exercise (Vuillerme et al., 2009). The other study by Strang and Berg (2007) showed that muscle fatigue generated by a dead-lift exercise performed to exhaustion has no effect on postural stability, and yet caused earlier anticipatory postural adjustment onsets in the contralateral paraspinals, ipsilateral paraspinals, and contralateral paraspinals.

These findings indicate that fatigue induced changes in the somatosensory system have to be taken into account when interpreting post-exercise changes in postural balance control. Prolonged exercise in a form of a 25-km run and ergocycle exercise of identical duration (on average 1 h 44 min) impairs postural stability during conflicting sensory conditions in well-trained triathletes, with some differences depending on the kind of exercise (Lepers et al., 1997). These athletes used vestibular inputs less effective after running than cycling while maintaining balance (Lepers et al., 1997). The authors suggest that adaptation to prolonged stimulation of proprioceptive, visual and vestibular inputs during exercise most likely occurred in the integrating centers. However, an impairment of motor efferents or hemodynamic changes can not be excluded. Also treadmill running at speed of 2.2 m/s tends to disturb postural stability more than walking at speed of 1.9 m/s, possibly due to more excessive head movement observed by larger vertical displacement and acceleration pattern, and disturbance of visual and vestibular information centers (Derave et al., 2002). As the study shown, exercise of moderate intensity increased two-dimensional postural sway in eyes open only. This deteriorated visual contribution to postural stability was evident as an initial destabilization in the sagittal direction and a less transient loss of latero-lateral stability (Derave et al., 2002). Likewise, Hashiba (1998) identified greater mean fore-back postural sway after treadmill running at a speed of 10 km/h than walking at a speed of 7 km/h. The author demonstrated that vision during treadmill locomotion plays an important role in evoking postural sway after such an exercise. Somatosensory/motor signals may be stored during visual-somatosensory/motor conflict and this stored information may evoke postural change and self-motion perception (Hashiba, 1998). Furthermore, prolonged (60 min) cycle ergometer exercise was found to increase the mean time interval between two consecutive peaks in sway density plot, thus decreasing the control rate but not changing the stability level (Mello et al., 2009). Conversely, the maximal oxygen uptake test caused a decrease of the mean duration of peaks in the sway density plot, decreasing the stability level, without modifying the rates of central and muscular torque controls (Mello et al., 2009). This means that visual privation has a more detrimental effect on body sway than these exercises, though it also depends on their intensity and duration (Mello et al., 2009). Fatiguing running also affects static and dynamic postural control in active athletes with previous ankle sprain (Steib et al., 2013). Their fatigue-induced alterations of dynamic postural control (Star Excursion Balance Test) were greater as compared to uninjured controls (Steib et al., 2013). This indicates that ongoing deficits in sensorimotor control may contribute to the enhanced ankle reinjury risk (Steib et al., 2013). Similarly, a latent impairment of balance performance was found following a bout of plyometric exercise consisting of 200 countermovement jumps designed to elicit symptoms of muscle damage (Twist et al., 2008). This has implications for both the use of skill-based activities and for increased injury risk following high-intensity plyometric training (Twist et al., 2008).

Another factor is dehydration. Prolonged exercise (cycled for 2 h at a power output equal to 57–63% VO_2max) without fluid ingestion negatively affects postural stability, whereas there is no effect after exercise with fluid replacement (intake of 1.9l of a carbohydrate-electrolyte solution) or after thermal dehydration induced by seven 15 min consecutive sauna sessions (85°C, 50% rh) with no fluid replacement (Derave et al., 1998). However, more recent study by Mtibaa et al. (2018) demonstrated that hyperthermia impairs the proprioception and balance parameters due to heat-induced alterations in efferent and afferent signals to and from the muscle. The exercise that induced a mild dehydration, which increases proteinemia and leads to body mass loss, was found to impair balance in the standard situation and when the vestibular cue is reliable (Lion et al., 2010). There was a correlation between the decreased use of vestibular input and the dehydration level (Lion et al., 2010). Even though muscular fatigue could explain the decrease in postural performances, vestibular fluid modifications may also be involved by its influence on the intralabyrinthine homeostasis, thus lowering the contribution of vestibular information on balance control (Lion et al., 2010). Fatigue mainly alters muscular effectors and sensory inputs, such as proprioception, resulting in poor postural regulation (Gauchard et al., 2002). Fluid ingestion could be responsible for the preservation of muscular functions and of sensory afferences accurately regulating postural control (Gauchard et al., 2002).

Last but not at least, increased ventilation plays an important role in postural sway response to exercise (Zemková and Hamar, 2014). In general, body movements associated with paced respiration disturb the postural control system (Hunter and Kearney, 1981). The magnitude of the respiratory contribution to sway is constant over the normal range of respiratory rates and linearly relates to respiration amplitude (Hunter and Kearney, 1981). Interestingly, the respiratory component of the sway path is larger in seated than in standing subjects, indicating that sitting entails less instantaneous steadiness (Bouisset and Duchene, 1994). Moreover, the sway distance is greater when holding breath after inspiration than expiration, and increasing the respiration rate produces a greater postural sway (Jeong, 1991). Specifically, breath holding that leads to activation of postural control is more pronounced in athletes (Malakhov et al., 2014). The athletes' postural system also compensates for hyperventilation more efficiently when compared to controls but with greater effort (Malakhov et al., 2014).

Investigation of postural sway after light (40 W), moderate (85 W), and heavy (125 W) work loads under conditions of wearing a full facepiece respirator but no respiratory protection device revealed that its values increase more quickly and in a more consistently linear fashion with increasing work load under the respirator than the non-respirator condition (Seliga et al., 1991). This may be attributed to the increasing work load-induced proprioceptive fatigue effect on the nervous system's ability to process signals from proprioception systems incongruent with body sway (Seliga et al., 1991). Another study evaluated postural response to a strenuous treadmill exercise and a 3 s bilateral soleus muscle vibration after the strenuous exercise (Bove et al., 2007). There was a linear relationship between sway path and oxygen uptake, which indicates that body instability may be due to rapid recovery of oxygen uptake (Bove et al., 2007). However, the fatigue-induced body instability was not associated with postural sway response to soleus muscle vibration (Bove et al., 2007). Ventilatory demands regulate diaphragmatic force-generation during exercise, whereas diaphragmatic fatigue must be attributed to non-ventilatory controlled feedback mechanisms (Kabitz et al., 2008). Evidence clearly indicates that hyperventilation impairs postural stability (Malakhov et al., 2014). For instance, hyperventilation induced by a maximum-intensity, incremental cycling exercise is accompanied by an increase in postural sway, indicating a reduction in postural stability following a change in ventilatory drive (David et al., 2010, 2015). It appears that the compensation of respiratory disturbances for erect posture is less effective when minute ventilation increases (David et al., 2015). A close link between sway velocity and ventilation in an initial phase of recovery was also found after resistance exercises (Zemková and Hamar, 2009a). The highest increase in the CoP velocity as well as ventilation was found after squats, followed by calf raises, voluntary hyperventilation, biceps curls, and presses behind the neck (Zemková and Hamar, 2009a). Thus, voluntary hyperventilation also increases postural sway (Sakellari and Bronstein, 1997). This increase may be mediated by derangement of both peripheral and central somatosensory signals from the lower limbs (Sakellari et al., 1997). Hyperventilation disrupts mechanisms mediating vestibular compensation. It seems to spare vestibular reflex activity and cerebellar-mediated eye movements (Sakellari et al., 1997). All these factors have to be taken into consideration when interpreting physiological mechanisms underlying postural sway response to exercise.

Gaps in Current Studies Investigating Acute Effects of Exercise on Postural Sway and Proposals for Future Research

Although there is a wide variety of studies investigating the effects of exercise on postural balance control, less attention has been paid to those performed under sport-specific conditions. Previous reviews revealed that an increase in post-exercise CoP velocity is lower and its readjustment to pre-exercise level is faster in experienced athletes, for instance after body rotations in dancers and synchronized swimmers or judo falls in judo competitors (Zemková, 2014). The magnitude of post-exercise balance impairment depends on the form of exercise simulating conditions of a particular sport (e.g., Latino American vs. rock and roll dancing), intensity (e.g., maximal vs. bouncing aerobic jumps), and their duration reflecting the duration of performance (Zemková et al., 2010). It seems that sensory functions are more profoundly affected by intensive jumps than motor functions of the task-oriented balance exercise based on visual feedback control of the center of mass (CoM) position (Zemková and Hamar, 2011). However, after reaching some level of deterioration of proprioceptive function, there is no further impairment of balance parameters.

In most studies, functional testing protocols have been usually used to evaluate the effect of exercise-induced fatigue on the postural control system. However, these laboratory experiments, including stepwise increasing exercise loads on a treadmill or cycle ergometer, in many ways represent artificial conditions. Nevertheless, these procedures provide standardized protocols and can also simulate the physiological demands of many sports. Despite the many advantages of laboratory diagnostics, such exercises do not reflect specific changes in the neuromuscular system induced by a particular sport. From both a practical and a theoretical point of view it is therefore equally important to investigate the effect of intermittent exercise on postural balance control, which better reflects the type of muscular activities encountered in most sports. The intermittent exercise at a high intensity level is an activity pattern, where periods of intense exertion are interspersed with periods of active or passive recovery. Therefore, simulated fatigue induced protocols should be used in order to be closer to specific conditions in a particular sport. Nonetheless, it seems that the intensity of exercise rather than its mode (e.g., continual vs. intermittent exercise) plays a role in an increased postural sway (Zemková and Dzurenková, 2009).

Based on the analysis of the literature, there are only few studies that evaluated the effects of sport-specific exercise on balance and how it may affect sport performance, for instance in biathletes (Hoffman et al., 1992; Sadowska et al., 2019a), pentathletes (Sadowska et al., 2019b), wrestlers (Mel'nikov et al., 2012), handball (Zech et al., 2012), soccer (Zemková and Hamar, 2009b), or football players (Clarke et al., 2015). Therefore, further studies should be focused on the investigation of changes in postural sway following exercises performed under sport-specific conditions.

Post-exercise balance impairment is often associated with fatigue induced by prolonged exercises. On the other hand, more marked ventilation is responsible for increased postural sway after short-term intensive exercises, though some contribution of fatigue cannot be excluded. This was demonstrated by the close correlation between sway velocity and ventilation in an initial phase of recovery after intensive cycling bouts (Zemková and Hamar, 2003) as well as after resistance exercises (Zemková and Hamar, 2009a). The contribution of respiration to postural sway is low during quiet breathing but linearly increases with the respiration amplitude (Hunter and Kearney, 1981; Hodges et al., 2002). Although a threshold for posture-destabilizing fatigue-effects does not appear to exist, sizeable and potentially dangerous destabilization does occur when exercise intensity exceeds 50% of VO_2max (Nardone et al., 1997). Besides fatigue of lower limbs (Lundin et al., 1993; Yaggie and McGregor, 2002; Corbeil et al., 2003; Dickin and Doan, 2008; Pinsault and Vuillerme, 2008; Springer and Pincivero, 2009; Paillard et al., 2010; Gimmon et al., 2011; Bisson et al., 2012; Chaubet et al., 2012; Chaubet and Paillard, 2012; Hlavackova and Vuillerme, 2012; Magalhães and Kohn, 2012; Yu et al., 2014; Thiele et al., 2015; Barbieri et al., 2019), trunk (Davidson et al., 2004; Pline et al., 2005; Parreira et al., 2013), neck (Schieppati et al., 2003; Vuillerme et al., 2005; Liang et al., 2014) and whole body (Nardone et al., 1997, 1998; Zemková et al., 2007; Fox et al., 2008; Demura and Uchiyama, 2009; Springer and Pincivero, 2009; Wojciechowska-Maszkowska et al., 2012; Wright et al., 2013; Hill et al., 2014, 2015; Marcolin et al., 2019; Güler et al., 2020), also hyperventilation (Zemková and Hamar, 2003; David et al., 2010, 2015; Malakhov et al., 2014) and/or increased oxygen uptake (Bove et al., 2007) play a main role in post-exercise postural stability. Other possible physiological mechanisms of post-exercise increase in postural sway include impairments of visual cues, vestibular system and proprioceptive functions (Lepers et al., 1997; Hashiba, 1998; Derave et al., 2002; Zemková et al., 2007; Mello et al., 2009), muscle damage (Twist et al., 2008), dehydration (Derave et al., 1998; Lion et al., 2010), hyperthermia (Mtibaa et al., 2018), and dizziness resulting from hyperventilation. Nevertheless, more research is needed to investigate the association of these factors with sport-specific exercises and their effect on athlete performance with high demands on postural stability.

Balance impairment after resistance exercises is also a consequence of more pronounced ventilation rather than fatigue (Zemková and Hamar, 2009a). This effect is more evident after exercises performed with lower (squats and calf rises) than upper extremities (biceps curls and presses behind the neck). The sway velocity after voluntary hyperventilation reaches a maximum at the end of exercise, and starts to decline immediately in the recovery phase. However, its values after resistance exercises, especially after those performed with lower limbs, remain temporarily elevated and a gradual decrease back to the resting level set in only after about 10 and 25 s. This effect is mainly a consequence of the delayed activation of ventilation in an early phase of recovery after such exercises. This assumption may be corroborated by the close correlation between the level of ventilation and sway velocity in the recovery phase (Zemková and Hamar, 2009a). Besides the type of exercise and muscle mass activated, postural sway response to resistance exercises depends on their contraction intensity (additional load used), rate of movement, number of repetitions and sets, and the intensity of proprioceptive stimulation. However, physiological mechanisms underlying changes following resistance exercises performed under sport-specific conditions have yet to be investigated.

These findings have to be taken into account in sports dependent upon post-exercise postural stability, such as biathlon, figure skating, rock and roll dancing, and so forth. Such investigations are essential for designing sport-specific training programs tailored to individual needs. For instance, a recent study by Hill et al. (2016) investigated whether high-intensity cycling training leads to adapted responses of balance performance in response to exercise-induced muscle fatigue. The authors found that 3 weeks of high-intensity training (HIT) on a cycle ergometer resulted in longer recovery times following fatigue compared to pre-training assessments. After 6 weeks of HIT, postural sway following fatigue was attenuated. A better understanding of acute and adaptive changes in postural sway after various exercises may help athletes to improve their performance and decrease the risk of injuries under fatigue conditions.

CONCLUSION

This scoping review revealed that among a variety of studies evaluating the effects of exercise on postural balance control, only few of them were tested under sport-specific conditions (i.e., while shooting in biathlon or pentathlon, and after simulated or match-induced protocols in combat and team sports). Therefore, more research is still needed to address this gap in the literature and aim research at the investigation of postural sway response to sport-specific exercises. Further analysis of the existing literature showed that the type, intensity and duration of exercise plays a main role in increased postural sway. Whole body and localized muscular fatigue of the trunk, neck and lower limbs is considered as a main factor responsible for the magnitude of balance impairment in an initial phase of recovery and speed of its readjustment to a pre-exercise level. Hyperventilation and deterioration of sensorimotor functions can also be included, though some contribution of muscle damage, dehydration, hyperthermia or dizziness cannot be excluded. A better understanding of the physiological mechanisms of balance impairment after exercises performed under simulated fatigue induced protocols close to conditions specific to a particular sport, has implications for designing smart exercise programs tailored to individual needs and to improve athlete performance with high demands on postural stability and/or decrease their risk of injuries.

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The author confirms being the sole contributor of this work and has approved it for publication.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2022.792875/full#supplementary-material

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The Role of Neuromuscular Control of Postural and Core Stability in Functional Movement and Athlete Performance

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Balance and core stabilization exercises have often been associated with improved athlete performance and/or decreased incidence of injuries. While these exercises seem to be efficient in the prevention of injuries, there is insufficient evidence regarding their role in sport-specific performance and related functional movements. The aim of this scoping review is (1) to map the literature that investigates whether currently available variables of postural and core stability are functionally related to athlete performance in sports with high demands on body balance and spinal posture and (2) to identify gaps in the literature and suggest further research on this topic. The literature search conducted on MEDLINE, Scopus, Web of Science, PubMed, and Cochrane Library databases was completed by Google Scholar, SpringerLink, and Elsevier. Altogether 21 articles met the inclusion criteria. Findings revealed that postural stability plays an important role in performance in archery, biathlon, gymnastics, shooting, and team sports (e.g., basketball, hockey, soccer, tennis). Also core stability and strength represent an integral part of athlete performance in sports based on lifting tasks and trunk rotations. Variables of these abilities are associated with performance-related skills in cricket, cycling, running, and team sports (e.g., baseball, football, hockey, netball, soccer, tennis). Better neuromuscular control of postural and core stability contribute to more efficient functional movements specific to particular sports. Training programs incorporating general and sport-specific exercises that involve the use of postural and core muscles showed an improvement of body balance, back muscle strength, and endurance. However, there is controversy about whether the improvement in these abilities is translated into athletic performance. There is still a lack of research investigating the relationship of body balance and stability of the core with sport-specific performance. In particular, corresponding variables should be better specified in relation to functional movements in sports with high demands on postural and core stability. Identifying the relationship of passive, active, and neural mechanisms underlying balance control and spinal posture with athlete performance would provide a basis for a multifaced approach in designing training and testing tools addressing postural and core stability in athletes under sport-specific conditions.

Keywords: body balance, core stabilization, neuromuscular functions, spinal stability, sport-specific performance

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INTRODUCTION

Postural and core stability is critical to almost all movements in sport (Sharrock et al., 2011), particularly when maintaining balance on an uneven surface or while responding to sudden perturbations (Zazulak et al., 2007). While most research has been devoted to the role of postural stability in athletic performance, far fewer studies have investigated the relationship between core stability and sport-specific skills.

The core that involves lumbopelvic–hip region maintains the vertebral column equilibrium within its physiological limit by reducing postural displacement after unexpected perturbations (Reeves et al., 2007). This requires instantaneous activation of the central nervous system to evoke optimal muscle recruitment for both stability and mobility. Core muscles provide the necessary stability for the production of force in the lower limbs and efficient control of body movements (Rivera, 2016). Deficiencies or imbalances in the core muscles can increase fatigue, decrease endurance, and increase the risk of injuries in athletes (Rivera, 2016).

Recently, widely promoted spine stabilization and core strengthening exercises have been seen to improve postural and core stability and/or reduce back problems in athletes (Akuthota et al., 2008). These exercises seem to be efficient in the prevention and rehabilitation of back pain, lumbar spine injuries, or other musculoskeletal disorders. However, there is a lack of evidence regarding their effectiveness for improvements of functional movements and consequently also athletic performance. This is mainly due to a limited number of appropriate tests evaluating postural and core stability that would be able to provide deeper insights into the understanding of exercise-induced changes in neuromuscular functions under sport-specific conditions.

Currently, motion analysis and accelerometry recordings allow monitoring of head, trunk, and limb movements and provide useful data for a complete assessment of postural and core stability during a variety of functional movements. Measurement of postural sway using accelerometry is strongly related to task-based variables (Whitney et al., 2011). The accelerometry combined with stochastic dynamics quantifies the time-varying structure of postural sway pattern (Lamoth et al., 2009). For instance, acceleration time-series are more stable, less variable, and less regular with greater gymnastic skills (Lamoth et al., 2009).

In the study by Glofcheskie and Brown (2017), a seated balance task was used to assess trunk postural control, electromyography, and kinematics to measure neuromuscular control in response to unexpected trunk perturbations, and active trunk repositioning tasks to examine proprioceptive ability. There was an interactive relationship between postural control, trunk neuromuscular control, and trunk proprioception in athletes of different training backgrounds (Glofcheskie and Brown, 2017). More specifically, greater trunk postural control (less CoP movement), less lumbar spine angular displacement, higher muscle activation amplitudes, and faster trunk muscle activation onsets in response to unexpected trunk perturbations were found in athletes (collegiate level long-distance runners and golfers) than non-athletes (Glofcheskie and Brown, 2017). Absolute and variable errors in trunk repositioning tasks were lower in golfers than runners and controls, which indicates their greater proprioceptive ability (Glofcheskie and Brown, 2017).

Usually, postural and core stability have been compared among athletes of different sports, their age, and/or performance level. For instance, the best body balance is found in gymnasts, then in soccer players, swimmers, physically active controls, and basketball players (Hrysomallis, 2011). Balance is related to the competition level of athletes, and the more proficient ones display better postural stability (Hrysomallis, 2011). Athletes of rifle shooting, soccer, and golf have better postural stability than their less-proficient counterparts (Hrysomallis, 2011). Paillard (2019) reported that the most successful athletes have the best postural performance, both in ecological and non-ecological postural conditions, that is, specific vs. decontextualized in relation to the sport practiced. They also have more elaborate postural strategies than those at lower competition levels (Paillard, 2019). Specific muscle synergies are of considerable value as a training strategy for hockey players who need to improve their postural stability and reduce their potential risk of injuries (Kim et al., 2018).

Balance is also associated with performance measures (Hrysomallis, 2011). Body sway measured during stance on a force plate is related to aim point fluctuation and shooting performance (Ball et al., 2003). As body sway increases, performance decreases and aim point fluctuation increases for most relationships in elite rifle shooters (Ball et al., 2003). Postural balance in the standing position is also related to the shooting accuracy, both directly and indirectly, through rifle stability (Mononen et al., 2007). Furthermore, a balance ratio (contact with floor to no contact time) during a 30-s wobble board test correlates with maximum skating speed in hockey players (Behm et al., 2005). Unipedal static balance, core strength, and stability correlate with golf performance in elite players (Wells et al., 2009). There is also a relationship between unipedal dynamic balance and the luge starting speed (Platzer et al., 2009a).

In general, practicing any kind of sport is associated with better postural stability (Andreeva et al., 2021). The center of pressure (CoP) velocity during a bipedal stance on a force platform with eyes open is lower in shooters, football players, boxers, cross-country skiers, gymnasts, runners, team sport players, wrestlers, tennis players, alpine skiers, rowers, speed skaters, and figure skaters when compared to the general population (Andreeva et al., 2021). Athletes usually display better postural stability in sport-specific conditions and sway measures may not reveal between and within-group differences when testing in a standard upright position (Zemková, 2014). There are also differences in the magnitude of postural sway increase after sport-specific exercises and the speed of its readjustment to pre-exercise level (Zemková, 2014).

Investigating the relationship of passive, active, and neural mechanisms underlying balance control and spine stabilization with sport-specific performance would provide a basis for a multifaced approach in designing training and testing tools addressing postural and core stability in athletes. The aim of this scoping review was (1) to review the existing literature that deals with sports with high demands on body balance and spinal posture and to investigate whether currently available variables of postural and core stability are functionally related to athletic performance and (2) to identify gaps in the literature and suggest further research on this topic.

METHODS

This article was proposed as a scoping review (Armstrong et al., 2011). The purpose was to provide an overview of the available research evidence and answer the following question: (1) Is there a relationship between postural and core stability and functional movement and/or athletic performance?

An electronic literature search was provided to analyze existing studies dealing with the role of neuromuscular control of both postural and core stability in functional movement and/or athlete performance. Studies were searched on Scopus, Web of Science, PubMed, MEDLINE, and Cochrane Library databases. This search was completed on Google Scholar, Elsevier, and SpringerLink. The articles in peer-reviewed journals were considered for analysis. References included in review articles were also searched to identify further relevant studies. If articles included overlapping data from the same or similar study, the one with the most recent publication date was analyzed. Articles or abstracts published in conference proceedings, theses, case studies, and books were excluded. Articles were also excluded if they did not contain original research or were incomplete. The inclusion criteria involved research articles that specified participants, experimental protocols, and measures relevant to this review. The literature search was limited to the English language. Articles published after 1990 were preferred. Articles were excluded if they failed to meet the eligibility criteria.

The initial search was confined to research articles closely related to the main purpose of this scoping review, that is, those dealing with the relationship between neuromuscular control of either postural or core stability and functional movement and/or athlete performance. However, this approach revealed only a limited number of articles that met the eligibility criteria. The search was, therefore, widened to investigations dealing with the effects of sport-specific and balance or the core-related exercises on functional movements and skills within a particular sport. In particular, neuromuscular mechanisms underlying these relationships were studied. This together helped us to identify gaps in the literature and formulate recommendations for further studies in this field of research.

The search and appraisal of selected studies on the basis of exclusion and inclusion criteria were performed by both authors of this review. Some concerns were related to sample size and its representativeness, incomplete information about the methods used, variables analyzed, and/or non-controlled compliance of experiments. The target population was athletes of a team and individual sports where balance and core stability play an essential role in their performance. Proposed sports were combined with the following keywords.

A combination of these terms was included in the search strategy: "postural stability" AND "core stability" AND "core endurance" AND "core strength" AND "core training" AND "body balance" AND "postural control" AND "spinal posture" AND "lumbopelvical stability" AND "athletes" AND "sport-specific exercise" AND "athletic performance" AND "functional movement," AND "neuromuscular control."

Further searches were conducted by using words from subheadings that specified the contribution of postural and core stability on performance in highly skilled athletes in comparison with those at a lower level of sport-specific skills. Following an initial screening of articles identified through database searching and assessing for their eligibility, those that failed to meet inclusion criteria were removed. Articles that investigated neuromuscular control of postural (14 out of 29) and core (7 out of 13) stability in association with functional movements and/or athlete performance were included in this scoping review. The search process phases are displayed in **Figure 1**.

RESULTS AND DISCUSSION

The Role of Neuromuscular Control of Postural Stability in Functional Movement and/or Athlete Performance

Analysis of the literature revealed (**Table 1**) that postural stability plays an important role in functional movements and/or athlete performance in shooting (Era et al., 1996; Ball et al., 2003; Mononen et al., 2007; Ihalainen et al., 2016a,b, 2018; Lang and Zhou, 2021), gymnastics (Opala-Berdzik et al., 2021), dancing (Munzert et al., 2019), and team sports, such as soccer (Jadczak et al., 2019a,b).

Shooting, Biathlon, and Archery

The majority of studies investigated postural stability in association with shooter performance. Postural stability and stability of hold were identified as the main factors influencing air rifle shooting performance (Era et al., 1996; Konttinen et al., 1998; Ball et al., 2003). High postural stability and small gun barrel movements determine shooting performance in novice shooters (Mononen et al., 2007). High postural stability is also important in elite rifle shooters (Lang and Zhou, 2021). Specifically, CoP variables measured during stance on a force plate negatively correlate with shooting score and aiming accuracy, whereas there is a positive correlation with the stability of hold and stability of triggering (Lang and Zhou, 2021). The timing of triggering, cleanness of triggering, and aiming accuracy then influence shooting score in elite-level air rifle shooters (Ihalainen et al., 2016a). Taking together, postural stability, cleanness of triggering, aiming accuracy, and stability of hold affect performance in both training and competition situations even in athletes at high–shooting skill levels (Ihalainen et al., 2016b). Particularly, body sway is related to aim point fluctuation in shooters (Ball et al., 2003). Aim point fluctuation increases and performance decreases as body sway increases for most relationships (Ball et al., 2003). However, Spancken et al. (2021) identified that body sway does affect shot score in national- and elite-level athletes in both small-bore and air-rifle shooting, whereas aiming time, aiming accuracy, and horizontal rifle stability influence shot score in national-level air-rifle athletes. A higher Romberg quotient in

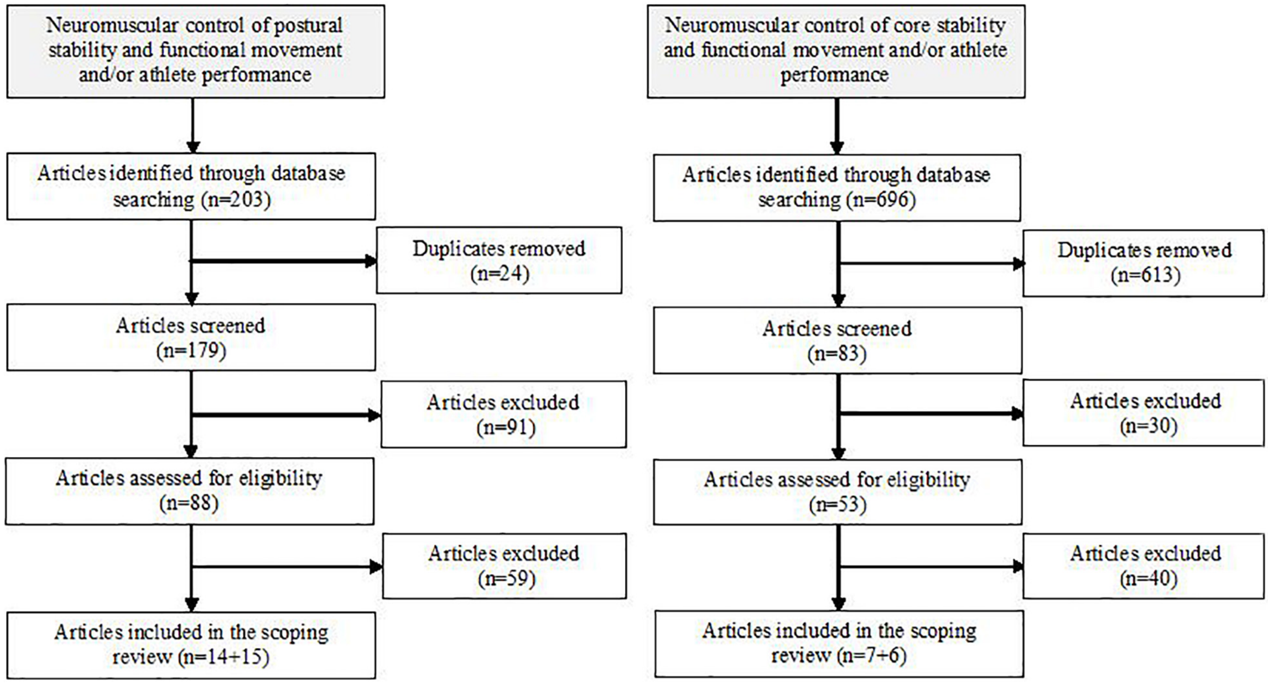

FIGURE 1 | Flow chart showing the literature search phases.

shooters than in controls indicates that they use an increased amount of vestibular and proprioceptive cues to stabilize their posture (Aalto et al., 1990). The coordination patterns of pistol motion and posture are more variable in the novice than in the skilled group (Ko et al., 2018). There are different quantitative and qualitative dynamics in pistol-aiming reflecting athlete's skill level with postural control foundation (Ko et al., 2018). The skill acquisition in pistol-aiming reduces the kinematic variables into a lower-dimensional functional unit over the posture and upper-limb system (Ko et al., 2017).

Vertical holding ability and cleanness of triggering are also important for shooting performance in the biathlon (Ihalainen et al., 2018). Postural stability in shooting direction is related to technical components of shooting, which indicates that athletes can reduce the aiming point movement in the holding and triggering phase by stabilizing their posture (Ihalainen et al., 2018). It seems that expert biathletes use a different strategy than expert rifle shooters, each of them adapting to the characteristics of their respective discipline (Larue et al., 1989).

The synchronization of bow and body sway plays a role in shot accuracy, which indicates that combined bow stability and balance exercises would contribute to better archery performance (Sarro et al., 2021). Reduced postural sway speed post-arrow release, greater bow draw force, and reduced clicker reaction time are predictors of higher scoring shots in elite recurve archers (Spratford and Campbell, 2017).

Gymnastics

Furthermore, specific postural stability control plays an essential role in acrobatic sports. Gymnastic experience during childhood is beneficial for the development of proprioceptive reweighting processes that lead to a more mature form of controlling and coordinating posture similar to adults (Busquets et al., 2021). Anthropometric characteristics and discipline-specific training experience are associated with postural steadiness (Opala-Berdzik et al., 2021). There is a relationship between anterior–posterior postural steadiness with eyes open and the artistic gymnasts' biological maturity, body mass, body height, greater age, and longer training experience (Opala-Berdzik et al., 2021). Overall postural steadiness regardless of visual conditions is associated with the acrobatic gymnasts' BMI percentiles and greater body mass (Opala-Berdzik et al., 2021).

The sport-specific task (i.e., single-leg back scale) is more sensitive in differentiating the level of expertise in young gymnasts than the simple task (i.e., bipedal standing) (Marcolin et al., 2019). While basic-level gymnasts have better postural stability in the bipedal standing, advanced-level gymnasts perform better in the single-leg back scale, particularly on balance time-dependent response to the rondade plus flic–flac (Marcolin et al., 2019). In addition, experts have a more efficient perception of body orientation in space in skills that require a fine postural adjustment than controls (Bringoux et al., 2000). Increasing expertise through specific gymnastic training increases the relevance of interoceptive and/or otolithic inputs (Bringoux et al., 2000). There is an expert advantage on sway areas for dance-like but not static postural tasks (Munzert et al., 2019). Their advantage is task specific, which provides new insights into the specificity of postural performance in highly skilled athletes (Munzert et al., 2019).

TABLE 1 | Neuromuscular control of postural stability and functional movement and/or athlete performance.

Authors	Study design	Participants	Main variables	Main findings	Postural stability control and athlete performance
Relationship between postural stability and functional movement and/or athletic performance					
Ball et al., 2003	20 shots under competition conditions	6 elite shooters	AMTI LG6-4 force plate for measuring body sway parameters; SCATT shooting analysis system for measuring aim point fluctuation and shooting performance	Body sway is related to performance in shooters; Body sway is related to aim point fluctuation in shooters; As body sway increases, performance decreases and aim point fluctuation increases for most relationships	Body sway and aim point fluctuation are essential in elite rifle shooting; Highly individual-specific are performance errors
Behm et al., 2005	Relationships between hockey skating speed and specific performance measures	30 competitive junior and secondary school hockey players	Off-ice measures including squat jump, drop jump, a 40-yd sprint, 1 RM leg press, flexibility, and balance ratio; Electromyographic (EMG) activity of the dominant vastus lateralis and biceps femoris while skating, performing a change-of-direction drill, stopping, and turning	There are significant correlations between sprint and balance tests and the skating performance; There are significant correlations between balance and players under the age of 19 years but not those over 19 years old	Significant correlations with balance suggest that stability may be associated with skating speed in younger players; Low correlations with drop jumps suggest that short contact time stretch-shortening activities is not an essential factor; EMG activities illustrate very high activation levels associated with maximum skating speed
Mononen et al., 2007	30 shots in the standing position at a distance of 10 m from the target	58 right-hand male conscripts from the Finnish Air Force Communications School	Postural balance and rifle stability assessed in terms of anteroposterior and mediolateral sway velocity of the CoP movement, and horizontal and vertical deviation of the aiming point	Shooting accuracy is related to postural balance and rifle stability, but only at the inter-individual level; There is a correlation between shooting score and behavioral performance variables; Postural balance is related to the shooting accuracy both directly and indirectly through rifle stability	High postural balance and minimal movement of the gun barrel are essential determinants of successful shooting performance among novice shooters
Edıs et al., 2016	Relationships between postural control variables and technical performance in different small-sided games (SSGs) - 1:1, 2:2 and 3:3	16 trained male amateur soccer players	Measuring of postural sway in anterior–posterior and medial–lateral directions during one-legged and both-legged quiet-stance using a Tekscan HR Mat™	There is a relationship between postural control and soccer-specific technical variables in 1:1, 2:2 and 3:3 SSGs	Higher postural control levels are essential variables that affect success in technical skills under rival pressure and suddenly changing conditions
Ihalainen et al., 2016a	A simulated air rifle shooting competition series	40 international- and national-level shooters	Optoelectronic device for measuring of shooting score and aiming point trajectory variables; Force platform for measuring of postural balance variables	Stability of hold, cleanness of triggering, timing of triggering and aiming accuracy are predictors of shooting performance, accounting for 81% of the variance in a shooting score; Direct effect of postural balance on performance is small, accounting for $<1\%$ of the variance in a shooting score; Indirect effects could be greater through a more stable holding ability that correlate with postural balance	Aiming accuracy, cleanness of triggering, and timing of triggering contribute to shooting score in elite-level air rifle shooters

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TABLE 1 | Continued

Authors	Study design	Participants	Main variables	Main findings	Postural stability control and athlete performance
Ihalainen et al., 2016b	Simulated series of air-rifle shooting-competition in three consecutive seasons	17 elite shooters	Optoelectronic shooting device for measuring of shooting score and aiming-point-trajectory variables; Force platform for measuring of postural-balance variables	Seasonal mean test results in stability of hold and cleanness of triggering are related to competition performances; Changes in stability of hold and cleanness of triggering are related to changes in performances; Cleanness of triggering is related to postural balance in shooting direction, whereas stability of hold is related to balance in cross-shooting direction	Stability of hold, cleanness of triggering, and postural balance affect performance in both training and competition situations in athletes at the elite level
Verhoeven and Newell, 2016	50 basketball free-throws with both their dominant and non-dominant hand	25 male college students with a range of skill levels	The free-throw shot recorded at 120 Hz through 8 VICON Bonita Optical motion capture cameras	Trial-to-trial variance in release parameters as well as postural stability of the shooter, synchronization of postural movement and ball release are strong predictors of performance, with non-elite shooters having a higher mean and variability of CoM speed at the time of ball release; The synchronization between the time of peak CoM and the time of ball release increases as a function of skill level and hand dominance, with the better performers releasing the ball more closely to the time of CoM peak height	The control of the trial-to-trial variability along the solution manifold of release parameters, as well as the coordination of postural control and ball release properties are important for shooting success changes as a function of skill level
Edıs et al., 2017	Relationships between parameters designating postural control levels and running speeds in SSG	16 youth soccer players	Measuring of postural sway in anterior-posterior and medial-lateral directions during one and both leg stance positions	There is a significant relationship between the running speeds of 0–6, 6–10 and 10–16 km.h^{-1} in 2 vs. 2 and 3 vs. 3 games	Combining practices that are designed to train balance with football specific exercises in a single training session can significantly contribute to athlete performance
Spratford and Campbell, 2017	The effect of postural stability pre- and post-arrow release, arrow length, flight time, draw force and clicker reaction time on scoring outcomes and the performance	39 recurve archers of an elite-level (23 male and 16 female) from four different countries prior to competition at a World Cup event	The CoP measurements 1 s prior to arrow release and 0.5 s post-arrow release using an AMTI force platform (1000 Hz); High-speed footage (200 Hz) for calculation of arrow flight time and flight score	Maximum sway speed, draw force and clicker reaction time are variables that predict performance of the shot; Higher bow draw force, reduced clicker reaction time and postural sway speed post-arrow release are predictors of higher scoring shots	The clicker time, draw force and mainly maximum sway speed post-arrow release play an important role in the scoring outcomes in elite-level recurve archery
Ihalainen et al., 2018	Factors determining performance in biathlon standing shooting at rest as well as after intense exercise	9 junior 8 national team biathletes	40 resting shots (REST) and 2 × 5 shots simulating the competition (LOAD) after 5 min of roller skiing at 95% of peak heart rate; Postural balance, aiming point trajectory and hit percentage measured from each shot	Cleanness of triggering (ATV) and vertical stability of hold (DevY) are the most important components affecting shooting performance both in REST and in LOAD; Postural balance, especially in shooting direction, is related to DevY and ATV	Cleanness of triggering and vertical holding ability are key factors in biathlon standing shooting performance; Postural balance especially in shooting direction is related to these shooting technical components; Athletes may be able to reduce the movement of the aiming point in triggering phase and in the holding phase by improving their postural stability

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TABLE 1 | Continued

Authors	Study design	Participants	Main variables	Main findings	Postural stability control and athlete performance
Caballero et al., 2021	Relationships between balance and tennis performance using linear and non-linear parameters through (1) the comparison of tennis players of different levels of expertise and ages and (2) the analysis of the association of balance and tennis serving speed and accuracy	106 recreational and expert male tennis players	Temporal dynamics of postural control during a balance task on an unstable surface analyzed through the mean velocity and the detrended fluctuation analysis (DFAV) of CoP	Traditional variables measuring balance performance only show differences according to age but not to sport performance; CoP show a reduction of auto-correlated variability with age in expert players; CoP dynamics is related to age and discriminates sport expertise	Sport experience induces balance adaptations that is characterized by a higher ability to perform postural adjustments; The lack of correlations indicates that balance measured with scattering variables under non-specific conditions is not a determinant of tennis serve performance
Lang and Zhou, 2021	60 shots under test conditions	12 elite athletes belonging to national team	Shooting score for indicating of performance; Footscan 1.0 force platform for measuring of postural balance variables; SCATT MX-02 optoelectronic training device for measuring of aiming technique parameters	Postural balance is negatively correlated with shooting score and aiming accuracy; Postural balance is positively correlated with the stability of hold and stability of triggering; There is a significant correlation between postural balance and performance, aiming accuracy and stability of hold for shooters; Postural balance is related to the stability of triggering for shooters; Postural balance is not significant with aiming time on an intra- and inter-individual basis	Postural balance is very important in aiming technique and shooting performance among elite rifle shooters
Opala-Berdzik et al., 2021	Differences in postural steadiness among young gymnasts practicing different disciplines, and their relation to the duration of their training experience, age, and their anthropometric characteristics	10 female artistic gymnasts, 10 female acrobatic gymnasts, and 10 female non-athletes	60-s quiet standing trials on a force platform with the eyes open and closed; Postural sway represented by directional components of CoP mean velocity	There are no differences in anterior-posterior (A-P) and medial-lateral (M-L) CoP mean velocities between the acrobatic and artistic gymnasts; The age, body height, body mass, duration of training experience, and maturity offset are negatively correlated with the A-P CoP mean velocity under eyes-open conditions in the artistic gymnasts; The body mass and BMI percentiles are negatively correlated with A-P and M-L CoP mean velocities under both visual conditions in the acrobatic gymnasts; The non-athletes' CoP mean velocities are non-significantly correlated with their age and anthropometric measures under both visual conditions	The artistic gymnasts' longer training experience, greater age, body height, body mass, and biological maturity are associated with better anterior-posterior postural steadiness when vision is available; The acrobatic gymnasts' greater body mass and BMI percentiles are associated with better overall postural steadiness regardless of visual conditions; There are relationships between postural steadiness and discipline-specific training experience and anthropometric characteristics

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TABLE 1 | Continued

Authors	Study design	Participants	Main variables	Main findings	Postural stability control and athlete performance
Sarro et al., 2021	The relationship between bow stability and postural control in recurve archery according to shooting performance	8 archers who participated in national-level competitions, and trained four times a week	6 shot of arrows at a 13-m distant target; 3-dimensional position of one marker attached to the bow and the CoP position of the archer measured during the aiming phase, representing bow and archer displacement, respectively	Length of the CoP trajectory (DCoP), CoP displacement in the direction across the target (CoPY), and length of the bow trajectory (Dbow) are higher in the lowest than the highest scoring shot; There is a significant correlation between CoPX and vertical displacement of the bow (DZ) during the highest scoring shot, and between CoP and bow displacement in the direction towards/away from the target (CoPX and DX)	Synchronization between body and bow sway may influence the accuracy of the shot, suggesting that combined balance and bow stability training exercises would be beneficial to improve archery performance
Effect of sport-specific training on functional movement and/or athletic performance					
Larue et al., 1989	The stability of body-gun up to the firing of a shot when shooting in standing position	2 novices and 2 experts in rifle shooting, 2 novices and 2 experts in biathlon	Electromyographic activity of the tibialis anterior, gastrocnemius and deltoid muscles; Rifle oscillation and CoP displacements	Expert biathletes use a different strategy than expert rifle shooters; No significant pattern emerges from tests with novice rifle shooters	Rifle shooters and biathletes adapt characteristics of their disciplines
Aalto et al., 1990	A simulated race	10 competition shooters	Force platform for measuring postural stability with and without competition clothing	Stability is significantly better in shooters than in untrained controls without supportive clothing; The Romberg quotient is higher in shooters than in controls, suggesting that they use an increased amount of vestibular and proprioceptive cues to stabilize their posture	Assiduous training aimed to improve balance contributes to good postural stability in shooters
Era et al., 1996	Posture control while aiming 7.5 s preceding the shot	National top-level, national and amateur rifle shooters	Speed and amplitude of the center of forces (CoF) movement	Top-level male shooters stabilize their posture better than top-level female and national level male shooters, who are more stable than naive shooters; Experienced shooters stabilize their posture better during the last seconds preceding the shot, whereas there are no differences when the successive windows are compared with each other in naive shooters; Naive shooters have more pronounced CoF movement in the less successful trials; Not-efficient whole-body posture stabilization is not a reason for a poor result in top-level shooters	Postural control is better in trained athletes who can improve their stability during the last seconds preceding the shot; A good body stabilization is the prerequisite for good shooting performance
Bringoux et al., 2000	How motor skills experts requiring a good postural control perceive their body orientation with few gravity based sensory cues	5 expert gymnasts (4 males and 1 female) and 5 non-gymnast subjects (3 males and 2 females)	The body tilt when pitching at 0.05 deg.s^{-1} in conditions of body restriction (strapped and body cast altering the somatosensory cues); The Subjective Postural Vertical (SPV) starting from different angles of pitch tilt	There is a larger body tilt when totally restrained in the body cast in controls than in experts; Controls exhibit significant errors of SPV judgmentwhereas the experts are very precise	More informative are somatosensory than otolithic cues for the body orientation perception; The efficiency of otolithic and/or interoceptive inputs can be improved through a specific training to compensate for the lack of somatosensory cues

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TABLE 1 | Continued

Authors	Study design	Participants	Main variables	Main findings	Postural stability control and athlete performance
Asseman et al., 2004	Comparing the level of performance and postural control of elite gymnasts in postures specifically trained or not	15 elite gymnasts	Surface and mean velocity of the CoP motions	The subject's level of postural performance and control in one condition is not correlated to the corresponded level in another condition; Postural ability of elite gymnasts in the handstand is not transferable to upright standing postures	Body movements and muscular force regulation to maintain balance are specific to the task characteristics; Postural performance and control in gymnastics' skills are not transferred or generalized to more usual upright stances; The level of athlete in this activity does not implicate a corresponding level in usual postures
Asseman et al., 2008	A comparison of body sway in bipedal and unipedal with eyes open and eyes closed	Two groups of 13 subjects: male elite gymnasts of international level and portsmen of regional level involved in different activities	Center of gravity motion computed from CoP motion, estimating postural control	The two groups differ significantly in the unipedal posture and with eyes open; Removal of vision affects similarly both groups	Gymnastics expertise improves postural performances in situations for which their practise is related to (i.e., unipedal with eyes open)
Croix et al., 2010	The effect of somatosensory and visual information on handstand performance; The link of general perceptual characteristics of gymnasts with their handstand performances	17 gymnasts: an expert group (6 women and 2 men), and a non-expert group (7 women and 2 men)	A handstand on a force platform in 4 conditions: open or closed eyes on a firm or foam support; The surface area covered by the CoP trajectory	Experts have significantly better postural performance during the handstand than nonexperts, whatever the visual condition, nonexperts are unable to maintain the handstand without vision, whatever the support, and the CoP surface is significantly greater on the foam surface than on the firm surface for both experts and nonexperts and, only for experts, whatever the visual condition; Experts are less field dependent than nonexperts, and the rod-and-frame test results are positively correlated with postural performance	Expert gymnasts use the remaining sensory modalities efficiently when vision is removed; Gymnastics training improves the ability to change the frame of reference
Butler et al., 2016	Differences in dynamic balance across competition levels in baseball players	90 professional (PRO), 78 collegiate (COL), and 88 high school (HS) baseball players	Lower Quarter Y-Balance Test	The PRO players exhibit greater posteromedial reach, posterolateral reach, and composite score than HS and COL groups; HS baseball players exhibit increased anterior reach compared with the COL and PRO cohorts; There are no significant differences in reach asymmetry among groups	Baseball players of different competition level differ in lower extremity dynamic balance performance
Omorczyk et al., 2018	Relationships between stability indices registered in two positions	46 athletes (23 juniors and 23 seniors) practicing gymnastics at various levels of advancement	Standing and handstand; Posturograph CQ-Stab 2P	CoP area, mean CoP amplitude, mean CoP displacement of the feet/hands in M-L direction and maximal CoP displacement of the feet/hands in M-L direction in both standing position and handstand is significantly lower in seniors; The statokinesiogram path length, both total and in A-P and M-L directions in the standing position is significantly lower in seniors	Ability to control the position of the body in both positions is better in seniors than in juniors; Stability variables in standing position significantly correlate with those in handstand in seniors but not in juniors

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TABLE 1 | Continued

Authors	Study design	Participants	Main variables	Main findings	Postural stability control and athlete performance
Jadczak et al., 2019a	Body balance control and recovery strategies in static and dynamic conditions	Professional and junior elite soccer players: 52 in PRO, 55 in U-21, 47 in U-19 group	Body balance control measured using a Delos Postural Proprioceptive System	Static and dynamic balance varies among players in different playing positions; Significantly higher differences in the static test with eyes closed are in central midfielders than goalkeepers; There is a difference in the dynamic postural priority test in favor of central midfielders relative to external defenders, central defenders, external midfielders and forward; The difference is greater in non-dominant than dominant leg	The higher the sport level of football players (PRO), the better their balance, which may contribute to more effective performance of actions related to the game and the prevention of injuries
Jadczak et al., 2019b	A comparison of balance profiles in elite soccer players across different field positions	101 elite professional soccer players (10 goalkeepers, 15 central defenders, 15 external defenders, 23 central midfielders, 15 external midfielders, 23 forwards)	Delos Postural System Test using the standard protocol (standing on a stable platform and on an unstable base unilaterally on non-dominant and dominant leg with eyes open and eyes closed)	Central midfielders have significantly higher differences than goalkeepers in the static test with eyes closed; There is a difference in favor of central midfielders relative to external defenders, central defenders, external midfielders and forwards in the dynamic postural priority test; There is a significantly greater difference in non-dominant compared to dominant leg in the dynamic postural priority test	Static balance performance and dynamic postural priority varies with playing position in elite soccer players; Midfield players have better postural priority than players in other positions; Professional soccer players present greater balance postural priority on the non-dominant leg
Marcolin et al., 2019	The effect of training on postural control in sport-specific and simple tasks	Eight female advanced-level gymnasts (ALG) and seven female basic-level gymnasts (BLG)	Bipedal standing (B) and single-leg back scale (BS) before and after two gymnastic elements (rondade plus flic-flac)	Better postural control in the B position in BLG, whereas in the BS position in ALG; CoP parameters increase after the rondade plus flic-flac in both BLG and ALG; Better performance on balance time-dependent response after the rondade plus flic-flac in BS in ALG	Postural control during the simple task (B) is not affected by expertise level; The sport-specific task (BS) is more selective in representing the level of expertise in young gymnasts
Munzert et al., 2019	Expertise-specific differences in postural tasks of various difficulty	12 intermediate non-professional and 13 professional dancers	Five dynamic dance-like and Six static everyday postural tasks	There is an expert advantage on sway area for dance-like but not for static everyday postural tasks; This effect is observed for the root mean square (RMS) velocity and RMS amplitude of the difference signal between CoP and CoG line location	The expert advantage is task-specific and provide new insights into the specificity of postural performance in experts
Caballero et al., 2020	The relationship between team-handball performance and balance ability according to expertise and age, applying a non-linear approach to balance assessment	114 male team-handball players	The CoP during a balance task; Sport performance measured by the speed and accuracy in throwing	There is a faster but not more accurate throw in expert than recreational players; Balance performance is better for 18+ than U12 players, whereas there are no differences according to their skill level; CoP velocity is slower during the balance task and moves are less irregular in players who throw with less accuracy; CoP movements are more irregular and less auto-correlated in players who throw faster	Balance performance is better in experienced team-handball players, and this is related to the maturation of the motor system more than to sport performance level; There is an exploratory behavior during balance in expert players, exhibiting more motion adjustments to reduce motor output error; A non-linear assessment reveals functional variability of balance as an intrinsic characteristic of individuals' motor control according to skill level and age

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TABLE 1 | Continued

Authors	Study design	Participants	Main variables	Main findings	Postural stability control and athlete performance
Andreeva et al., 2021	Postural stability in athletes of various sports	936 athletes of different sports and performance level	The CoP sway area (AS) and velocity (VCP) during bipedal stance with eyes open (EO) and eyes closed (EC) on a stabiloplatform (50 Hz)	VCP-EO increases in shooters, footbal players, boxers, cross-country skiers, gymnasts, runers, team sport players, wrestlers, tennis players, alpine skiers, rowers, speed skaters, and figure skaters compared to controls	Practicing sport is associated with increased postural stability in bipedal stance

Specific postural control in gymnasts' skills (i.e., postural ability in the handstand) is not transferred to basic upright stances (Asseman et al., 2004). Gymnastics expertise seems to improve postural performance only in conditions to which their practice is related (Asseman et al., 2008). There is lower frequency and variation of body sway in the handstand in more than less-experienced gymnasts (Sobera et al., 2019). While the first group concentrates on reducing anterior–posterior body sway with minimum medial–lateral movements, its values in a medial–lateral direction are irregular in the second group (Sobera et al., 2019). Postural performance in the handstand is significantly better in expert gymnasts than non-experts, whatever the visual condition (Croix et al., 2010). Experts are less field-dependent than non-experts, and this is positively correlated with postural performance (Croix et al., 2010). They use the remaining sensory modalities more efficiently under eyes closed conditions (Croix et al., 2010). The ability to change the frame of reference is improved through a high level of gymnastics training (Croix et al., 2010). Variables obtained in the handstand and standing position significantly correlate in the senior but not in the junior gymnasts (Omorczyk et al., 2018). Disabling visual control in the handstand and free-standing position in seniors deteriorates postural sway and increases CoP displacement in the anteroposterior and both directions. Lack of differences in CoP variables in the mediolateral direction in a free-standing position indicates that eye control is not important for body stability in the frontal plane in seniors practicing gymnastics CoP movement control in the mediolateral direction (Puszczałowska-Lizis and Omorczyk, 2019).

Team Sports

Postural stability control is also important for performance in team sports and may vary among athletes of different competition levels. Both dynamic and static tests should be used for the assessment of balance as postural control performance in these two cases is not related (Pau et al., 2015). For instance, the measures from the Star Excursion Balance Test may not reflect the balance performance in well-trained athletes (i.e., professional basketball and football players) who have a better balance when performing sport-related skills (Halabchi et al., 2020). However, this test includes static postures, which may better reflect postural deficits in more experienced athletes than dynamic tests (Halabchi et al., 2020).

The hockey skating performance significantly correlates with balance and sprint tests, which demonstrates the important role postural stability plays in skating speed in young players (Behm et al., 2005). Improving postural control by decreasing CoM speed at ball release is important for a higher level of shooting in basketball (Verhoeven and Newell, 2016). Incorporating postural control in the free throw shot is a critical qualitative change in coordination resulting from practice (Verhoeven and Newell, 2016). Also, volleyball players may develop a unique postural control (Borzucka et al., 2020b). Their sensory resources should be optimally distributed between sport-specific skills on the court and postural control (Borzucka et al., 2020b). They use diversified postural strategies for the maintenance of balance whereby reducing the contribution of proprioception for more challenging posture-motor tasks (Borzucka et al., 2020a). A different model of sensory integration is used by volleyball players for postural control compared to non-athletes, which may be explained by their better "dynamic" visual acuity (Agostini et al., 2013). Dynamic balance is better in professional than collegiate and high school baseball players (Butler et al., 2016).

Soccer players have superior postural control when compared to those involved in contact sport and no sport at all (Liang et al., 2019). Contact sports increase postural control through increased use of vestibular and proprioceptive information (Liang et al., 2019). Players with soccer-specific training improve executive control and proprioceptive functions, which results in better single-support balance during a dynamic visuomotor lower limb-reaching task (Snyder and Cinelli, 2020).

The contribution of vision in the maintenance of balance is less important in the professional national-level than amateur regional-level soccer players (Ben Moussa et al., 2012). Balance performance in terms of more efficient and faster stabilization after a forward jump is better in young national-level soccer players, whereas a one-leg static standing test is not sensitive enough to reveal differences in postural control associated with the combination of physical and technical features (Pau et al., 2018). Stabilometric variables improve with age until maturity (Zago et al., 2020). The higher the sport level of football players, the better their balance (Jadczak et al., 2019a). Greater balance in professional soccer players is on the non-dominant leg (Jadczak et al., 2019b). Static balance in elite soccer players varies across playing positions with better postural control in midfield players than those in other positions (Jadczak et al., 2019b). The level of playing experience influences postural control in test conditions specific to playing soccer (Paillard et al., 2006). Postural regulation changes from visual to vestibular and proprioceptive contribution, which allows better visual control of game situations in the field (Paillard et al., 2006).

Experienced team-handball players exhibit better balance performance, which is more associated with the maturation of the motor system than their performance level (Caballero et al., 2020). It seems that players with a higher level of expertise exhibit a better ability to perform motion adjustments to reduce motor output error (Caballero et al., 2020). Although postural adjustments during a balance task have a differential feature in expert players, this ability is not crucial for a tennis serve performance (Caballero et al., 2021). However, this does not reject the association of balance with other tennis drills such as pivoting maneuvers, sudden decelerations, and fast cutting maneuvers (Caballero et al., 2021).

Other Sports

Furthermore, balance, core strength and stability, flexibility, and peripheral muscle strength are associated with golf performance (Wells et al., 2009). Using concurrent mental tasks, differences in balance performance between expert surfers and controls can be found, whereas standard balance tests may not be able to elucidate whether surfing expertise facilitates balance adaptations (Chapman et al., 2008). When sharing attention with a concurrent mental task, sway path length increases in expert surfers compared to controls (Chapman et al., 2008). A different model of sensory integration was found in young kayaking and canoeing athletes than in non-athletes, which may be ascribed to a subtle re-adaptation deficit after disembarking to a stable surface with diminished sensitivity of vestibular apparatus and vision (Stambolieva et al., 2012). Better postural stability is also present in pentathletes who are less vision-dependent than untrained individuals. Conscious control of body alignment and a high level of concentration are the main factors responsible for minimizing body oscillations in pentathletes (Sadowska et al., 2019). Horseback riding may develop better postural muscle tone and particular proprioceptive abilities on standing posture during bipedal dynamic perturbations (Olivier et al., 2019). Interestingly, less anteroposterior movement during chair rising was found in master runners compared with young athletes, suggesting that they are not spared from the age-associated decline in postural stability and may benefit from specific balance training (Leightley et al., 2017).

However, some studies found no significant relationship between postural balance control and athlete performance. For instance, both the isokinetic core power and a one-legged static balance do not correlate with overall World Cup points in competitive snowboarders (Platzer et al., 2009b). Furthermore, unilateral stance with eyes closed demonstrates a positive correlation with pitch velocity, whereas there is no significant correlation between unilateral stance with eyes open or eyes closed and pitching error in college baseball pitchers (Marsh et al., 2004). Similarly, there is a lack of correlation between balance, measured with scattering variables in a non-specific task, and tennis serving speed and accuracy (Caballero et al., 2021). Furthermore, balance is not associated with team-handball performance (Caballero et al., 2020). Although the accuracy of the throws revealed a slight positive correlation with mean CoP velocity magnitude (players with better throw accuracy moved faster during the balance task), there was a negative correlation between the ball speed and bivariate variable error in experts (Caballero et al., 2020). Nevertheless, low-to-moderate correlations between unipedal balance ability and the players' technical level suggest that some technical soccer skills improve more after balance than typical soccer training (Cè et al., 2018). This discrepancy in findings may be mainly ascribed to the degree of physical development of a particular group of athletes or their exposure to sport-specific tasks. Also, a variety of methods used for balance assessment may play a role in a weak relationship between postural stability and functional movement or athlete performance. While static balance tests may be suitable for shooters, biathletes, or archers, for athletes of freestyle sports, snowboarding, skateboarding, windsurfing, or cycle acrobacy, the dynamic balance tests may represent a more appropriate alternative. Additionally, measurement of CoP variables using laboratory diagnostic systems may not be specific enough for most athletes, namely, those at a high level of competition. Moreover, postural stability may not be a key factor of athletic performance, for instance, a tennis serve or the accuracy and speed in throwing.

The Role of Neuromuscular Control of Core Stability in Functional Movement and/or Athlete Performance

Analysis of the literature revealed (**Table 2**) that out of 13 selected studies, seven (54%) investigated the relationship between core (trunk) stability-related variables and functional movement and/or athletic performance (Abt et al., 2007; Nesser et al., 2008; Nesser and Lee, 2009; Chaudhari et al., 2011; Ozmen, 2016; Anand et al., 2017; de Bruin et al., 2021). Three of them (43%) included only variables of athletic performance (Chaudhari et al., 2011; Anand et al., 2017; de Bruin et al., 2021), another three studies (43%) incorporated variables of functional movement and athletic performance (Nesser et al., 2008; Nesser and Lee, 2009; Ozmen, 2016), and one study (14%) focused on changes in the functional movement resulting from compromised core stability (Abt et al., 2007).

The remaining six studies (46%) evaluated the effects of various core or neuromuscular training programs on core stability, functional movement, and athletic performance (Stanton et al., 2004; Saeterbakken et al., 2011; Sannicandro and Cofano, 2017; Vitale et al., 2018; Kuhn et al., 2019; Felion and DeBeliso, 2020). Three of them (50%) investigated the effects of core stabilization exercises on functional movement and performance variables, strength, or core stability (Stanton et al., 2004; Sannicandro and Cofano, 2017; Kuhn et al., 2019). Two studies (33%) examined the effects of core stabilization exercises only on variables of athletic performance (Saeterbakken et al., 2011; Felion and DeBeliso, 2020). One study (17%) evaluated

TABLE 2 | Neuromuscular control of core stability and functional movement and/or athlete performance.

Authors	Study design	Participants	Methodology/Main variables	Outcomes	Main findings
Relationship between core stability and functional movement and/or athletic performance					
Abt et al., 2007	Changes in pedaling forces and lower extremity joint kinematics as a result of compromised core stability	15 cyclists, members of local road cycling team	3D Motion Analysis System, dependent kinematic variables: total frontal and sagittal plane motion of the hip and knee and total sagittal plane motion of the ankle; Core fatigue: Isokinetic Torso Rotation Test: Biodex System 3 Multi-Joint Testing and Rehabilitation System; Core fatigue workout: 32 min. circuit of 7 exercises targeting core stabilizer muscles	After the core fatigue workout: significant decrease (30.0–43.3%) in peak torque, total work, average power, maximal repetition total work, and average peak torque; an increase in total frontal plane knee motion and total sagittal plane knee and ankle motion (13.4–54.3%); no significant differences for any pedal force or work data	Core fatigue results in altered cycling mechanics that could increase the risk of knee injury; Improved core stability and endurance could promote greater alignment of the lower extremity when riding for extended duration as the core is more resistant to fatigue
Nesser et al., 2008	The relationships between core stability and various strength and power variables in strength and power athletes	29 male football players of the National Collegiate Athletic Association Division I	3 strength variables: 1 RM squat, 1 RM bench press, 1 MR power clean; 4 performance variables: countermovement jump, 20- and 40-yd sprints, 10-yd shuttle run; Core stability: trunk flexion, back extension, and left and right bridge	There is a number of significant but not consistent and weak to moderate correlations between core strength/stability and strength and performance measures	Significant correlations between core strength/stability, even weak to moderate, suggesting that core strength/stability contributes to strength and power performance
Nesser and Lee, 2009	The relationship between core stability and various strength and power variables	16 National Collegiate Athletic Association Division I female football players trained specifically for strength and power	2 strength variables: 1RM squat, 1RM bench press; 3 performance variables: countermovement jump, 10-yd shuttle run, 40-yd sprints; Core stability: trunk flexion, back extension, left and right bridge	There are no significant correlations between core strength/stability and the strength and performance measures	Determination of the effectiveness of core strength or stability requires further research and sport-specific means
Chaudhari et al., 2011	The relationship between lumbopelvic control and pitching performance	48 pitchers who pitched 50 or more innings in Minor League competition of A, AA, or AAA levels	Lumbopelvic control: Level Belt secured to the waist, transition from two-leg to single-leg pitching stance, balance while maintaining a stable pelvic position; Pitching performance: number of innings pitched (IP) during season; Median Level Belt score for the study group 7°	Significantly fewer walks plus hits per inning and significantly more IP during the season in subjects scoring $<7°$ on the Level Belt test than those scoring $>7°$	Lumbopelvic control influences performance of baseball pitchers; Simple test of lumbopelvic control can identify individuals with better chance of pitching success
Ozmen, 2016	The relationships between core stability, jumping performance and dynamic balance	17 male soccer players	Dynamic balance: Star excursion balance test (SEBT); Core stability: McGill's protocol; Jumping ability: squat jump test on contact mat	Significant negative correlation between trunk flexion test and jumping height ($r = -0.705$); No significant correlation between side bridge, trunk extension tests and jumping height, and between trunk flexion, side bridge, trunk extension tests and SEBT results	Trunk flexion is associated with squat jump height but not with side bridge and trunk extension tests; Core stability does not contribute to dynamic balance
Anand et al., 2017	The relationship between bowling speed in cricket and core stability	82 cricket medium and medium fast bowlers of district and universities	Core stability: plank test (prone plank, left side plank and right side plank); Bowling speed: BUSHNELL Velocity Speed Gun	There is significant positive correlation between core stability and the bowling speed ($r = 0.736$)	Bowling speed is significantly higher in subjects with well-developed than poorly-developed core stability

(Continued)

TABLE 2 | Continued

Authors	Study design	Participants	Methodology / Main variables	Outcomes	Main findings
de Bruin et al., 2021	The relationship between athletic performance and core stability	83 female athletes from the university teams: hockey (*n* = 24), netball (*n* = 16), running (*n* = 11), soccer (*n* = 15), and tennis (*n* = 17)	Core strength and endurance: Biering-Sørensen tests - isometric back extension (IBE), lateral flexion (LF) and abdominal flexion (AF); Core neuromuscular control (NMC): Welch Allyn FlexiPort pressure biofeedback unit; Athletic performance: T-test, 40 m sprint, medicine ball chest throw (MBCT) and vertical jump (VJ)	Most weak correlations in all sports (r = 0.10–0.39); Very strong correlation between VJ and LF (r = 0.90); Moderate correlations in all sports between core strength, endurance and motor control and certain athletic performance tests (r = 0.40–0.69)	Correlations between core stability and athletic performance are negligible or weak; Athletic performance in different sports is associated with different components of core stability
Effect of core stability training on functional movement and/or athletic performance					
Stanton et al., 2004	The effect of short term Swiss ball training (SBT) on core stability and running economy	18 male athletes from Basketball and Touch Football School of Excellence in Sport program: EG (*n* = 8), CG (*n* = 10)	SBT sessions 6 weeks, two times a week, approximately 25 min. during regular training, supervised by researcher; Core stability: 5 level Sahrmann core stability test with Stabilizer Pressure Biofeedback Unit; Maximal aerobic power (VO_2max) and running economy (RE): incremental treadmill running test to volitional exhaustion	Significant effect of SBT on core stability in the EG; No significant differences in myoelectric activity of abdominal and back muscles, treadmill VO_2max, RE, or running posture in both EG and CG	SBT has positive effect on core stability without improvements of physical performance
Saeterbakken et al., 2011	The effect of core stabilization training (CST) on maximal throwing velocity	24 female high-school handball players randomly divided into a CST (*n* = 14) and a control group (CG, *n* = 10)	6-week regular handball training in both groups; Additional progressive core stability training program in the CST group (twice a week for 75-min, 6 unstable closed kinetic chain exercises); Throwing velocity: 2 photocell arrays with an accuracy of ± 0.001 s	There is a significant increase of maximal throwing velocity in the CST group (4.9%) but not in the CG	CST using unstable, closed kinetic chain movements improves maximal throwing velocity; Stronger and more stable lumbopelvic hip complex may contribute to higher rotational velocity in multisegmental movements
Sannicandro and Cofano, 2017	The effects of integrative training of core stability on jump performance	44 young basketball players (19 female, 25 male); EG, *n* = 21 (11 female, 10 male), CG, *n* = 23 (11 female, 12 male)	4-week CST in stable and unstable conditions during warm-up (8 sessions, twice a week), followed by basketball drills with CG (60 min); Jump performance: monopodalic jumps (triple hop test, side hop test, and 6m timed hop test) and bipodalic jump (Seargent vertical jump)	Significant improvements in the right and left hop test, the 6m-timed hop left and right test in the EG; A significant improvement in vertical jump in the CG	Core stability program is effective in improving monopodalic jump ability in prepubertal basketball players
Vitale et al., 2018	The effects of neuromuscular training program on dynamic balance and vertical jump performance	24 elite junior male skiers randomized in an experimental group (EG, *n* = 12) and a control group (CG, *n* = 12)	8-week training program (16 sessions, 3 phases); partly different exercises on core stability, body-weight strengthening and plyometric exercises on dynamic postural control and vertical jump performance in each phase; circuit training form during warming up (30 min); Dynamic balance: lower quarter Y-Balance test (YBT) with standardized testing protocol; Jumping performance: countermovement (CMJ) and drop jump (DJ) on Optojump Next	Positive effects on pre to post measures in anterior, postero-medial, postero-lateral directions, and composite YBT score for both lower limbs in the EG; No significant changes in the CG; No significant changes in CMJ and DJ in both EG and CG	There is a positive effect of neuromuscular training on dynamic balance ability but not on vertical jump performance; It may be effective in increasing lower limb joint awareness and postural control

(Continued)

TABLE 2 | Continued

Authors	Study design	Participants	Methodology / Main variables	Outcomes	Main findings
Kuhn et al., 2019	The effects of core stabilization training (CST) on maximal throwing velocity and core strength parameters	20 female handball players from German non-elite handball squad	6-week CST (twice a week for 45 min., 9 specific core and rotational exercises for ventral, dorsal and lateral core muscles chain on an unstable surface); Maximum voluntary isometric strength (MIS) of the trunk using isometric dynamometer Beck-check 607; Endurance strength of ventral, dorsal and lateral core chains using a Swiss Olympic Medical Center core performance test battery; Throwing velocity using OPTOJump Next	A significant improvement in MIS of left lateral core muscle chain in the EGcompared to the CG; A significant improvement in MIS of ventral core endurance (35%) and the lateral right core muscles (21%) in the EG; A significant increase in throwing velocity of jump throw in both EG (12%) and CG (8%) but not velocity of standing throw	CST effectively increases isometric strength and endurance of core muscles but does not enhance throwing velocity when compared to standard training
Felion and DeBeliso, 2020	The effect of core training (CT) program on force production in torsional movements	Students, members of baseball team at Granger HS, UT, United States	Experimental group (EG): 6-week CT program (twice a week, 1 h/day), in addition to specific training; Control group (CG): 6-week baseball specific training only (twice a week, 2 h/day). Throwing velocity (TV) and ball-exit velocity (BEV) using Stalker Sport II radar gun; BEV: speed of the ball immediately after being struck by the baseball bat	Neither EG nor CG increase in TV following the 6-week CT intervention; A significant increase in BEV in the EG but not in the CG	Implementing of CT with additional rotational exercises with free weights, resistance bands, or medicine balls leads to additional gains in torso rotational strength and potentially improvement in BEV

the effect of neuromuscular training on selected parameters of functional movement (Vitale et al., 2018).

Regarding the sport, eleven studies (85%) were conducted in team sports, such as baseball, basketball, cricket, football, handball, soccer, and touch ball (Stanton et al., 2004; Nesser et al., 2008; Nesser and Lee, 2009; Chaudhari et al., 2011; Saeterbakken et al., 2011; Ozmen, 2016; Anand et al., 2017; Sannicandro and Cofano, 2017; Kuhn et al., 2019; Felion and DeBeliso, 2020; de Bruin et al., 2021) and two studies (15%) were carried out in individual sports, such as cycling and skiing (Abt et al., 2007; Vitale et al., 2018).

The Relationship Between Core Stability and Functional Movement and/or Athletic Performance

Among seven studies, six investigated the association of core stability with variables of athletic performance (Chaudhari et al., 2011; Anand et al., 2017; de Bruin et al., 2021) or both functional movement and athletic performance (Nesser et al., 2008; Nesser and Lee, 2009; Ozmen, 2016), whereas one study dealt with changes in functional movement resulting from compromised core stability (Abt et al., 2007).

The most investigated characteristics of core stability (Nesser et al., 2008; Nesser and Lee, 2009; Ozmen, 2016; Anand et al., 2017) were core or lumbopelvic neuromuscular control (Chaudhari et al., 2011; de Bruin et al., 2021), and core strength and endurance (Abt et al., 2007; de Bruin et al., 2021). Among functional movement characteristics, it was the kinematics of movement (Abt et al., 2007) and jumping abilities that stood out (Ozmen, 2016; de Bruin et al., 2021), whereas factors of athletic performance included ball speed (Anand et al., 2017), running speed, agility, and explosiveness of upper body (de Bruin et al., 2021). All studies dealing with the association of core stability with functional movement and athletic performance used a cross-sectional design. In all seven studies, only one selected group of athletes of a certain type of sport was tested.

Regarding the methodology of core stability characteristics, the following tests were used: trunk flexion, back extension, left and right bridge (Nesser et al., 2008; Nesser and Lee, 2009), core stability McGills protocol (Ozmen, 2016), prone plank, left and right side plank (Anand et al., 2017), and Biering-Sørensen test (de Bruin et al., 2021). The “Level belt” was used for the lumbopelvic control (Chaudhari et al., 2011), isokinetic torso rotation test on a Biodex system and 32 min circuit of exercises targeting the core muscles evaluated core muscle fatigue (Abt et al., 2007), and biofeedback unit was applied for the core neuromuscular control (Ozmen, 2016). Regarding the functional movement characteristics, three-dimensional motion analysis (Abt et al., 2007) and star excursion balance test for dynamic balance (Ozmen, 2016) were used. Athletic performance characteristics were evaluated using the radar speed gun (Anand et al., 2017), 20-m run, 40-m run, *T*-test, agility test, shuttle run, medicine ball throw (Nesser et al., 2008; Nesser and Lee, 2009; de Bruin et al., 2021), squat jump (Ozmen, 2016), and the number of innings pitched during a season (Chaudhari et al., 2011).

Core stability provides a foundation for force production in the lower and upper limbs (Willardson, 2007). This is a requisite for optimal functional movement and consequently also for better athletic performance (Abt et al., 2007; Chaudhari et al., 2011; Anand et al., 2017). However, some studies do not find this link between core functions and the movement of lower and upper limbs. In general, two research approaches exist that examine the association of core stability (lumbopelvic control) with functional movement and athletic performance. Some studies examined the importance of core stability or lumbopelvic control using strength, endurance, agility, speed, or other physical abilities tests as surrogate measures of functional movement and athletic performance (Nesser et al., 2008; Nesser and Lee, 2009; Ozmen, 2016; de Bruin et al., 2021). It has been proposed that well-trained athletes have general level of abilities, such as agility, explosive power, and speed, in addition to core stability, regardless of the specificity of the sport, and that there is a relationship between them. Other studies used direct measures of functional movement or athletic performance (Abt et al., 2007; Chaudhari et al., 2011; Anand et al., 2017). For instance, investigating the relationship between cycling mechanics and core stability revealed that improved core stability and endurance could promote greater alignment of the lower extremity when riding for extended durations as the core is more resistant to fatigue (Abt et al., 2007). Lumbopelvic control influences overall performance for baseball pitchers, thus a simple test of lumbopelvic control can potentially identify individuals who have a better chance of pitching success (Chaudhari et al., 2011). Throwing accuracy is significantly better in cricket bowlers with well-developed than poorly developed core stability (Anand et al., 2017).

The association of core stability with some variables of athletic performance in sports such as hockey, netball, tennis, soccer, and running supports the fact that its significance regarding some motor abilities in particular sports partly differs. However, when these sports were analyzed separately, there were similar moderate correlations between core strength or endurance and motor abilities in the tests used (de Bruin et al., 2021). There were strong correlations between abdominal flexion endurance and the vertical jump in runners, and between isometric back extension strength and the sprint in tennis players. However, the core strength and/or stability does not correlate with the strength and performance measures (10 yard shuttle run, 40 yard sprint, countermovement jump, 1RM squat, 1RM bench press) in athletes who train for strength and football skills. Despite these non-significant correlations, it is not reasonable to neglect the core. Nonetheless, it seems that the core musculature is no more important than any other part of the body (Nesser and Lee, 2009).

A belief that core stability is important for strength and power production in sport was not corroborated in male football players. The core stability was significantly but not strongly correlated with power and strength variables (10-yd shuttle run, 20- and 40-yd sprints, countermovement jump, 1RM squat, 1RM bench press, 1RM power clean). Correlations between core stability and strength or power, and sprints or shuttle run were moderate to weak but significant. This indicates that core strength contributes to power and strength performance and therefore should be taken into account (Nesser et al., 2008). However, there was a negative correlation between the jump height and trunk flexion, and no significant association was found between the jump height and side-bridge trunk extension in male soccer players. Similarly, the relationship between core stability and dynamic balance was not significant (Ozmen, 2016). These findings indicate that an understanding of the role of core stability in body movements most likely requires testing under sport-specific conditions. All of the athletic performance measures were mainly one repetition of explosive movements or sprints lasting a few seconds. The core stability was evaluated using isometric muscle contractions or muscle endurance tests. However, the core stability and performance of these two variables should not be compared. While sub-maximal muscle contractions, activation of more slow-twitch muscle fibers, and anaerobic glycolysis are typical for most core stability tests, the agility, power, strength, and running tests involve primarily maximum force production, activation of fast twitch muscle fibers, and the ATP–CP energy system (Nesser et al., 2008; Nesser and Lee, 2009; de Bruin et al., 2021).

The second type of study represents relationships between core stability and bowling speed in cricket, walks, hits per innings pitched and total innings pitched in baseball and functional movement in cycling. Cricket players with well-developed stability of the core manifest high quality in the kinetic chain of movements when bowling that probably results in increased bowling speed. They can better control their trunk position and motion over the pelvis and leg. This allows optimum generation and transfer of force to the terminal segment in the kinetic chain of specific movements. Core stability provides integration of proximal and distal segments in increasing bowling speed (Anand et al., 2017). Lumbopelvic control is related to performance in baseball pitchers (Chaudhari et al., 2011). The study revealed differences between lumbopelvic control and walks plus hits per innings pitched and total innings pitched. Significantly lower walks plus hits per innings pitched were found in the group with better than those with poorer lumbopelvic control. Furthermore, core stability also plays a role in the functional movement in cycling. For instance, lower extremity cycling mechanics is influenced by the core fatigue workout. Several kinematic variables were altered whereas work variables and the pedal force remained unchanged (Abt et al., 2007).

Core Stability Training and Functional Movement and/or Athletic Performance

Training programs were usually aimed at the increase of athletic performance factors (Saeterbakken et al., 2011; Kuhn et al., 2019) or variables of functional movement (Stanton et al., 2004; Sannicandro and Cofano, 2017; Vitale et al., 2018; Felion and DeBeliso, 2020) and were often combined with the development of core stability and strength (Stanton et al., 2004; Kuhn et al., 2019; Felion and DeBeliso, 2020) or the dynamic balance (Vitale et al., 2018).

The duration of intervention was from 4 to 6 weeks (Stanton et al., 2004; Saeterbakken et al., 2011; Sannicandro and Cofano, 2017; Kuhn et al., 2019; Felion and DeBeliso, 2020) or 8 weeks

(Vitale et al., 2018), two times per week with a duration of 25–45 min (Stanton et al., 2004; Sannicandro and Cofano, 2017; Vitale et al., 2018; Kuhn et al., 2019) to 60–75 min (Saeterbakken et al., 2011; Felion and DeBeliso, 2020). While the 25–30 min programs were a part of warming up, the 45–75 min programs were organized apart from standard training. Core stabilization training programs were supervised by coaches, conditioning specialists, or researchers.

Core stabilization training programs included core exercises (Stanton et al., 2004; Saeterbakken et al., 2011; Sannicandro and Cofano, 2017; Kuhn et al., 2019; Felion and DeBeliso, 2020), or core exercises combined with plyometrics and body strengthening (Vitale et al., 2018). These exercises were often performed in unstable conditions or in both stable and unstable conditions (Stanton et al., 2004; Sannicandro and Cofano, 2017; Vitale et al., 2018; Kuhn et al., 2019).

The assessment of athletic performance or measurement of functional movement variables was focused on throwing velocity (Saeterbakken et al., 2011; Kuhn et al., 2019; Felion and DeBeliso, 2020) and jumping performance (Vitale et al., 2018). Running economy was assessed with a test to exhaustion on a treadmill (Stanton et al., 2004). The core assessment included the Swiss Olympic test (Kuhn et al., 2019) and the Sahrmann core stability test (Stanton et al., 2004); dynamic balance was tested by means of Y-Balance test (Vitale et al., 2018); and jump abilities by using the side hop test, triple hop test, 6-m timed hop test, and the Sargent vertical jump test (Sannicandro and Cofano, 2017).

Core stability and core strength training have been used for the improvement of functional movement and consequently also athletic performance. The purpose of core stability exercises is to control the lumbar spine, whereas core strength exercises improve the transfer of muscle power, activation of local stabilizers, and global mobilizers (Saeterbakken et al., 2011; Sharrock et al., 2011; Sannicandro and Cofano, 2017). The training programs incorporating core stability exercises performed under stable or unstable conditions showed improvements in core muscle strength, muscular endurance, and body balance (Stanton et al., 2004; Vitale et al., 2018; Kuhn et al., 2019). However, there is a controversy as to whether an increase in core stability and strength is transferred to athletic performance.

For instance, a 6-week isolated resistance training program in young baseball players did not improve the throwing velocity in contrast to the ball-exit velocity (Felion and DeBeliso, 2020). Similar findings were found also after a 6-week core stabilization training in adult female handball players. Both experimental and control groups significantly increased throwing velocity of the jump throw, but their throwing velocity of the standing throw remained unchanged (Kuhn et al., 2019). Furthermore, an integrated short-term Swiss ball training failed to enhance the running economy measured by VO_2max, vVO_2max, or running economy at speeds of 60, 70, 80, or 90% vVO_2max (Stanton et al., 2004). On the other hand, an isolated progressive core stability training in unstable conditions improved the throwing velocity significantly in young handball players (Saeterbakken et al., 2011). Also, an 8-week integrated neuromuscular training focused on core stability, plyometrics, and dynamic postural control led to an improvement of postural stability but not of jump performance in junior alpine skiers (Vitale et al., 2018). Furthermore, a 4-week integrated core stabilization program improved the one-leg jump abilities but not the bipedal vertical jump in the prepubertal athletes (Sannicandro and Cofano, 2017).

Depending on how the special core stabilization programs were integrated into standard training, it is possible to distinguish two variants, that is, either integrated programs conducted during warm-ups within a training session (Stanton et al., 2004; Sannicandro and Cofano, 2017; Vitale et al., 2018; Kuhn et al., 2019) or isolated additional programs carried out as the standard training (Felion and DeBeliso, 2020; Saeterbakken et al., 2011). With regard to these differences in integrated and isolated core stabilization training programs, findings in the literature are not consistent.

Most studies investigated the association of core stability with functional movements and athletic performance or the effect of specific core stabilization programs on functional movement and athletic performance in junior or younger age groups (Stanton et al., 2004; Nesser et al., 2008; Nesser and Lee, 2009; Chaudhari et al., 2011; Saeterbakken et al., 2011; Anand et al., 2017; Sannicandro and Cofano, 2017; Vitale et al., 2018; Felion and DeBeliso, 2020; de Bruin et al., 2021). However, only a few studies have investigated the role of core stability in the functional movement and athletic performance in adult athletes and the changes induced by core stabilization training (Abt et al., 2007; Chaudhari et al., 2011; Ozmen, 2016; Kuhn et al., 2019). The reason for this disproportionality could be the accessibility of young athletes compared to the elite ones for participating in intervention studies.

Limitations in the Current Studies Investigating the Relationship Between Postural and/or Core Stability and Athlete Performance and Proposals for Further Research

An analysis of the literature revealed several gaps in the existing studies (**Table 3**). There is still a lack of research that seeks to investigate the relationship of body balance and stability of the core with sport-specific performance. Although the importance of the core musculature for spine stabilization and postural control has been emphasized during the past decade, the supporting evidence is still scarce. Recently, increased research efforts have been accomplished to investigate effective exercise programs for improving spinal stability and body balance. Practitioners suggest that a strong core could contribute to better balance and proper posture with a positive impact on increasing their athletic performance and/or decreasing the occurrence of back pain. While postural and core stability may be a key factor in the prevention of musculoskeletal disorders, it seems that much less evidence exists on their role in sport-specific performance and related functional movements. These gaps revealed in the literature should be addressed in future studies.

TABLE 3 | Research gaps identified in the literature and suggestions for future studies.

Gaps and limitations revealed in the literature	Suggestions for future studies
There is a lack of studies investigating the relationship between core stability or core strength and functional movement and/or athletic performance.	In comparison with balance research, more attention should be paid to investigations related to the role of core stability and core strength in functional movement and/or athletic performance.
There is still inconsistent definition of core stability and core strength in spite of an increased number of studies in this field of research.	The authors should use uniform terminology of core stability and core strength based on their characteristics, similarly as it is in the case of balance research.
The research in this field has been conducted mainly in shooting, biathlon, archery, gymnastics, and team sports. More research has been carried out in team than individual sports.	As core stability and strength represent an integral part of athlete performance in sports based on lifting tasks and trunk rotations, their role in performance in acrobatic, combat, power and water sports should also be investigated.
Small sample sizes occur in most studies, which could reduce its power and increase the margin of error.	The number of participants in research studies should be increased.
There are only few studies conducted on competitive athletes of a high performance level. The analysis of the literature in this scoping review related to the role of core stability in functional movement and/or athlete performance revealed that nine studies (69%) were conducted with regularly competing high school or university athletes or athletes from lower competitions (Stanton et al., 2004; Abt et al., 2007; Chaudhari et al., 2011; Saeterbakken et al., 2011; Ozmen, 2016; Anand et al., 2017; Kuhn et al., 2019; Felion and DeBeliso, 2020; de Bruin et al., 2021), three studies (23%) with young elite athletes (Nesser et al., 2008; Nesser and Lee, 2009; Vitale et al., 2018), and one study (8%) with very young athletes (Sannicandro and Cofano, 2017). Regarding the age of participants, eight studies (62%) included adult athletes (Abt et al., 2007; Nesser et al., 2008; Nesser and Lee, 2009; Chaudhari et al., 2011; Ozmen, 2016; Anand et al., 2017; Kuhn et al., 2019; de Bruin et al., 2021) and five studies involved young athletes (Stanton et al., 2004; Saeterbakken et al., 2011; Sannicandro and Cofano, 2017; Vitale et al., 2018; Felion and DeBeliso, 2020).	The research studies should include more elite athletes. In such a case, adults should be preferred over young participants.
There is a lesser number of research studies conducted on female than male participants. Regarding gender, the analysis of the literature related to the role of core stability in functional movement and/or athlete performance revealed that eight studies (62%) included male athletes (Stanton et al., 2004; Abt et al., 2007; Nesser et al., 2008; Chaudhari et al., 2011; Ozmen, 2016; Anand et al., 2017; Vitale et al., 2018; Felion and DeBeliso, 2020), four studies (31%) female athletes (Nesser and Lee, 2009; Saeterbakken et al., 2011; Kuhn et al., 2019; de Bruin et al., 2021), and one study (7%) both girls and boys (Sannicandro and Cofano, 2017).	The number of female participants should be increased.
The control group is rarely included. Non-sporting population cannot be included in most of the studies because of the athletic performance tests used.	The control group should be included, especially in intervention studies.
There is a missing information on the level of athlete performance.	Information on the degree of physical development of athletes and their exposure to sport-specific tasks should be included.
General physical fitness tests rather than sport-specific tests are used.	Better understanding the role of postural and core stability in athletic performance requires testing under conditions specific to a particular sport. Therefore, testing under sport-specific conditions should be preferred, particularly in athletes at a high level of competition.
Variables analyzed are not precisely described.	Corresponding variables should be better specified in relation to functional movements in sports with high demands on postural and core stability.
Experiments are conducted in different training periods (pre-season, in-season, post-season) what limits a comparison of findings.	Studies investigating the association of postural and core stability with functional movement and performance in athletes should be carried out during a period of their high level of sport-specific skills.
The average duration of training programs is from 4 to 8 weeks while the training sessions vary from 25 to 30 min, in some cases even from 60 to 75 min, which very often combine balance, strength and core muscle exercises.	The duration of training sessions and training programs should be separately specified for balance exercises and core stabilization or core strengthening exercises.
There is insufficient analysis of neuromuscular mechanisms underlying significant associations of postural and core stability with functional movement and/or athlete performance.	Greater attention should be paid to the interpretation of findings.
Balance, strength, plyometric or endurance exercises are usually associated with training induced improvements of postural and core stability but not with athlete performance.	Further research is needed to investigate the relationship between postural or core stability and sport-specific performance and their changes after sport-specific training.

CONCLUSION

This scoping review revealed that among a variety of studies investigating the role of neuromuscular control of postural and core stability in functional movement and/or athlete performance, only a few revealed the relationships between them. Postural stability was found to play an essential role in performance in archery, biathlon, gymnastics, shooting, and team sports (e.g., basketball, hockey, soccer, tennis). Also, core stability and strength represent an integral part of athlete performance in sports based on lifting tasks and trunk rotations. Variables of these abilities are associated with performance-related skills in cricket, cycling, running, and team sports (e.g., baseball, football, hockey, netball, soccer, tennis). Better neuromuscular control of postural and core stability contribute to more efficient functional movements specific to particular sports. Training programs incorporating general and sport-specific exercises that involve the use of postural and core muscles showed an improvement of body balance, back muscle strength, and endurance. However, there is controversy about whether the improvement in these abilities is translated into athletic performance. Identifying the relationship of passive, active, and neural mechanisms underlying balance control and spinal posture with athlete performance would provide a basis for a multifaced approach in designing training and testing tools addressing postural and core stability in athletes under sport-specific conditions.

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Both authors have made a substantial, direct, and intellectual contribution to the work, and approved it for publication.

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Effects of Hip Joint Angle on Quadriceps Recruitment Pattern During Knee Extension in Healthy Individuals: Analysis by Ultrasound-Based Shear-Wave Elastography

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Purposes: To detect the effects of hip joint position on the quadriceps recruitment pattern of different resistance levels of rectus femoris (RF), vastus intermedius (VI), vastus lateralis (VL), and vastus medialis obliquus (VMO) in healthy people during knee extension.

Methods: Twenty healthy females performed isometric knee extension contractions at 0, 10, 20, and 30% of maximal voluntary isometric contraction (MVIC) with a 90° and 0° hip angle. Ultrasound shear-wave elastography was used to evaluate the shear elastic modulus of RF, VI, VL, and VMO during resting and contraction states.

Results: At resting state, stiffness of RF was about 50% higher at 0° compared with at 90° of the hip ($p < 0.01$). There were significant differences in comparisons between 0 and 10% MVIC, 10 and 20% MVIC, and 20 and 30% MVIC in the four muscles, except that there was no significant difference between 20 and 30% MVIC for RF. There was a significant positive correlation between muscle stiffness and resistance level ($r = 0.78–0.94$, $p < 0.001$).

Conclusions: Hip joint position had effects on the quadriceps recruitment pattern of different resistance levels in healthy people during knee extension.

Keywords: quadriceps, shearwave elastography, shear modulus, hip angle, maximal voluntary isometric contraction

INTRODUCTION

The quadriceps are the most important muscle group in our human thighs. It is not only responsible for the main force when we walk, go up and down stairs, squat up, etc., but also for the stability of our knee joints, especially the stability of the patella and knee joint in the anterior and posterior directions, so if the quadriceps muscle atrophies significantly, the most common is the medial head (the quadriceps is divided into four heads: medial head, lateral head, intermedius, and rectus femoris), likely to cause knee imbalance, strength loss, patellar instability, and other functional imbalances can further lead to accelerated abrasion of the articular cartilage and degenerative changes in the knee joint (eg, premature aging) and many other related diseases (Pietrosimone et al., 2019). Therefore, it is very important to maintain normal muscle condition and strength in the quadriceps. Quadriceps strength can be restored and strengthened with exercise rehabilitation training and isometric knee extension to increase joint stability and improve knee function (Herrera H et al., 2020). Ordinary people should practice on a regular basis to maintain normal functions such as going up and down stairs and squatting that we require in our daily lives; people with knee joint diseases should practice more to improve the knee joint's protection ability and motor function, which can also promote recovery from knee joint injury.

Exercise is one of the most important ways to build muscles, and it has been shown to be effective in increasing strength, reducing pain, and improving function (Jakobsen et al., 2019). However, the quadriceps consists of four muscles: rectus femoris (RF), vastus intermedius (VI), vastus lateralis (VL), and vastus medialis (VM), which differ in their ability to contract. The role of the vastus medialis oblique muscle (VMO) is to stabilize the medial side of the patella and avoid the increased lateral joint contact stress and decreased medial side due to lateral translation and tilting of the patella during knee extension, so we chose the VMO as the target muscle (Stephen et al., 2018). Since RF spans the knee and hip joints, the method of muscle recruitment for knee extension should also depend on hip angle, and it is unclear whether hip angle affects isometric knee extension (Ema et al., 2017). Proprioceptive and vestibular inputs of the hip are important for lower extremity muscle activity (Lewek et al., 2006). The position of the hip joint modulates the contraction of the associated muscles that control the movement of the knee joint. It was found that thigh muscles are activated by about 20% MVIC most of the time in our daily life (Otsuka et al., 2019). Thus, we investigated the variation of shear modulus during 10–30% of MVIC on quadriceps performing isometric knee extension using ultrasound shear-wave elastography.

Ultrasound shear-wave elastography (SWE) is a useful technique to reflect the function of muscle contraction by quantitatively evaluating the shear elastic modulus in skeletal muscle (Botanlioglu et al., 2013). Studies have shown that during knee MVIC extension, the muscle becomes stiffer as the level of muscle contraction increases, positively correlated with the relative isometric level of the knee extensors (Andonian et al., 2016; Wang et al., 2017; Otsuka et al., 2019). On the other hand, hand-held dynamometry for isometric knee flexor strength assessment was found to have good intertester reliability with an ICC range of 0.80–0.87 (van der Made et al., 2019). Another study showed that hand-held dynamometry demonstrated moderate to excellent intra- and inter-rater reliability for the assessment of isometric knee extensor muscle strength in a healthy population (Mentiplay et al., 2015). However, the research on the isometric resistance contraction of the quadriceps by the angle of the hip joint has not been confirmed, so this study mainly explored the effect of changing the angle of the hip joint on the isometric resistance contraction of the quadriceps.

MATERIALS AND METHODS

Subjects

Twenty healthy female subjects participated in the present study (aged: 20.75 ± 2.02 years; height: 1.60 ± 0.06 m; body mass: 51.64 ± 5.10 kg). Subjects who had surgery or neurological disease in the lower limbs were excluded (Kawai et al., 2018). The procedure and purposes of this study were explained to the subjects, and written informed consent was obtained. They were asked to avoid participating in any training the day before the experiment (Wang et al., 2017).

Data Collection

We used a randomized repeated-measures experimental design to compare the effects of two hip joint angles (0° *vs*. 90°) on the contracted state (0% MVIC *vs*. 10% MVIC vs. 20% MVIC *vs*. 30% MVIC) and the shear elastic modulus of the quadriceps muscle (RF, VI, VL, and VMO), as shown in **Figure 1**.

Subjects were placed on a bed with the knee at 90° of flexion (0° = full knee extension) while hip joint angles at 90° of flexion (neutral position) or 0° of flexion (supine position). To assess isometric quadriceps strength, a hand-held dynamometer Hoggan MicroFET2 (Hoggan Scientific, LLC, UT, United States) was placed on the anterior aspect of the shank, proximal to the ankle joint (Mentiplay et al., 2015). Prior to data collection, subjects were asked to perform several submaximal isometric quadriceps contractions (~75% effort) with the dominant leg (the one used for kicking the ball) to familiarize them with testing procedures (Lewek et al., 2006). Each subject produced three maximal voluntary isometric quadriceps contractions (MVICs) at each of two hip positions, taking the average of three MVICs as the mean MVIC valueat one hip position for a total of six MVICs. The testing order of the different hip positions was randomized among subjects, with each subject producing all three MVICs with the hip in one condition (e.g., either 90° or 0°) before progressing to the next randomly determined position. During the test, the researcher obtained real-time feedback of the MVIC value by observing the dynamometer screen and verbally encouraged the subject to produce the maximum value. The ergometer value did not diminish over the three trials at each hip position, suggesting that subjects were given sufficient rest to avoid fatigue. The mean MVIC value was used to calculate the resistance value required for each contraction, and then subjects randomly performed four tasks with MVIC of 0, 10, 20, and 30%. Calculate as follows: 10% MVIC = 0.1MVIC, 20% MVIC = 0.2MVIC, 30% MVIC = 0.3MVIC. Each contraction lasted about 5 s and rest was allowed for 1 minute between each task to avoid muscle fatigue (Otsuka et al., 2019). The examiner should prevent the subject's pelvic region from rising due to quadriceps contraction when the hip is at 0° (Sung et al., 2019).

The shear elastic modulus in the muscle belly of RF, VI, VL, and VMO were measured under resting and contract conditions using ultrasonic shear wave elastography. Shear modulus was measured by ultrasonic shear wave elastography scanner (AixPlorer, SuperSonic Imagine, Aix-en-Provence, France) with a linear array probe (50 mm, 4–15 MHz, SL15-4, SuperSonic Imagine, Aix-en-Provence, France). The probe for RF and VI was placed at the midpoint of the line connecting the anterior superior iliac spine and the superior pole of the patella (Wang et al., 2017; Kawai et al., 2018).

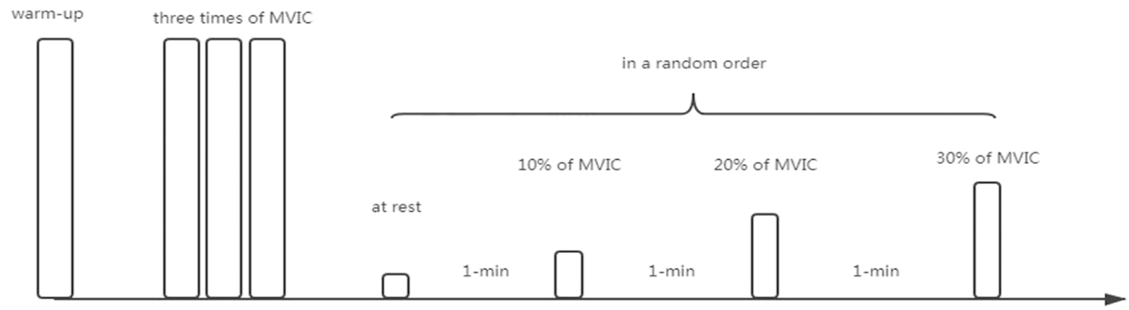

FIGURE 1 | Experimental design. The first is a warm-up to prepare for MVICs, followed by 3 times of MVIC, and the average of the three MVICs is used for subsequent resistance settings, and the knee joint isometric resistance stretching tasks (0, 10, 20%, and 30%MVIC) are performed in random order. Each task lasts 5 s, with 1 min intervals between tasks.

TABLE 1 | The mean shear modulus (kPa) of muscles at resting and contraction states show separately for 90° and 0° of the hip joint angles.

Contraction Level	Hip joint angle	Shear elastic modulus (kPa)			
		RF	VI	VL	VMO
At rest (0%)	90°	17.15 ± 5.25	19.06 ± 6.30	17.81 ± 4.61	14.66 ± 3.70
	0°	24.87 ± 11.40	17.49 ± 3.32	16.06 ± 3.34	16.01 ± 3.76
	p-value	**<0.01***	0.31	0.17	0.13
10%	90°	51.55 ± 25.51	51.15 ± 25.53	65.58 ± 21.27	52.06 ± 19.78
	0°	61.21 ± 21.01	56.38 ± 21.16	62.27 ± 12.15	57.33 ± 23.04
	p-value	0.28	0.38	0.62	0.39
20%	90°	87.14 ± 34.45	75.34 ± 29.55	103.92 ± 20.10	86.31 ± 26.39
	0°	89.15 ± 29.49	87.78 ± 27.08	101.74 ± 20.36	92.76 ± 32.87
	p-value	0.85	0.07	0.72	0.48
30%	90°	120.84 ± 44.88	119.75 ± 44.57	152.12 ± 42.94	127.60 ± 37.08
	0°	121.49 ± 39.96	118.45 ± 31.30	143.91 ± 26.25	129.57 ± 34.90
	p-value	0.96	0.90	0.40	0.79

**Means significant difference.*
Bold value means significant difference as well.
Values are means ± SD.
KPa, kilo Pascal; RF: rectus femoris; VI: vastus intermedius; VL: vastus lateralis; VMO: vastus medialis obliquus.

For VL, the probe was placed at the midpoint between the head of the greater trochanter and the inferior border of the patella (Kawai et al., 2018). For VMO, the probe was placed 4 cm superiorly and 3 cm medially to the patella (Botanlioglu et al., 2013). Young's modulus, quantified in kilopascals (kPa), is color-mapped in a region of interest (ROI) of $15 \times 15\ mm^2$ per muscle fasciculus. A Q-Box™ with a diameter of 10 mm was set inside the ROI and the mean Young's modulus was measured by machine. The images were saved when the colour in Q-Box™ was uniform.

Statistical Analysis

Normality of the data distribution and homogeneity of variances using the Shapiro-Wilk and Levene tests, respectively. The paired *t*-test was used to compare different hip joint angles. The one way-ANOVA or the Welch analysis of variance (Welch ANOVA) with stiffness (same measuring position) as dependent variable and 0% MVIC, 10%MVIC, 20%MVIC, and 30%MVIC as independent variables was used to judge whether there were differences in stiffness between different MVICs. When the one way-ANOVA or the Welch ANOVA was significant, post hoc Tukey's test or Games-Howell test was performed. The statistical significance level for all tests was set at $p < 0.05$ and all measurement data were expressed as mean ± SD (standard deviation).

RESULTS

Effect of Hip Joint Angle

The mean shear modulus values in resting and different contraction states for the RF, VI, VL, and VMO at 0° and 90° of the hip were shown in **Table 1**. For the shear modulus of the RF muscle, the paired *t*-test revealed a significant difference among two hip joint angles at resting states shown in **Figure 2**.

Effect of Contraction Intensity

The stiffness of the quadriceps increases with the strength of the contraction. One-way repeated ANOVA indicated a significant difference between each %MVIC state except that there was no significant difference between 20 and 30% MVIC for RF at hip 90°. **Figure 3** shows the change in shear modulus between

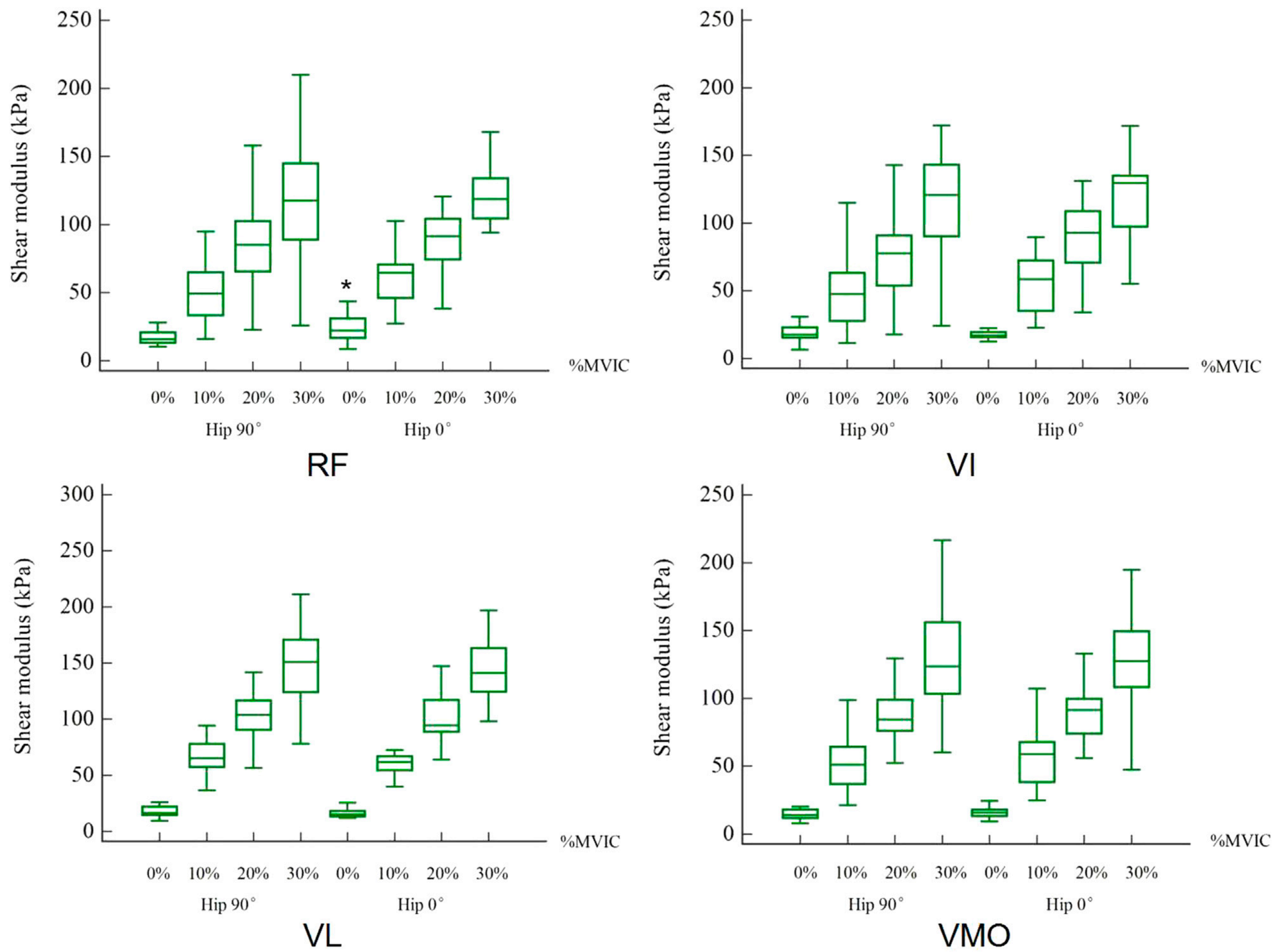

FIGURE 2 | Effects of hip angle on shear modulus of muscles. * means significant difference.

muscles during rest and contraction (SWE plot). **Table 2** shows a comparison of shear modulus between the various contraction states. There were significant differences in comparisons between 0 and 10% MVIC, 10 and 20% MVIC, and 20 and 30% MVIC in the four muscles, except that there was no significant difference between 20 and 30% MVIC for RF (**Figure 4**).

Relationships Between %MVIC and Shear Modulus

Pearson correlation test showed that shear modulus positively correlated with %MVIC. Correlation was a little bigger at 0° of hip for RF (r = 0.80, $p < 0.001$), VI (r = 0.85, $p < 0.001$), VL (r = 0.94, $p < 0.001$) and VMO (r = 0.85, $p < 0.001$) than that of at 90° of hip for RF (r = 0.79, $p < 0.001$), VI (r = 0.78, $p < 0.001$), VL (r = 0.89, $p < 0.001$) except VMO (r = 0.86, $p < 0.001$) (**Table 3**).

DISCUSSION

Among the muscles we choose to measure, only the RF is the muscle that spans both joints, so the change in the angle of the hip joint only affects the RF. In addition, quadriceps shear modulus increased as the muscle contraction level did.

Comparison With Previous Studies Based on Joint Angle

Few studies have explored the relationship between the quadriceps shear modulus and the angle of the hip joint, especially at different isometric contractions. In our research, we found that the shear modulus of RF was significantly different under two different hip joint angles at rest. According to reports, the effect of the knee joint angle is that the stiffness of VI measured at 90° of the knee is always significantly greater than the 60° of the knee at 90° of the hip (Wang et al., 2017). In other words, the shear modulus of the muscle increases as the angle of the knee joint increases. On the contrary, another study on the relationship between the passive range of motion of the ankle joint and tissue stiffness found that the shear wave velocities of the medial gastrocnemius (MG) and lateral gastrocnemius (LG) in young people were negatively correlated with the increase in ankle dorsiflexion angle, which means the shear wave velocity of MG decreased with the increase in ankle dorsiflexion angle

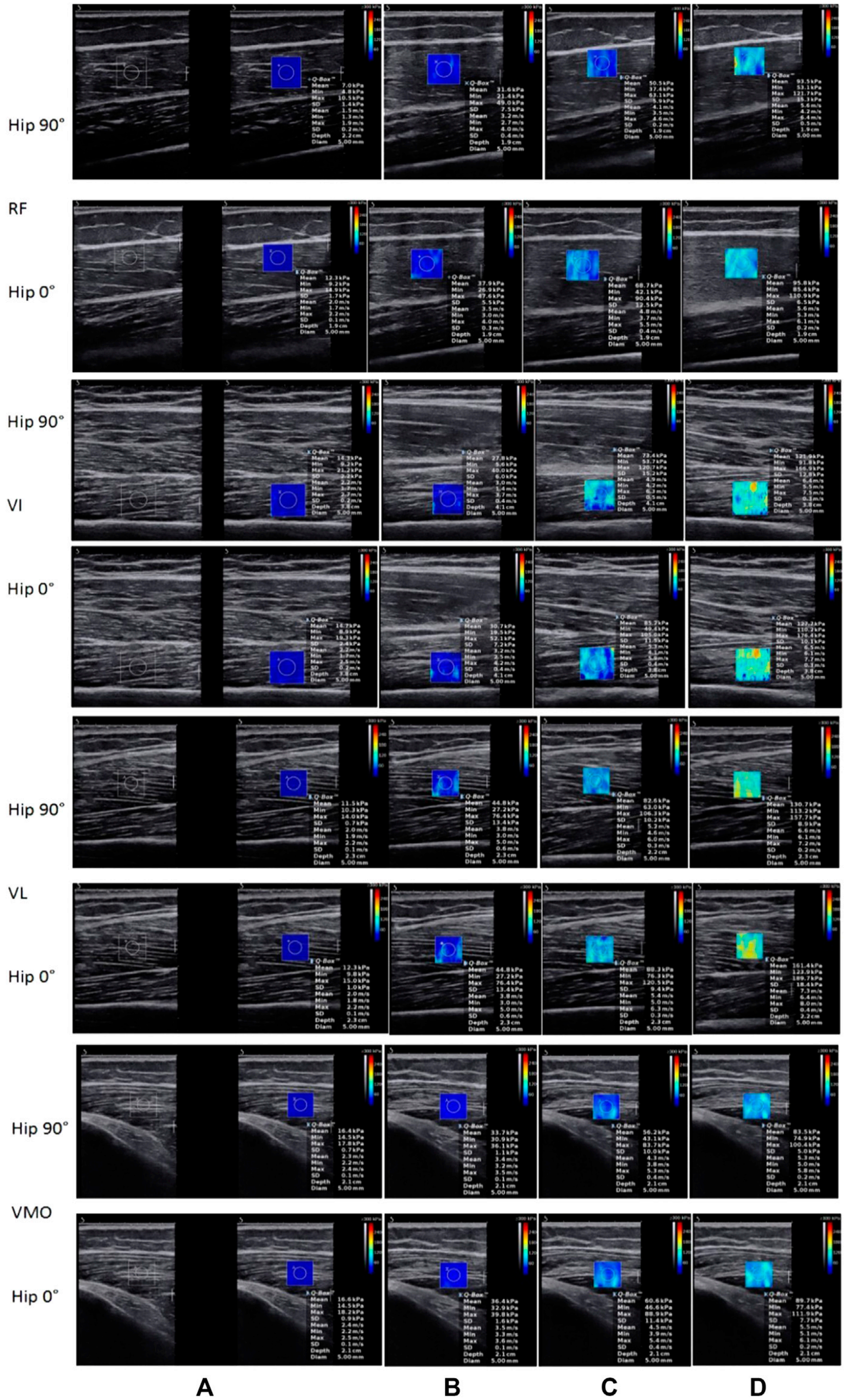

FIGURE 3 | The shear modulus of muscles were shown as resting (0%) and 10, 20, and 30% MVIC from **(A–D)** with blue means softer and red means stiffer. Using SWE, we can see real-time changes in muscle stiffness through real-time color changes.

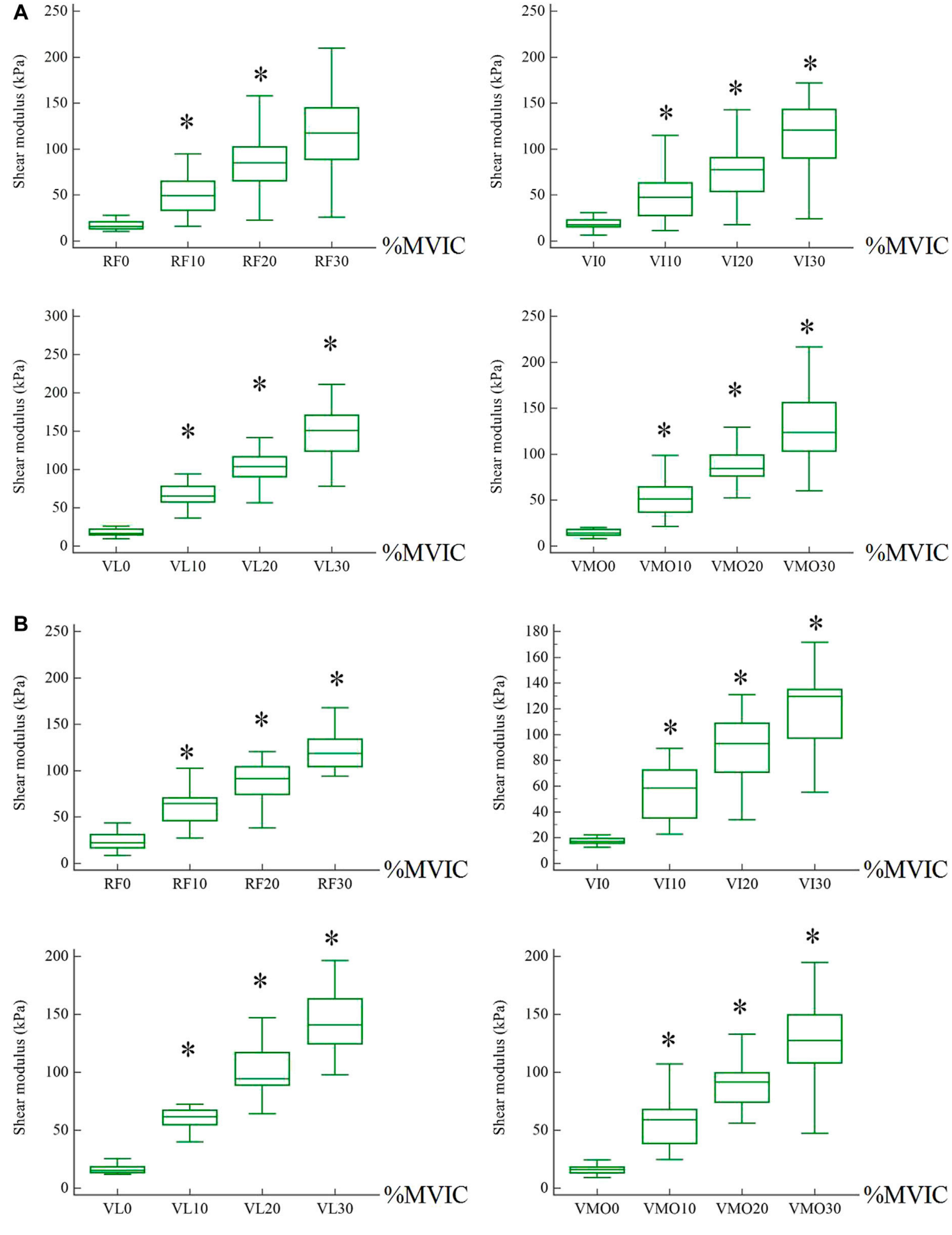

FIGURE 4 | The shear modulus of muscles between different states. * means a significant difference from a smaller contraction state. **(A)** Hip was 90°. **(B)** Hip was 0°.

TABLE 2 | Contraction intensity comparisons.

Muscle	Contraction intensity comparison	Hip 90°	Hip 0°
RF	0–10%MVIC	**<0.001***	**<0.001***
	10–20%MVIC	**0.004***	**0.010***
	20–30%MVIC	0.054	**0.002***
VI	0–10%MVIC	**<0.001***	**<0.001***
	10–20%MVIC	**0.041***	**0.001***
	20–30%MVIC	**0.004***	**0.011***
VL	0–10%MVIC	**<0.001***	**<0.001***
	10–20%MVIC	**<0.001***	**<0.001***
	20–30%MVIC	**0.001***	**<0.001***
VMO	0–10%MVIC	**<0.001***	**<0.001***
	10–20%MVIC	**<0.001***	**0.002***
	20–30%MVIC	**0.001***	**0.008***

**Means significant difference.*
Bold value means significant difference as well.

TABLE 3 | The relationship between quadriceps shear modulus and relative muscle contraction level (% MVIC).

	Hip 90°		Hip 0°	
	r-value	p-value	r-value	p-value
RF	0.79	**<0.001***	0.80	**<0.001***
VI	0.78	**<0.001***	0.85	**<0.001***
VL	0.89	**<0.001***	0.94	**<0.001***
VMO	0.86	**<0.001***	0.85	**<0.001***

RF, rectus femoris; VI, vastus intermedius; VL, vastus lateralis; VMO, vastus medialis obliquus.
**Means significant difference.*
Bold value means significant difference as well.

(Hirata et al., 2020). According to the shear-wave propagation velocity c, the Young's modulus E is derived from $E = 3\rho c^2$, where ρ is the muscle mass density (1,000 kg/m^3), and then quantified in kPa (kilo pascal) units. Since skeletal muscle is not an isotropic material, the shear modulus can be analyzed by dividing the Young's modulus by 3 (Kawai et al., 2018). Therefore, as we have seen, the shear wave velocity is positively related to the shear modulus, or Young's modulus. For hip joint angle, one research discovered that there was no significant difference regardless of the contraction capacity or contraction ratio of VMO and VL between neutral and 30° abduction of the hip position (Botanlioglu et al., 2013). In our study, we measured the shear elastic modulus of RF, VI, VL, and VMO at 90° of the knee and 90° and 0° of the hip, respectively. In the above-mentioned study of changing the abduction angle of the hip joint from the 0° neutral position of the hip joint to the 30° of the hip joint abduction, the shear modulus of VL varied from large to small. This result echoed our findings. At 70° and 90° of the hip and knee joint angle, the shear elastic modulus of VL at rest was 8.8 ± 1.4 kPa (Chalchat et al., 2020). In our study, under the same angle of the knee joint at a rest state, the shear modulus of the VL is 8.8 ± 1.4 kPa at 70° of the hip joint, while the shear modulus of the VL is 17.8 ± 4.6 kPa at 90° of the hip joint. This may be due to the fact that our subjects were younger than the subjects in that study, and the younger age had a larger shear modulus (Wang et al., 2017). Previous research results indicate that during a single joint exercise, the excitation of monoarticular muscles will be affected by the position of the adjacent joint, even if the muscle length does not change (Pui and Jason, 2010), so future single joint exercise training may need to consider the potential impact of the adjacent joint position.

Comparison With Previous Studies Based on Contraction Intensity

In our current study, we explored the shear modulus of RF, VI, VL, and VMO of the quadriceps muscle at 0, 10, 20, and 30% MVIC. Under eleven step levels of isometric contraction, the stiffness of VI was measured in the entire range of 0–100% MVIC (Wang et al., 2017). The shear modulus of VI at rest was 16.2 ± 8.1 kPa in their study, while in the present study it was 19.1 ± 6.3 kPa. However, they found that the shear modulus of the elderly subjects was smaller than that of the young subjects, so our results agreed well with theirs. This can be explained by the effect of age on muscle strength (Domínguez-Navarro et al., 2020), because the age of their participants (28.5 ± 4.9 years) is higher than the age of our participants (20.8 ± 2.0 years). Decreased weakness of quadriceps muscles with increasing age is a risk factor for falling (Domínguez-Navarro et al., 2020). Patients with quadriceps dysfunction had reduced muscle stiffness during contraction (Kawai et al., 2018). Otsuka et al. studied the shear wave velocities of RF and VL at 0, 20, 40 and 60% of MVIC. Converting the shear wave velocity in their study to kPa, in units consistent with our study, corresponds to 17.28 ± 0.27 kPa for RF at 0%, 81.12 ± 3 kPa for 20% and VL at 0% is 20.28 ± 0.27 kPa, and 20% is 126.75 ± 3 kPa (Otsuka et al., 2019). The different posture and knee joint angle they used were difficult to use to directly compare the stiffness values they measured with ours. However, our conclusion that there was a significant increase in the shear modulus of the four muscle with the increase in the isometric contraction level was in good agreement with their finding. The stiffness of the VI of 0% MVIC measured at 90° of the knee and 90° of the hip in young female was 19.0 ± 10.2 kPa (Wang et al., 2017), which was similar to our study the that stiffness of the VI for 0%MVIC was 19.06 ± 6.30 kPa.

The Relationship Between Quadriceps Shear Modulus and Relative Muscle Contraction Level (% MVIC)

Our results showed that, the stiffness of RF, VI, VL, and VMO muscle bellies along the direction of muscle action increased with increasing contraction levels and were significantly and moderate positively correlation with the relative contraction level (% MVIC) which close to a linear relationship. The stiffness of skeletal muscle was positively correlate to non-fatigue contraction intensity, that is, no more than 60% MVIC (Wang et al., 2017). It was found that most reports show a linear relationship when measuring a smaller range of 3-5 isometric contraction levels (Wang et al., 2017), which is consistent with ours. With the increasing contraction levels, the shear modulus of RF, VL, and VI were increased (Wang et al., 2017; Otsuka et al., 2019). As **Table 3** shown, the correlation coefficients of RF and VL muscle at 90° of hip were 0.79 and 0.89, respectively. The correlation coefficients of RF ($r = 0.72$) and VL ($r = 0.88$) when the probe placed longitudinally was similar to our research results (Otsuka et al., 2019). The correlation coefficient of VL is the largest regardless of the hip angle, since the VL muscle has the largest cross-sectional area in the quadriceps, which is important for force generation during knee extension. The correlation coefficient at 0° of the hip joint is greater than the correlation coefficient at 90°, so it is recommended to perform quadriceps contraction exercises at 0° of the hip.

Limitations

Any neuromuscular activity generated during rest and contraction was evaluated and recorded as surface electromyography (EMG) signals (Wang et al., 2017; Chalchat et al., 2020). However, EMG cannot quantify the mechanical properties of muscles. One of the limitations of the present study was that we did not use EMG while we paid more attention to the mechanical properties of skeletal muscle, especially elasticity, rather than neuromuscular features and morphological characteristics. We preferred to use SWE as a non-invasive tool to estimate the degree of myoelectric activity instead of EMG as a invasive technique. The VI is an important knee extensor, but its depth in the inner thigh does not allow for surface EMG measurements. Other study that did not use EMG was in agreement with ours (Botanlioglu et al., 2013). Furthermore, the vastus medialis longus (VML) contributed to knee extension, whereas evaluations of shear elastic modulus of quadriceps in the current study did not take it into account because we focused on the patella's line of action, which influences stress distribution on the patellofemoral joint. When stress on the patellofemoral and knee joints changes, the VMO that is attached to the aponeuroses would be affected, and therefore, we could detect the altered stress by evaluating the elastic properties of the VMO using SWE (Castanov et al., 2019). Since there was no statistically significant difference between the right and left extremities in healthy control regardless of females or males (Botanlioglu et al., 2013), our study was coincident with previous studies that all measurements were taken from the dominant leg (right leg for all participants) (Otsuka et al., 2019; Kawai et al., 2018; Chalchat et al., 2020).

CONCLUSION

The effect of hip joint angle was only observed in the RF muscle. Our research shows that the correlation coefficient between shear elastic modulus and %MVIC were greater at 0° of the hip. Quantitative research on quadriceps muscle elastic properties may be of great significance in rehabilitation medicine for better understanding the way of quadriceps muscle recruitment in resting and isometric contraction phases.

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 authors.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethics Committee of Guangdong Provincial Hospital of Chinese Medicine. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

WD designed the study and drafted the manuscript; CL helped to conceive the study; CT helped to perform statistical analysis; ML, SY, and HL participated in the data collection; ZZ revised the article critically for important intellectual content. All authors read and approved the final manuscript.

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The experiments comply with the current laws of the country in which they were performed.

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Prospective Study on Dynamic Postural Stability in Youth Competitive Alpine Skiers: Test-Retest Reliability and Reference Values as a Function of Sex, Age and Biological Maturation

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This study aimed 1) to assess the test-retest reliability of dynamic postural stability index (DPSI) assessments using a ski-specific jump protocol that consists of single-leg landings on a three-dimensional force plate after forward-performed double-leg drop jumps from a box over a hurdle (DJSLLs), 2) to provide reference values for female and male youth competitive alpine skiers; 3) to explore their changes in DPSI over 3 years during adolescence; and 4) to investigate potential associations of DPSI with age and biological maturation. Using three-dimensional force plates, 16 healthy subjects were tested on the same day (test-retest reliability experiment; five test-retest assessments of right leg landings), and 76 youth skiers aged 13–15 years were tested 3 times within 2 years (main experiment; average of two trials per leg each time). The test-retest reliability experiment revealed an ICC(3,1) and 95% CI of 0.86 [0.74, 0.94] for absolute DPSI assessment. The within-subject SEM of absolute DPSI was 16.30 N [13.66 N, 20.65 N], and the standardized typical error was moderate (0.39 [0.33, 0.50]). Both absolute and relative DPSI values were comparable between male and female youth competitive alpine skiers. The mean absolute DPSI in year 1 (195.7 ± 40.9 N), year 2 (196.5 ± 38.9 N) and year 3 (211.5 ± 41.3 N) continuously increased (i.e., worsened) ($p < 0.001$). Mean relative, i.e. body weight force normalized, DPSI values significantly decreased, i.e., improved, from year 1 to 2 (0.42 ± 0.01 vs. 0.36 ± 0.004; $p < 0.001$) and year 1 to 3 (0.42 ± 0.01 vs. 0.36 ± 0.01; $p < 0.001$). Absolute DPSI correlated with age and biological maturation, while no such correlations were found for relative DPSI values. Our findings suggest that DPSI is a reliable and sensitive measure of dynamic postural control during DJSLLs and that relative DPSI improves annually in competitive youth skiers when accounting for body weight. Future work should consider biological maturation testing during the growth spurt, and normalizing to body weight force could be a possible solution.

Keywords: athletes, physical fitness, postural balance, athletic performance, injury prevention, alpine skiing

INTRODUCTION

Competitive alpine skiing is a sport with a relatively high risk of injury (Florenes et al., 2009; Bere et al., 2014b; Haaland et al., 2016; Alhammoud et al., 2020; Fröhlich et al., 2020), even at the youth level (Schoeb et al., 2020). The most frequent health problems among youth and elite skiers are knee injuries, with the anterior cruciate ligament (ACL) being most commonly ruptured (Florenes et al., 2012; Raschner et al., 2012; Westin et al., 2012; Hildebrandt and Raschner, 2013; Müller et al., 2017; Schoeb et al., 2020). The prevention of such severe injuries should therefore play a key role in the training of competitive alpine skiers. The mechanisms of ACL injuries in alpine skiing have been studied over the last few years, with three injury mechanisms described: "slip-catch", "landing back-weighted" and "dynamic snowplow" mechanisms (Bere et al., 2011a; Bere et al., 2013; Jordan et al., 2017; Spörri et al., 2017). These situations mainly occur during turns and jump landings and are in most cases the result of out-of-balance situations (Bere et al., 2011a; Bere et al., 2011b; Bere et al., 2014a).

With respect to the mediolateral direction, a skier gets out-of-balance if the force vector resulting from gravitational force and centrifugal force does not direct through the area of support that is spanned by the two skis and the standing width (Howe, 2001; Reid et al., 2020). To maintain balance in the anteroposterior direction, skiers need to react to disturbances due to rapidly changing ski-snow interactions and air resistance, as well as perturbations caused by uneven snow surfaces and gate contacts while skiing in a certain direction, and to bring the body back into dynamic equilibrium as quickly as possible (LeMaster, 2009; Gadient et al., 2010). In the vertical direction, skiers need to compensate for the accelerations and cushion the impacts resulting from convex terrain transitions (compressions) or landing from jumps (Heinrich et al., 2014; Gilgien et al., 2015).

Such skiing-specific balance patterns require excellent neuromuscular control in the lower extremities and trunk to achieve sufficient dynamic postural control under the various challenging and often unpredictable constantly changing conditions of alpine skiing. Youth skiers are especially worthy of investigation, since they are known to have fluctuating neuromuscular control performance during the growth spurt (Quatman-Yates et al., 2012). Because there is no direct equipment connection between the left and right leg as in snowboarding, for instance, the aforementioned skiing-specific dynamic postural control tasks need to be performed unilaterally and independently on each leg. As such, there are certain similarities with the motion task of a single-leg landing on a force plate after a forward-performed double-leg drop jump from a box over a hurdle, hereinafter called drop jump single leg landing (DJSLL), which challenges the dynamic equilibrium in all three directions (**Figure 1**).

A suitable measure to quantify the balance performance during DJSLL could be found in the Dynamic Postural Stability Index (DPSI), which has already been described several times in previous studies (Goldie et al., 1989; Wikstrom et al., 2005; Hellmers et al., 2017). The DPSI is a measure that determines a person's ability to regain balance while transitioning from a dynamic to a static state of single-leg stabilization. Over a 3-s period after impact, the fluctuation of ground reaction force (GRF) around the origin during landing is determined, thereby quantifying neuromuscular control in the anteroposterior, mediolateral, and vertical directions (Wikstrom et al., 2005). Previous studies assessed dynamic postural control primarily during basic stance positions, counter movement jumps or single leg landings after two-leg vertical jumps (Goldie et al., 1989; Wikstrom et al., 2005; Hellmers et al., 2017) and reported the latter test to be highly reliable (intraclass correlation coefficient = 0.96) and very precise (SEM = 0.03) (Wikstrom et al., 2005).

Despite the aforementioned research related to DPSI in athletes, to date, no study has analysed DPSI in youth competitive alpine skiers and/or using the slightly modified, forward-performed DJSLL jumping task. Overall, such a task might better reflect the high demands on skiers' unilateral and leg-independent dynamic postural control mechanisms, in which their centre of mass moves steadily forward while their skis repeatedly and leg-independently change frictional resistance as they glide down the slope (Lind and Sanders, 2004).

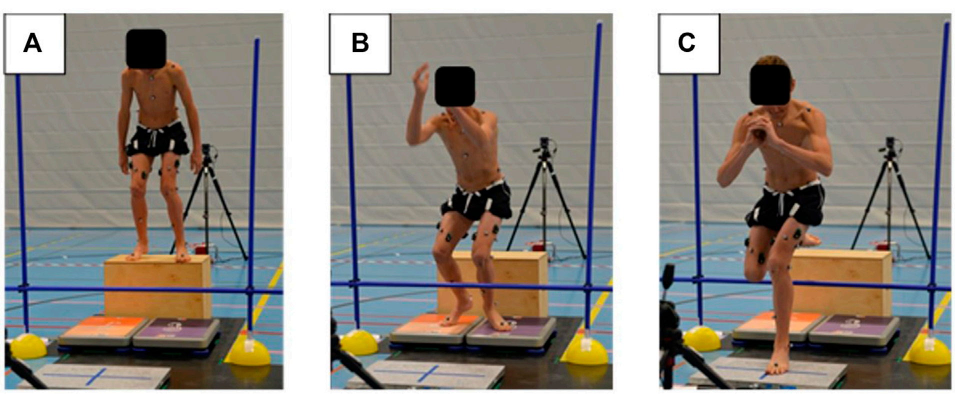

FIGURE 1 | Youth competitive alpine skier performing a DJSLL **(A)** starting on the box **(B)** dropping off the box and landing on both feet and **(C)** jumping over the hurdle with a single leg landing on the force plate in front.

TABLE 1 | Baseline characteristics of the youth competitive alpine skiers at the beginning of the study.

	Overall (n = 76)	Females (n = 30)	Males (n = 46)
Age [y]	13.7 ± 0.6 (12.5–14.9)	13.6 ± 0.7 (12.5–14.9)	13.8 ± 0.5 (12.9–14.8)
Body height [cm]	161.0 ± 8.0 (143.0–185.0)	160.3 ± 6.4 (143.0–171.5)	161.4 ± 8.9 (146.0–185.0)
Body weight [kg]	48.5 ± 10.1 (30.0–81.0)	46.8 ± 7.6 (35.0–66.0)	49.7 ± 11.4 (30.0–81.0)
BMI [kg/m^2]	18.5 ± 2.4 (13.0–24.7)	18.1 ± 1.9 (14.4–23.4)	18.8 ± 2.6 (13.0–24.7)
Maturity offset [y]	0.2 ± 1.2 (-2.0–2.8)	1.3 ± 0.7 (0.1–2.8)	-0.5 ± 0.8 (-2.0–1.6)
APHV [y]	13.5 ± 1.1 (11.3–15.4)	12.4 ± 0.5 (11.3–13.4)	14.3 ± 0.6 (12.8–15.4)

Data are expressed as the mean ± SD (minimum-maximum). BMI, body mass index; APH, age at peak height velocity.

However, the test-retest reliability of such an alternative test protocol is currently unknown.

Therefore, the aims of the current study were 1) to study the test-retest reliability of DPSI assessments using a ski-specific jump protocol that consists of single-leg landings on a three-dimensional force plate after forward-performed double-leg drop jumps from a box over a hurdle (DJSLLs); 2) to provide reference values for female and male youth competitive alpine skiers; 3) to explore their changes in DPSI over 3 years during adolescence; and 4) to investigate potential associations of DPSI with age and biological maturation.

MATERIALS AND METHODS

Study Design, Setting and Participants

Test-Retest Reliability Experiment

To verify the test-retest reliability of DPSI assessments during DJSLL, a cross-sectional study was set up with 16 young, healthy adults (9 females and seven males) who consecutively completed a DJSLL test 5 times on the same day. All instructions and the underlying measurement setup were identical to the “main experiment”. The only exception was the number of repetitions and leg-related unilaterality of the data underlying the reliability calculations (see below). Recruitment of subjects took place via public tender ahead of the experiment. The inclusion criteria were 18–40 years of age and being physically active (i.e., at least 30 min of moderate activity per day or one-time intense physical activity per week). Exclusion criteria were BMI greater than 45 or inability to perform DJSLLs without restrictions due to pain or injury. Participants’ detailed characteristics are described in the corresponding paragraph of the results section.

Main Experiment

Out of the potential pool of approximately 220 youth competitive alpine skiers of the U16 category in Switzerland, 76 skiers annually performed the DPSI tests in three consecutive years. In contrast to the test-retest reliability experiment, a total of four trials (2 left leg landings and two right leg landings) were performed and used for further analysis. Skiers were eligible for study participation if they were members of a regional performance centre (RLZ) certified by Swiss-Ski. In Switzerland, children typically start skiing at age 3, participate in their first fun competitions at age 6, and enter a semiprofessional youth development programme with sport-specific training plans at age 12. Each year, tests were conducted before the start of the youth alpine competition season in November. None of the participants were excluded from the study due to predefined exclusion criteria (i.e., health problems that limit current load tolerance, being in the return-to-sport process, or systemic diseases such as inflammatory arthritis or diabetes mellitus). The detailed characteristics of the participants are further described in **Table 1** in the results section.

Ethical Approval

Participation was voluntary, and all participants signed an informed consent form before taking part in this study. For participants under 14 years of age, legal guardians signed the forms. The current study was approved by an institutional review board and the local ethics committee (KEK-ZH-NR: 2017-01395). All procedures were in accordance with the Declaration of Helsinki and national laws.

Data Collection and Evaluation

Baseline Data and Determination of Biological Maturity

All participants of the test-retest reliability experiment underwent a baseline assessment of age, body mass (0.1 kg increments, Seca, Hamburg, Germany), and body height (0.5 cm, determined by measuring tape). In the youth skiers of the main experiment, in addition to the abovementioned baseline measures, sitting height (0.5 cm increments, determined with a tape measure) was measured during each of the three test sessions. Maturity offset was obtained using the noninvasive, anthropometric method proposed by Mirwald et al. (2002) that predicts the age at peak height velocity (APHV) and which had been validated in athletes, as well as in youth competitive alpine skiers previously (Malina et al., 2007; Müller et al., 2015). To calculate maturity offset, sex-specific formulas were used, building upon the following input data: chronological age, body weight, body height, sitting height, and leg length (body height–sitting height). Maturity offset thereby represents a point in time before or after the age at peak height velocity (APHV).

Dynamic Postural Stability Index

Biomechanical data were obtained using a three-dimensional (3D) force plate (Kistler Group, Winterthur, Switzerland) to track ground reaction forces during landing after the DJSLL. The 3D force plate data were collected at a sampling rate of 2000 Hz during the landing phase of 5 seconds. All participants underwent a series of trials, all within one test session on the same day. Participants were instructed to start on a 32 cm high box in an upright position (**Figure 1A**) and drop themselves from the box. The drop was followed by a double-legged landing (**Figure 1B**) followed by an immediate forward jump over a 37 cm high hurdle with a one-leg landing in the middle of the 3D force plate (**Figure 1C**). Participants were allowed to move their hands freely in the jump phase. After the hurdle jump, participants should maintain a rather upright position with their hands together in front of the body and stabilize for 5 seconds. Only the first 3 seconds were used for further analysis, as it was found to be the most meaningful period to assess DPSI (Wikstrom et al., 2005). The trial was invalid if participants 1) initiated the drop from the box by actively jumping off, 2) the force plates were not correctly hit during landing, 3) hesitated upon landing before the hurdle jump or 4) stabilized for less than 5 s after the second landing, lost balance or made a second hop. If the jump was invalid, the trial was repeated until a total of two valid trials for each leg were recorded. Between the trials, there was a recovery time of at least 10–15 s.

Raw data from all analogue channels of the force plate were transferred to MATLAB (MATLAB R2016b, The MathWorks, Inc.) to calculate the DPSI values with customized MATLAB scripts. The ground reaction force data were filtered using a second order Butterworth low pass filter with a cut-off frequency of 200 Hz. All trial data were reduced to a 3 s interval of the landing phase after the hurdle jump, beginning at the time of initial ground contact, which was defined as the point in time where the vertical GRF crossed a threshold of 25 N. Absolute DPSI was then calculated over the corresponding interval as a composite of force in all three directions, which are mean square deviations assessing fluctuations around a zero point, with the formula according to Hellmers et al. (2017):

$$Absolute\ DPSI = \sqrt{\frac{\sum (0 - F_x)^2 + \sum \left(0 - F_y\right)^2 + \sum (F_{BW} - F_z)^2}{n}}$$

n represents the number of data points (i.e., 2000 Hz × 3 s = 6000 frames in our study), F_{BW} is the body weight force, and F_x, F_y, F_z are the forces in the anteroposterior (x), mediolateral (y) and vertical (z) directions. The relative DPSI was determined by dividing the absolute DPSI by body weight force.

Statistical Analysis

Test-Retest Reliability Experiment

The assessment of the absolute DPSI test-retest reliability was based on five trials consecutively performed on the same day, and only data from the right leg were used. The Shapiro–Wilk test, graphical techniques (i.e., histograms and quantile-quantile plots) and shape parameters (i.e., skewness and kurtosis coefficients) were used to assess the normal distribution of the data. The dependent absolute DPSI differences between the five test repetitions of the 16 young, healthy adults were tested for significance using a repeated-measures ANOVA ($p < 0.05$). These analysis steps were performed in IBM SPSS software (Version 26). For the calculation of the within-subject raw standard error of measurement (SEM, also called "typical error"), within-subject standardized SEM (namely, "standardized typical error"), and the interclass correlation coefficient ICC(3,1) of absolute DPSI assessments during DJSLL, the consecutive pairwise spreadsheet of Hopkins (2015) and the values of five repeated DPSI tests were used. This spreadsheet allows an appropriate test-retest reliability assessment of performance tests where habituation is an issue. SEM was calculated as the degrees of freedom-weighted average of the typical errors (i.e., the SD of the score differences between two adjacent consecutive trials divided by √2) across the five repeated trials assessed (Hopkins, 2015). With respect to the ICC(3,1) calculation, "3" refers to the type of ICC in which the subjects are a random effect and the trials are a fixed effect, while "1" refers to the reliability of single repeated measurements (not the mean of several measurements) (Hopkins, 2015). For the classification of the standardized typical errors, values were doubled before interpreting their magnitude in relation to the common thresholds of 0.2, 0.6, 1.2, 2.0, and 4.0 for the labels *small*, *moderate*, *large*, *very large*, and *extremely large*, as proposed by Smith and Hopkins (2011). The interpretation of the ICC values was based on the classification of Koo and Li (2016) (<0.5, *poor* test-retest reliability; 0.5-0.75, *moderate* test-retest reliability; 0.75–0.9 *good* test-retest reliability; and >0.9 *excellent* test-retest reliability).

Main Experiment

The absolute and relative DPSI values of a youth competitive alpine skier, on which further analysis of the main experiment was based, were derived by averaging his/her four performed trials (i.e., two trials with landing on the left leg and two trials with landing on the right leg) into representative mean values. Normal distribution of the data was checked using the Shapiro–Wilk test, graphical techniques (i.e., histograms and quantile-quantile plots) and shape parameters (i.e., skewness and kurtosis coefficients). All accompanying baseline and biological maturity data were evaluated with respect to sex (female vs. male) and time (year 1 vs. year 2 vs. year 3) differences by the use of a multivariate analysis of variance (MANOVA) for repeated measures at $p < 0.05$, with Bonferroni correction for pairwise comparisons. Sex (female vs. male) and time (year 1 vs. year 2 vs. year 3) differences in absolute and relative DPSI values were analysed by two-way repeated-measures analyses of variance (ANOVAs) at $p < 0.05$; again, with Bonferroni correction. Pairwise comparisons were additionally illustrated by mean and 95% confidence interval (CI) plots. Correlations of absolute and relative (i.e., body weight force normalized) DPSI with age and biological maturation were tested using Pearson's correlation coefficients (r) and coefficients of determination (R^2). The level of significance was set at $p < 0.05$. IBM SPSS software (Version 26) was used for statistical analysis.

RESULTS

Test-Retest Reliability

On a group level, the repeated-measures ANOVA revealed no significant differences in absolute DPSI between the five repetitions at $p < 0.05$ (235.6 ± 40.2 N, 237.5 ± 41.9 N, 232.2 ± 38.3 N, 235.8 ± 49.4 N, 227.8 ± 36.1 N). The test-retest reliability of the absolute

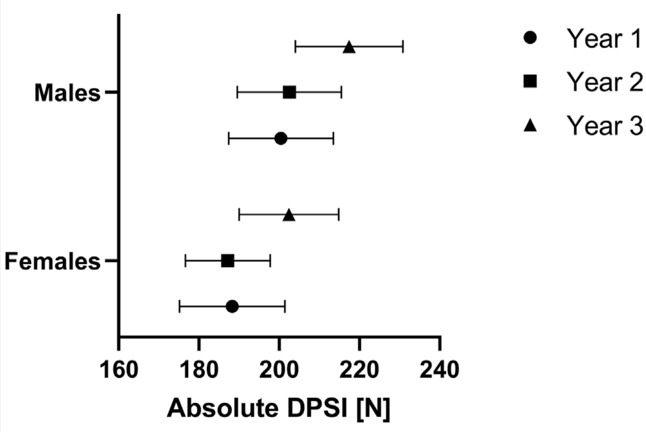

FIGURE 2 | Absolute dynamic postural stability index (DPSI) of youth competitive alpine skiers over 3 years separated by sex. Data are expressed as the mean ±95% CI and represent the sex-specific reference values for the U14, U15 and U16 categories [year 1: females 188.3 N (175.2 N, 201.5 N), males 200.5 N (187.4 N, 213.5 N); year 2: females 187.3 N (176.7 N, 197.8 N), males 202.6 N (189.6 N, 215.6 N); and year 3: females 202.4 N (190.0 N, 214.8 N), males 217.4 N (204.0 N, 230.8 N)].

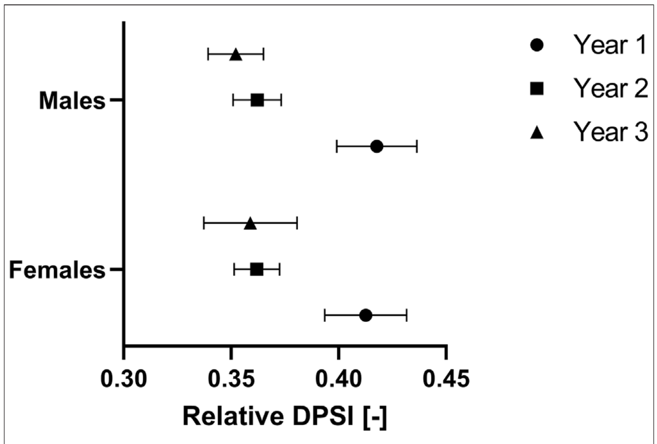

FIGURE 3 | Relative dynamic postural stability index (DPSI) of youth competitive alpine skiers over 3 years separated by sex. Data are expressed as the mean ±95% CI and represent the sex-specific reference values for the U14, U15 and U16 categories [year 1: females 0.41 (0.39, 0.43), males 0.42 (0.40, 0.44); year 2: females 0.36 (0.35, 0.37), males 0.36 (0.35, 0.37); and year 3: females 0.36 (0.34, 0.38), males 0.35 (0.34, 0.36)].

DPSI assessment during DJSLL was found to be *good* [ICC(3,1) was 0.86 with a 95% CI of (0.74, 0.94)]. The within-subject SEM was 16.30 N [13.66 N, 20.65 N], and the standardized typical error was found to be *moderate* [0.39 (0.33, 0.50)].

Baseline Data and Biological Maturity

The baseline characteristics of the young, healthy adults participating in the test-retest reliability experiment were as follows: age 28.9 ± 4.7 years, body height 171.7 ± 8.4 cm, body weight 68.0 ± 10.9 kg, and BMI 22.9 ± 2.0 kg/m^2. Baseline characteristics for the skiers that participated in the main experiment can be found in **Table 1**. A repeated-measures ANOVA showed significant differences on a multivariate level between male and female skiers in the main experiment ($p <$ 0.001; partial eta^2 = 0.962) and between the years assessed ($p <$ 0.001; partial eta^2 = 1.000), and there was an interaction effect for year * sex ($p < 0.001$, partial eta^2 = 0.485). On a univariate level, body height ($p = 0.016$; partial eta^2 = 0.077), maturity offset ($p <$ 0.001; partial eta^2 = 0.517) and APHV ($p < 0.001$; partial eta^2 = 0.693) were significantly different between males and females. Over the years, age ($p < 0.001$; partial eta^2 = 1.000), body height ($p < 0.001$; partial eta^2 = 0.748), body weight ($p < 0.001$; partial eta^2 = 0.851), BMI ($p < 0.001$; partial eta^2 = 0.718), maturity offset ($p < 0.001$; partial eta^2 = 0.922) and APHV ($p < 0.001$; partial eta^2 = 0.266) were significantly different on a univariate level.

Absolute DPSI

A repeated-measures ANOVA (sphericity not assumed: Mauchly W (2) = 0.921, $p = 0.049$) showed that absolute DPSI was significantly different over the years ($p < 0.001$, partial eta^2 = 0.202). Bonferroni corrected pairwise comparisons showed that absolute DPSI was significantly smaller in year 2 (196.5 ± 38.9 N) compared to year 3 (211.5 ± 41.3 N) and in year 1 (195.7 ± 40.9 N) compared to year 3 (211.5 ± 41.3 N), both at $p < 0.001$ (**Figure 2**). In contrast, absolute DPSI was not significantly different between years 1 and 2 (p = 1.000). No significant differences in absolute DPSI values between males and females were observed ($p = 0.112$, partial eta^2 = 0.034).

Relative DPSI

A repeated-measures ANOVA (sphericity not assumed: Mauchly-W (2) = 0.867, $p = 0.005$) shows that the relative DPSI is significantly different over the years ($p < 0.001$, partial eta^2 = 0.451). Bonferroni corrected pairwise comparisons showed that the relative DPSI was significantly lower in year 2 (0.36 ± 0.03) than in year 1 (0.42 ± 0.06) and in year 3 (0.35 ± 0.05) than in year 1 (0.42 ± 0.06), both at $p < 0.001$ (**Figure 3**). In contrast, the relative DPSI was not significantly different between years two and 3 ($p = 0.183$). There were no significant differences in the relative DPSI values between males and females ($p = 0.949$).

Association Between Absolute/Relative DPSI, Age and Maturity Offset

In **Figure 4**, correlations of absolute/relative DPSI and biological maturity are presented, and correlations of absolute/relative DPSI and age are presented in **Figure 5**. While for the absolute DPSI values of both sexes over all 3 years, there were significant correlations with biological maturation ($p < 0.01$) and age ($p <$ 0.001), such relations were not present for relative, i.e., body weight force normalized, DPSI values.

DISCUSSION

The purpose of this study was to provide a prospective observation of dynamic postural control ability in 76 youth competitive alpine skiers. The main findings were that 1) the test-retest reliability of absolute DPSI assessments during a jump landing exercise may be considered *good*; 2) both absolute and

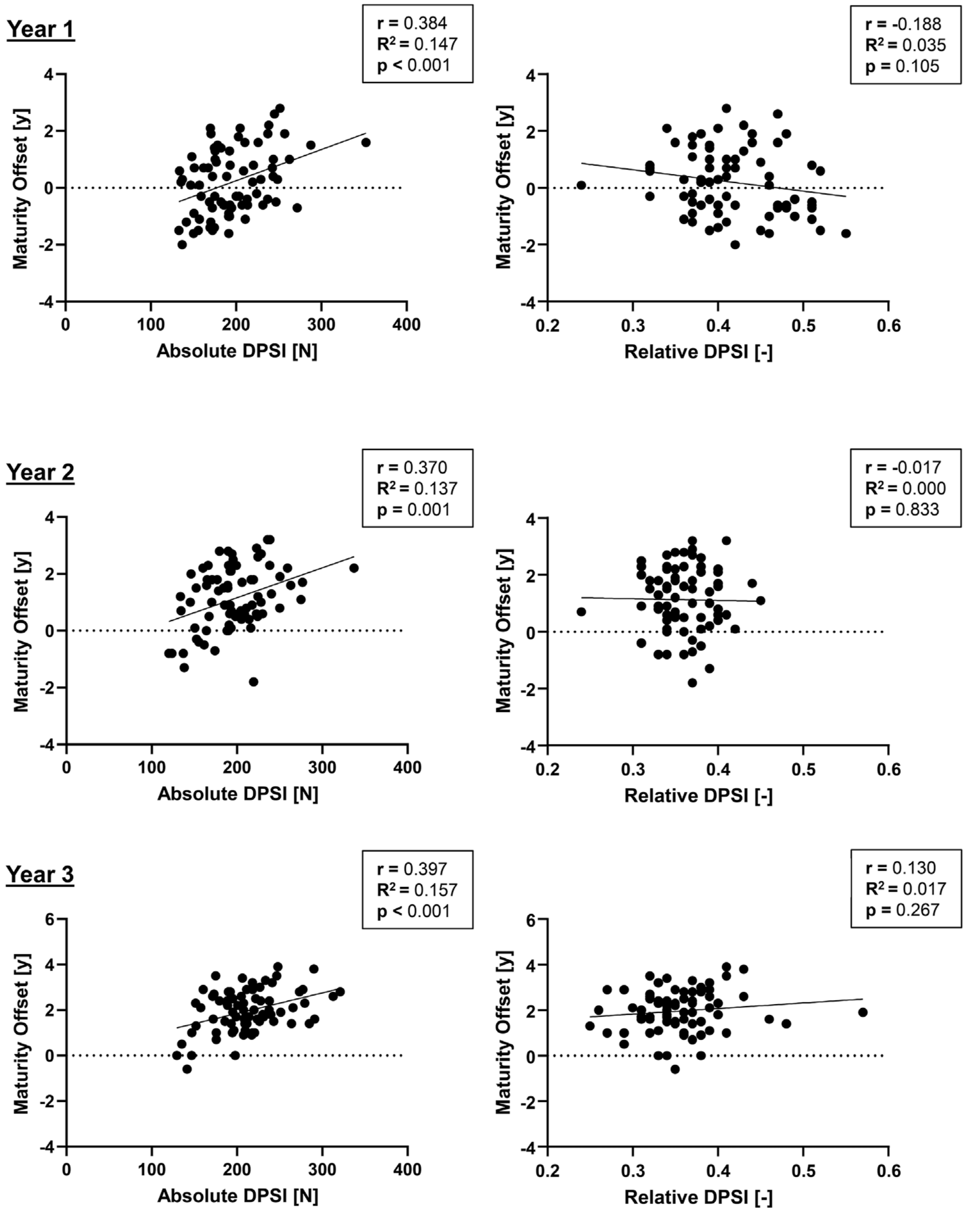

FIGURE 4 | Correlations of absolute and relative dynamic postural stability index (DPSI) with maturity offset in youth competitive alpine skiers.

relative DPSI values were comparable between male and female youth competitive alpine skiers 3) absolute DPSI values increased (i.e., worsened); over 3 years, and relative DPSI values decreased (i.e., improved) when normalized for body weight force; and 4) absolute DPSI correlated with age and biological maturation, while no such correlations were found for relative DPSI values.

Test-Retest Reliability of Determining DPSI During DJSLL Tasks

The reliability experiment yielded an ICC(3,1) and 95% CI of 0.86 [0.74-0.94] for determining absolute DPSI during DJSLL tasks, which may be considered *good*. In comparison, an earlier study by Wikstrom et al. (2005) reported a slightly better ICC(3,1) of 0.96 [0.91–0.98] for relative DPSI assessments. However, it must also be stressed that the jumping task (i.e., a two-legged jump to a height equal to 50% of the maximum vertical jump and landing on one leg) was probably less challenging and less specific to the demands of alpine skiers than the task included in the current study (including a hurdle jump forward before landing on one leg).

Sex Differences in Absolute and Relative DPSI Values During DJSLL Tasks

Comparable DPSI values were found for males and females. One may have expected sex differences in DPSI, since males have

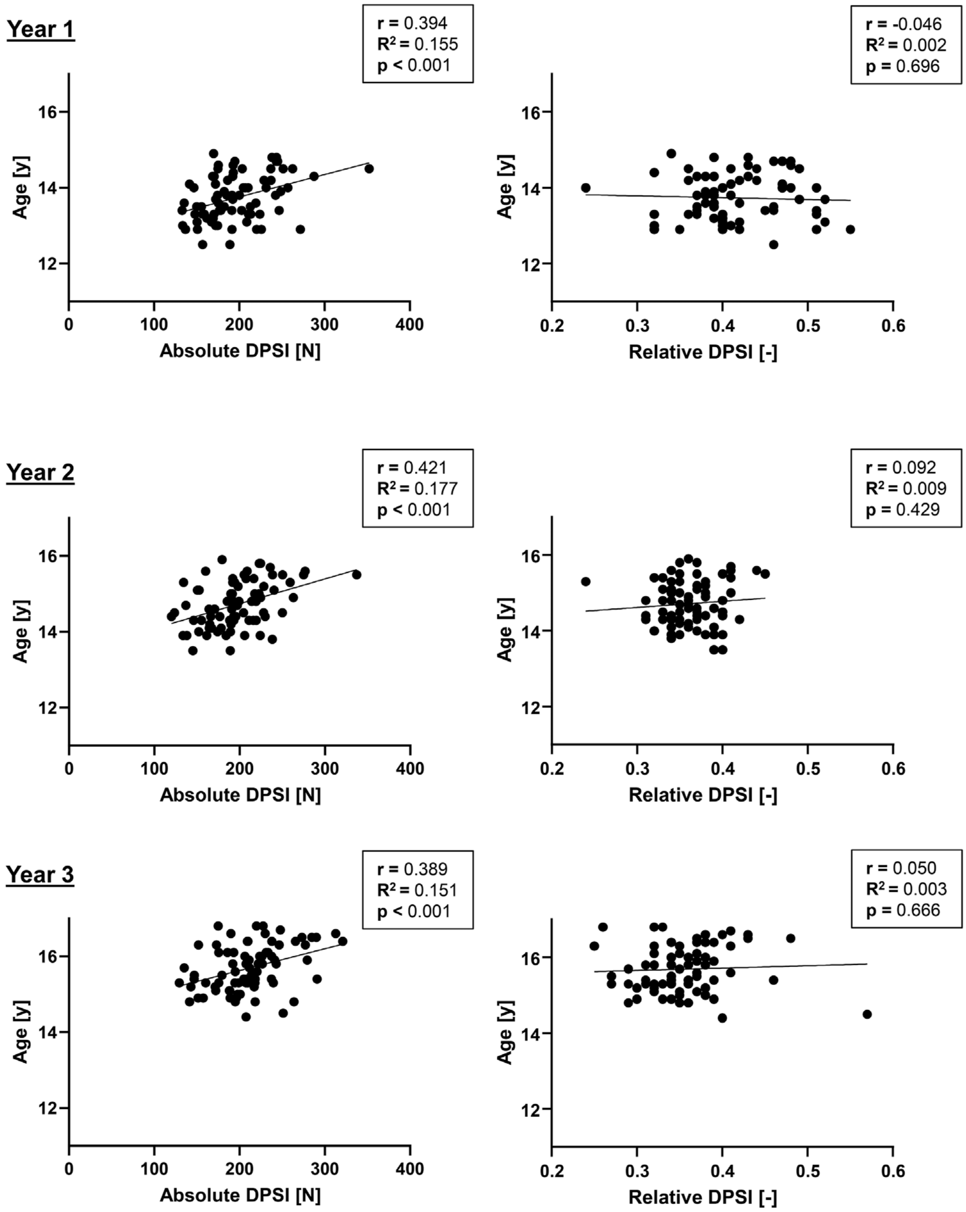

FIGURE 5 | Correlations of absolute and relative dynamic postural stability index (DPSI) with age in youth competitive alpine skiers.

greater neuromuscular control in general (Hewett et al., 2005; Hewett et al., 2006). During the growth spurt, both males and females gain length and body mass, but sex differences occur around the growth spurt. Puberty hormones affect males and females differently along with the important physical difference that muscle mass and strength increase more prominently in males than in females (Philippaerts et al., 2006). Given this difference in muscle mass, neuromuscular control may differ between males and females, with greater neuromuscular control in males. At the start of this study, female skiers had already passed the growth spurt (1.2 ± 0.7), in contrast to male skiers (-0.5 ± 0.8). The DPSI differences between males and females might therefore be compensated by biological maturity with a lagging neuromuscular control in the male skiers.

Absolute and Relative DPSI Values Over the three Consecutive Years Around the Growth Spurt

For the absolute DPSI values, a significant increase was observed between years 1–3 and 2–3. When normalizing for body weight force, a significant decrease in relative DPSI values between years 1–2 and 1–3 was revealed. Since lower DPSI values represent good dynamic postural stability per definition, dynamic postural stability seems to worsen over the three consecutive years around the growth spurt when looking just at the absolute values. In contrast, when relative DPSI values are considered, an improvement in dynamic postural stability is observed, with greater improvement between years 1 and 2 than between years 2 and 3. Thus, the increase in body weight between years one to two probably exceeds the decrease in relative DPSI (i.e., the actual improvement in the ability of dynamic postural control), ultimately leading to increased absolute DPSI values (i.e., seemingly worsened dynamic postural control). From a purely theoretical standpoint, this is entirely plausible, given that the absolute DPSI depends on the fluctuations of the measured forces in the x, y and z directions while landing with reference to the values of the same person standing quietly on the force plate, and thereby increases when gaining weight (remember: force (F) is equal to the mass (m) of an object times its acceleration 1) according to the following formula: F = m×a). In addition, it is well known that neuromuscular control changes during the growth spurt, since peak height velocity is the time that the body allows longitudinal growth by the weakening of ligaments, muscles and bones resulting in poor neuromuscular control (Backous et al., 1988; Watkins and Peabody, 1996). Thereby, the legs reach their maximum growth rate before the age at peak height velocity, and in contrast, the trunk reaches its maximum growth rate typically 1 year after the age at peak height velocity (Malina et al., 2004; Philippaerts et al., 2006).

Correlation of DPSI With Age and Maturity Offset

For the absolute DPSI values of all 3 years, there were significant correlations with age and biological maturation, with the strongest correlation in year two for age (r = 0.421) and in year three for maturity offset (r = 0.397). The corresponding R^2 values were relatively low, with values up to 0.16 (year 3) for maturity offset and 0.18 for age (year 2), explaining only 16 and 18% of the variability in DPSI. However, when normalized for body weight force, relations between relative DPSI, age and maturity offset were no longer present for all 3 years, suggesting that DPSI is strongly dependent on body weight, which increases during puberty. Regular testing of athletes such as youth competitive alpine skiers is an important part of sports. When testing physical fitness in youth skiers around the growth spurt, performance is dependent on the status of biological maturation and thereby body weight, which makes it challenging to interpret the results with only chronological age as a dependent factor between individuals. In this study, the DPSI values increased but decreased when normalized for body weight force. This further stresses the importance of body weight normalization of physical fitness tests in youth skiers around the growth spurt.

Study Limitations

First, the methodological approach used in this study to evaluate postural stability in a controlled laboratory environment is only a very rough imitation of the real ski situation, but with significant advantages in terms of standardization of test conditions.

Second, by definition, for the single-leg landings, only trials in which the foot was in continuous contact with the force plate for at least 3 s were valid. However, when the field tests were conducted in the first year, not enough priority was given to on-site monitoring of this aspect, so the number of subjects with three valid annual test scores was significantly limited, which may have led to some selection bias, with subjects with very poor dynamic postural stability abilities being more likely to have been excluded.

Third, since the underlying cohorts of the reliability experiment and the main experiment differ in age and sporting level, direct transferability of the reliability experiment results to the main experiment is limited and should be undertaken with caution. However, the young, healthy adults' cohort of the reliability experiment may serve as a representative sample for many application contexts of future studies, and at least the DPSI average and standard deviation magnitudes in year three of the main experiment (average absolute DPSI values: 211.5 ± 41.3 N) are comparable to those of the young, healthy adults in the reliability experiment (average absolute DPSI values: 233.8 ± 41.4 N). In year three of the main experiment, the skiers were approaching the age of 16 years, and therefore the anthropometry and balance abilities of the competitive skiers (at this time still semiprofessional athletes) might have been comparable to those of the young, healthy adults.

Fourth, the different number of repetitions and leg-related unilaterality versus bilaterally of the data underlying the reliability and main experiment calculations, respectively, may further limit a direct transferability between the two experiments. However, both experiments had a clearly defined purpose, which may justify the different analysis approaches. The reliability experiment aimed to account for habituation effects while avoiding bias due to increasing fatigue. Conducting five test repetitions on both sides on the same day, instead of five test repetitions on one side on the same day, likely would have introduced increasing exhaustion. The main experiment sought to increase the individual representativeness of the test results while minimizing testing efforts, which is crucial when screening large cohorts of athletes such as youth competitive alpine skiers. Accordingly, in the main experiment, a youth skiers' representative absolute/ relative DPSI values were obtained by averaging his/her four performed trials (i.e., two trials per leg) into a representative mean value. Notwithstanding, the one-sided DPSI values (as described in the reliability experiment) may provide further valuable insight and clearer guidelines for potential countermeasures once a global balance deficit has been identified in a particular skier.

Fifth, despite a clear theoretical and conceptual link between DPSI, skiing performance, and the risk of injury, the present study did not investigate such a potential relationship. This study also did not explore DPSI leg asymmetries and the influence of leg dominance on unilateral DPSI values. Since these aspects have not been the focus of previous studies, they should be the subject of future research.

CONCLUSION

The DPSI during DJSLLs may basically be considered a reliable and sensitive measure for assessing dynamic postural control. However, test-retest reliability may differ between specific cohorts, and biological maturation should be considered a potential confounder when interpreting absolute DPSI test values of youth skiers during the growth spurt. This is also apparent, among other things, from the fact that in the present study, an increase in dynamic postural control ability over the years was observed for body weight force normalized, relative DPSI values, whereas a deterioration occurred for absolute DPSI values.

DATA AVAILABILITY STATEMENT

The datasets presented in this article are not readily available because their access is restricted to protect the interests of the project partner Swiss-Ski and their athletes. Requests to access the datasets should be directed to joerg.spoerri@balgrist.ch.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Cantonal Ethic Commission Zurich (KEK-ZH-NR: 2017-01395). Written informed consent to participate in this study was provided by the participants' legal guardian/next of kin.

AUTHOR CONTRIBUTIONS

JS and WF conceptualized and designed the study. JS recruited the participants, and LE, JJ and JS collected the data. LE, JJ and FO processed the data, and KK and JS performed the statistical analysis. All authors substantially contributed to the interpretation of the data. KK and JS drafted the current manuscript; all authors revised it critically, approved the final version of the manuscript, and agreed to be accountable for all aspects of the work.

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Sinusoidal Optic Flow Perturbations Reduce Transient but Not Continuous Postural Stability: A Virtual Reality-Based Study

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Optic flow perturbations induced by virtual reality (VR) are increasingly used in the rehabilitation of postural control and gait. Here, VR offers the possibility to decouple the visual from the somatosensory and vestibular system. By this means, it enables training under conflicting sensorimotor stimulation that creates additional demands on sensory reweighting and balance control. Even though current VR-interventions still lack a well-defined standardized metric to generate optic flow perturbations that can challenge balance in a repeatable manner, continuous oscillations of the VR are typically used as a rehabilitation tool. We therefore investigated if continuous sensory conflicts induced by optic flow perturbations can challenge the postural system sustainably. Eighteen young adults (m = 8, f = 10, age = 24.1 ± 2.0 yrs) were recruited for the study. The VR was provided using a state-of-the-art head-mounted display including the virtual replica of the real environment. After familiarization in quiet stance without and with VR, bipedal balance was perturbed by sinusoidal rotations of the visual scenery in the sagittal plane with an amplitude of 8° and a frequency of 0.2 Hz. Postural stability was quantified by mean center of mass speed derived from 3D-kinematics. A rmANOVA found increased postural instability only during the first perturbation cycle, i.e., the first 5 s. Succeeding the first perturbation cycle, visual afferents were downregulated to reduce the destabilizing influence of the sensory conflicts. In essence, only the transient beginning of sinusoidal oscillation alters balance compared to quiet standing. Therefore, continuous sinusoidal optic flow perturbations appear to be not suitable for balance training as they cannot trigger persisting sensory conflicts and hence challenge the postural system sustainably. Our study provides rationale for using unexpected and discrete optic flow perturbation paradigms to induce sustainable sensory conflicts.

Keywords: postural control, virtual reality, sensory reweighting, sensory conflicts, optic flow perturbation

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INTRODUCTION

To inhabit the world with all its unpredictable, variable environmental and situational contexts, a powerful yet flexible postural system is crucial (Horak, 2009). This flexibility of the postural system is guaranteed by appropriate changes in muscle activation that generate joint torques correcting for deviations from the desired orientation (Peterka, 2002). To orchestrate those adjustments, particularly in response to balance challenges, the central nervous system (CNS) requires reliable sensory feedback to generate efferent commands that produce corrective muscle torque to stabilize the human body. For this purpose, multiple sensory channels are simultaneously integrated in the CNS, including visual, vestibular, and somatosensory input. It has been demonstrated that the integration of sensory information appears to be dynamically regulated to adapt to changing environmental conditions and available sensory information (Hwang et al., 2014). Hereby, reliable sensory information from one sensory system is preferred over less reliable information from another sensory system (Kiemel et al., 2002). This is referred to as sensory reweighting (Nashner and Berthoz, 1978).

To improve the postural control system, whether to prevent falls in old age, to regain performance after injury, or for training purposes in sports, balance training is recommended to perturb the different sensory systems required for balance. Mostly, unstable support surfaces are used for this purpose (Taube et al., 2008), which force the subject to utilize the optimal source of sensory information. Closing the eyes or pitching the head (Johnson and Van Emmerik, 2012) can increase the difficulty of the balance task by further modifying the reliability of the visual and/or vestibular system, thus creating sensory conditions that are more challenging. The ability to select and reweight sensory information adaptively is considered one of the most important factors for postural stability, e.g. in the elderly (Horak et al., 1989b). Balance exercises that challenge the sensory systems and specifically target multisensory integration mechanisms were shown to improve sensory reweighting and balance control (Allison et al., 2018).

To increase the variability of balance exercises and provide broader access to sensory perturbations, virtual reality (VR) offers completely new possibilities (Chander et al., 2019). VR provides an interface between humans and computer systems that enables natural and intuitive interaction within the simulated three-dimensional environment, thereby allowing researchers to systematically modulate the visual input almost without limitations and by this means to manipulate the interaction between the organism and the environment in an arbitrary but still standardized way (Hedges et al., 2011). Whereas the visual input in conventional balance training usually is binary (eyes closed or eyes open), VR applications have the ability to decouple the visual from the somatosensory and vestibular systems in a more fine-grained manner by providing manifold possibilities of optic flow perturbations (Canning et al., 2020). By this means, it can induce conflicting sensorimotor stimulation that creates additional demands on sensory reweighting and balance control (Martelli et al., 2019) that are necessary to evoke cortical reorganization and neuroplasticity (Adamovich et al., 2009).

For instance, oscillating visual fields, i.e. moving room paradigms, can trigger these conflicts and the associated postural instability. The visual field in the virtual environment can be spatially manipulated to target the neuromuscular skills required for balance (Osaba et al., 2020). Adamovich et al. (2009) describe the incorporation of this element in balance training as the logical "next step", as it may open a new direction in balance training and yields valuable implications to prevent falls or (sports-) injuries. In this context, Allison et al. (2018) examined the effect of sensory-challenge balance exercises on sensory reweighting capability in older adults. The authors found significant improvement in sensory reweighting and balance following balance exercises specifically targeting multisensory integration mechanisms through computerized, variable surface and/or visual environment motion. They concluded that their results provide a scientific rationale for sensory-challenge exercises to reduce fall risk. Based upon such findings, improvement in multisensory interactions has been suggested as a potentially fruitful area for new interventions (Bugnariu and Fung, 2007).

Consequently, the use of VR as a rehabilitation tool has advanced substantially within the last decade (Juras et al., 2019). There is growing evidence that when combined with conventional rehabilitation, VR offers improved benefits for balance and gait rehabilitation in neurological patients (for review, see Cano Porras et al., 2018). However, despite its growing popularity and proven efficacy to perturb balance and induce postural instability (Horlings et al., 2009; Chiarovano et al., 2017), information about optimal intervention programs (e.g., dosage and tasks) and defined paradigms for disturbing balance is still scarce. This hampers the development of both standardized interventions and the optimal use of VR in balance rehabilitation and balance training. Specifically, no established metric exists for creating a virtual environment that can perturb balance in an effective and repeatable manner. To be useful in balance training, however, the perturbation effects have to be well preserved in order to provide a permanent challenge to the postural system (Heidner et al., 2020).

Therefore, the purpose of this study was to investigate if continuous sensory conflicts can challenge the postural system sustainably and can thus deliver paradigms with additional demands on sensory reweighting and balance control processes. To provide a paradigm with conflicting sensorimotor stimulation, we used VR to create a synthetic replica of our real laboratory and to generate continuous rotatory oscillations of this virtual laboratory in the sagittal plane.

MATERIALS AND METHODS

Subjects

Eighteen young adults (m = 8, f = 10, age = 24.1 ± 2.0 yrs) with no conditions affecting balance were recruited for this study. All volunteers had no previous experience regarding VR. The study was approved by the ethics committee of the University of Freiburg and in accordance with the declaration of Helsinki. The participants provided their written informed consent to participate in this study.

Virtual Environment

The virtual environment was provided via a head-mounted display (HMD) (HTC Vive pro eye, HTC Corporation, Taoyuan City, Taiwan), with a resolution of 1440 × 1600 pixels per eye, a field of view of 110° and an update rate of 90 Hz. The stereo graphics were rendered with an AMD Ryzen

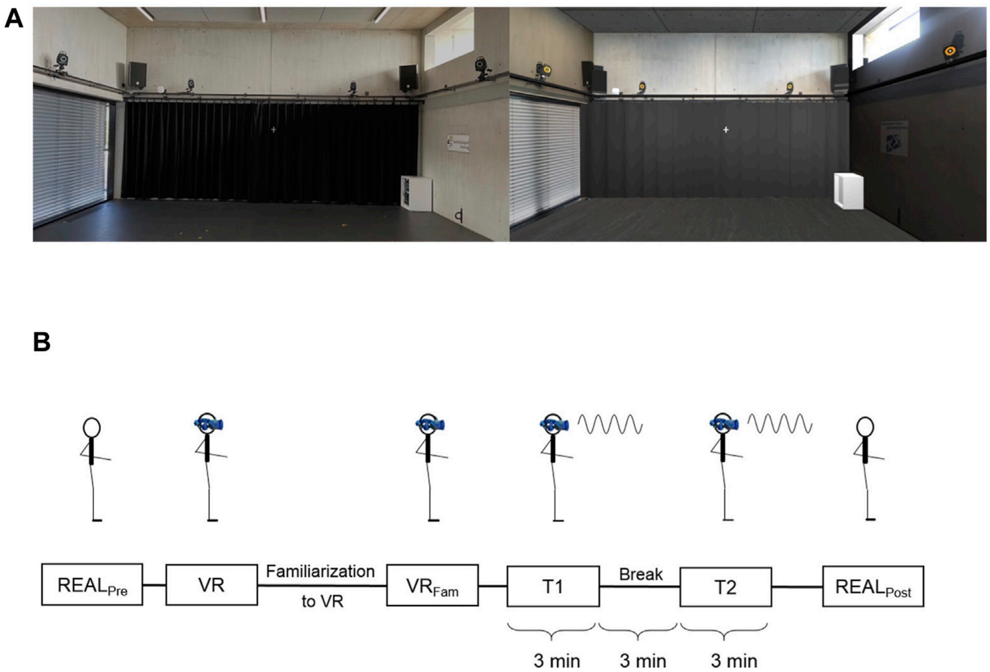

FIGURE 1 | Experimental setup **(A)** Real environment (left) and virtual replica of the real environment (right) that was provided to the subjects via HMD **(B)** Schematic display of the experimental protocol.

9 3900X processor and Nvidia GeForce RTX 2080 graphics card. The HTC Vive system includes a lighthouse tracking system, which tracks head position and orientation. This data updated the perspective displayed in the HMD, enabling the participants to move freely in the VR. The visual content in the HMD was a synthetic replicate of the real environment, thus a virtual measurement laboratory. The size of the virtual laboratory was scaled to match the real environment. The VR space was rendered in Unity3d (Unity Technologies, San Francisco, CA). **Figure 1A** illustrates a comparison of the real environment and our virtual replicate.

Experimental Procedure

Initially, subjects were informed about the protocol and the measurement conditions. For each measurement condition, the participants were instructed to take a comfortable bipedal stance with feet shoulder-width apart and to keep arms to the side. During data acquisition, participants were asked to stand still, in a relaxed manner and to look at a "+" placed on the wall in front of them at eye level (in VR and real environment).

Figure 1B summarizes our experimental protocol. First, participant's balance was assessed in the real environment with eyes open ("REAL$_{Pre}$") and in the virtual environment. In the virtual environment ("VR"), we measured balance immediately after the participant put on the HMD and after a 3-min familiarization phase in which the subject could move freely in the virtual space ("VR$_{Fam}$"). Here optic flow provided by the HMD was not manipulated and provided reliable visual perception. Each of these conditions included three trials with each trial lasting 20 s.

In the following perturbation session, participants completed two 3-min trials (T1 and T2) while being exposed to continuous anterior-posterior rotation of the VR in the sagittal plane (pitch of the virtual room). The rotation was prescribed as a sinusoidal signal with an amplitude of 8°, a frequency of 0.2 Hz and no phase shift. These specifications were chosen as high amplitudes are known to evoke larger postural responses (Dokka et al., 2009; Chiarovano et al., 2015) and frequencies of 0.2 Hz were shown to be within a comfortable range with maximum entrainment in healthy adults (Peterka, 2002). The rotation axis of the visual field was 8.8 cm above the floor, thus approximately through the ankle joint axis. Trial length was chosen according to Amiri et al. (2019) who reported 3 min to be optimal for perturbation design, as it generally does not cause fatigue provided that adequate rest is allowed between trials. In-between both trials a 3-min break was implemented, where the subjects were asked to take a seat while still being immersed in the VR. After the break, a second 3-min perturbation trial (T2) was made. Subsequent to T2, subjects removed the HMD and balance was assessed in the real environment three times à 20 s ("REAL$_{Post}$").

Measurements

Kinematic data were captured by a Vicon MX digital optical motion capture system with nine infrared cameras (Vicon Motion Systems Ltd., Oxford, United Kingdom) operating at 200 Hz. Thirty-nine retroreflective markers were attached to the subjects according to the Vicon Plug-in Gait Model. In combination with anthropometric measurements, this model allows to compute the three-dimensional coordinates of the joint centers, the segments' center of mass as well as the whole body's center of mass (COM).

Data Analysis

Recorded data was analyzed using Matlab (The Mathworks, Natick, United States). Kinematic data were filtered using a fourth order low-pass Butterworth filter with a cutoff frequency of 8 Hz. For all trials, the filtered trajectory of the COM was used to calculate the mean speed of COM sway.

To compare COM sway during the perturbation trials to the non-perturbation trials, i.e. natural standing trials, data were segmented into nine blocks; each including four perturbation cycles (= 20 s as in the conditions preceding the perturbations). Mean COM sway speed was calculated for each block to show the temporal course over the exposure to perturbations.

Furthermore, the amplitudes of the anterior-posterior translations of the COM were converted into angles describing the rotation of the COM around the ankle joint. This conversion allows relating the amplitude of rotation angles to the amplitude of the visual stimulus. Herein, the so-called gain represents the ratio of COM response amplitude at the vision stimulus frequency (0.2 Hz) to the vision amplitude (8°) (Peterka, 2002). The magnitude of the response provides information about the relative weight of the visual contribution to balance and therefore enables a description of the dynamic characteristics of the balance control system. Hereby, a change in gain is interpreted as reweighting of the visual modality, that is, a decrease in gain indicates lower weighting (decreased coupling) to the visual stimulus. The frequency response function (FRF) at the stimulus frequency was obtained by dividing the discrete Fourier transform (DFT) of the time series of COM rotatory displacement and of the oscillatory component of stimulus motion. Ultimately, gain is the absolute value of the FRF at the stimulus frequency. As described in Jeka et al. (2010), we followed the standard cycle-by-cycle analysis, whereby the FRF of each cycle (in our case, 36 cycles with 5 s each) is processed separately. Gain values were also grouped in nine blocks and for each block the mean gain was calculated as the average of the DFT coefficients over the cycles.

Statistics

We evaluated the dependent variables COM mean sway speed and gain. To test for the effect of the VR itself on postural stability, we first used a one-way repeated measure analysis of variance (rmANOVA) to compare between the visual conditions $REAL_{Pre}$, VR and VR_{Fam}.

To test for effects of optic flow perturbation, we conducted a rmANOVA on the average of each outcome measure taken at the time of interest: VR_{Fam}, the mean of the first block ("First_T1" and "First_T2") and the mean of the last block ("Last_T1" and "Last_T2") of both trials, respectively, and the Post condition ($REAL_{Post}$). Given a significant main effect of time, we performed the following planned, Šídák's corrected post-hoc, pairwise comparisons: VR_{Fam} versus First_T1/First_T2, VR_{Fam} versus Last_T1/Last_T2, First_T1 versus Last_T1, First_T2 versus Last_T2, and VR_{Fam} versus $REAL_{Post}$. To test for differences in visual afferent integration between First and Last block of T1 and T2, we used a rmANOVA of the gain values. We report effect size using partial eta squared (η_p^2) for main effects.

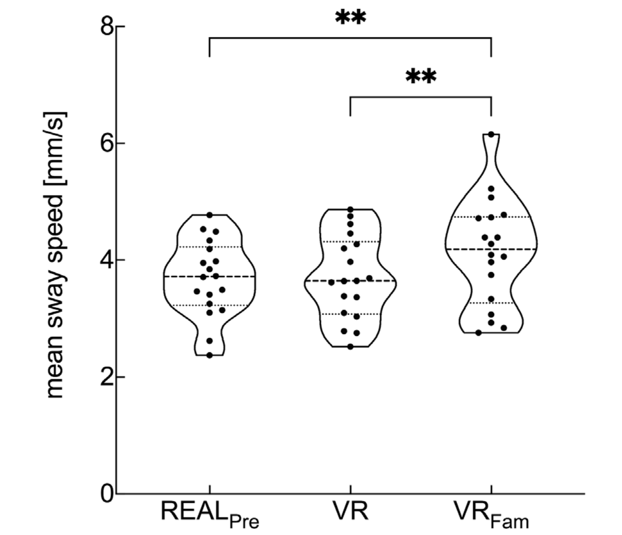

FIGURE 2 | Violin plots of mean COM sway speed for conditions with reliable visual input preceding the optic flow perturbations. Visual conditions are real environment ($REAL_{Pre}$), virtual replica of the real environment (VR), and virtual replica of the real environment after 3-min familiarization to VR (VR_{Fam}). Violin plots represent data from each subject and show median values (dashed horizontal line), lower and upper 25th and 75th percentile values (dotted lines) and error bars (spanning smallest to largest individual values). $^{**}p < .01$.

All statistical analyses were conducted using Prism 9 (GraphPad Software, San Diego, CA). All data are presented as mean value and 95% confidence intervals. For all statistical tests, the level of significance was set to $p = .05$.

RESULTS

The visual condition showed a statistical main effect on the COM mean sway speed ($F_{1.75,\ 31.5} = 12.27$, $p < .001$, $\eta_p^2 = .405$). We found no significant difference in mean sway speed between $REAL_{Pre}$ and VR. However, there was a significant difference between $REAL_{Pre}$ and VR_{Fam} ($p = .003$) as well as between VR and VR_{Fam} ($p = .002$), respectively (**Figure 2**).

Figure 3A shows the effects of prolonged optic flow perturbations on postural stability ($F_{3.606,\ 61.3} = 5.08$, $p = .002$, $\eta_p^2 = .23$). In First_T1, optic flow perturbation elicited no greater mean sway speed ($p = .238$) compared to VR_{Fam}, whereas First_T2 showed greater mean sway speed as in VR_{Fam} ($p = .001$). From First_T1 to Last_T1, mean sway speed did not change ($p = .963$). In the second trial, mean sway speed decreased from First_T2 to Last_T2 ($p = .014$) and did not show differences to VR_{Fam} anymore ($p = .856$). No difference existed in the COM mean sway speed of First_T1 and First_T2 ($p = .686$). Immediately following cessation of optic flow perturbations of the second trial ($POST_{EO}$), participants exhibited similar ($p = .999$) mean sway speed as during VR_{Fam}.

For the gain values we found no main effect of time ($F_{2.362,\ 40.15} = 5.08$, $p = .496$, $\eta_p^2 = .043$) (**Figure 3B**).

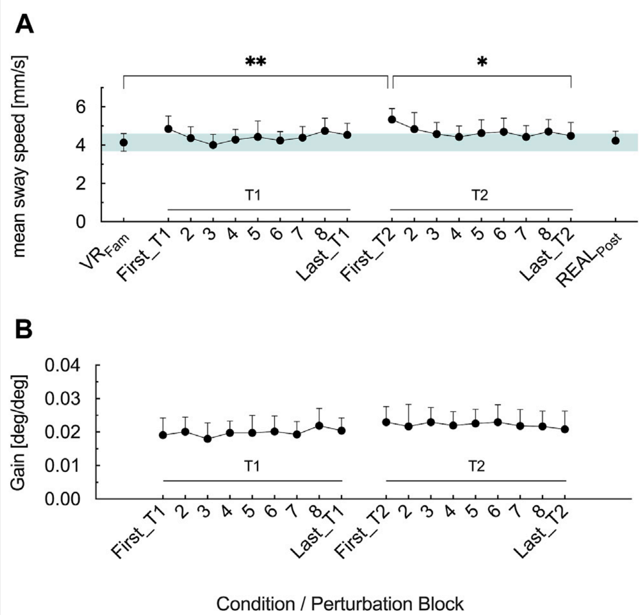

FIGURE 3 | (A) Mean COM sway speed for the reference condition (VR_{Fam}) and the nine perturbation blocks in T1 and T2 **(B)** Gain values for the nine perturbation blocks in T1 and T2. Error bars are the 95% CI of the mean. Green shaded horizontal bar in **(A)** highlights the 95% CI of VR_{Fam}. **p* < .05, ***p* < .01.

To verify whether significances were merely masked by block building (nine blocks à four perturbation cycles) especially at the beginning of representing the optic flow stimulus, we subsequently "zoomed in" the first block of T1 and T2 (= first four perturbation cycles) and checked for perturbation-to-perturbation differences. Therefore, we conducted a rmANOVA for both First_T1 and First_T2 and compared it with the reference condition VR_{Fam}.

Across perturbation cycles, mean sway speed exhibited significant main effects in First_T1 and First_T2 ($F_{2.646,\ 44.98}$ = 8.826, $p < .001$, $\eta_p^2 = .342$ and $F_{2.087,\ 35.48}$ = 7.168, $p = .002$, η_p^2 = .297, respectively) (**Figure 4A**). Results revealed that the first cycle of optic flow perturbation elicited 50% greater sway speed in the T1 ($p = .003$) and 69% in T2 ($p = .003$), respectively, compared to VR_{Fam}. In T1, the first perturbation cycle elicited significantly greater mean sway speed than the second ($p = .025$), the third (p = .017) and the fourth (p = .001) perturbation cycle. In T2, mean sway speed during the first perturbation cycle was greater than during the second ($p = .022$) and the third perturbation cycle (p = .036). There tended to be a difference between the first and the fourth perturbation cycle (p = .051).

Similarly, as for mean sway speed, time exhibited a significant main effect on the gain values in First_T1 and First_T2 ($F_{1.848,\ 31.41}$ = 5.548, $p = .01$, $\eta_p^2 = .246$ and $F_{1.950,\ 33.14}$ = 3.691, $p = .037$, $\eta_p^2 = .178$, respectively) (**Figure 4B**). In T1, gain values in the first perturbation cycle were similar as in the second ($p = .154$) and the third ($p = .069$) but greater than in the fourth perturbation cycle ($p = .006$). In T2, greater gain values were measured for the first perturbation cycle compared to the second (p = .045) and the third ($p = .038$) perturbation cycle. The fourth cycle did not differ to the first (p = .16).

DISCUSSION

The main purpose of this study was to investigate the effect of continuous optic flow perturbations as a trigger of sensory conflict on postural stability. We employed a swinging room paradigm where a scaled and lifelike VR rotated sinusoidally around the subjects' ankle joint in the sagittal plane to induce the visual perception of self-motion that contradicts vestibular and somatosensory sensory input. By this means, we aimed to elicit sensory re-weighting processes that could be used for balance training purposes.

The results suggest that these perturbations produce impaired balance control compared to quiet standing with reliable visual information, but only in a specific time domain. The main findings of this study where that 1) only the first perturbation cycle created sensory conflicts that were strong enough to elicit postural instability and 2) visual afferents were downregulated after the first perturbation cycle to reduce the sensory conflict and therefore postural instability.

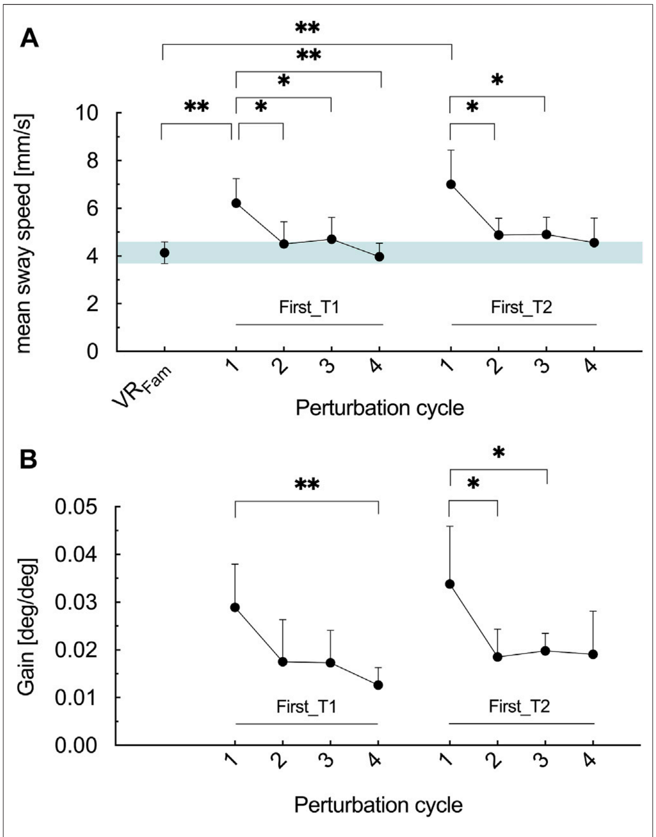

FIGURE 4 | (A) Mean COM sway speed for the reference condition (VR_{Fam}) and the four perturbation cycles of block one in T1 (First_T1) and T2 (First_T2) **(B)** Gain values for the four perturbation cycles of block one in T1 (First_T1) and T2 (First_T2). Error bars are the 95% CI of the mean. Green shaded horizontal bar in **(A)** highlights the 95% CI of VR_{Fam}. **p* < .05, ***p* < .01.

VR-induced optic flow perturbation is thought to be one mean for balance training, as the motor system increases robustness of motor control in the presence of perturbations (Santuz et al., 2018; Munoz-Martel et al., 2019). Training programmes using sensory perturbations to exercise dynamic stability can enhance sensory information processing within the motor system (Hamed et al., 2018). To trigger the additional response and thus have a training effect, however, the perturbations must be challenging enough (Hamed et al., 2018). Chiarovano et al. (2015) and Heidner et al. (2020) demonstrated that 25 or 30 s exposure to optic flow perturbations in VR can cause postural instability. The authors also suggest a clinical application of these perturbations as their findings demonstrate that motor control strategies can be challenged by optic flow perturbations without physically perturbing the subject (Heidner et al., 2020). Yet, past investigations provided rational that balance control in VR is per se compromised with respect to the real environment (Horlings et al., 2009; Kelly et al., 2019). Therefore, one could argue that postural instability induced by moving room paradigms is not solely due to the optic flow perturbations, but also a consequence of the VR itself. In our study, however, we found no difference in body sway between $REAL_{Pre}$ and VR indicating that our VR itself did not trigger any postural instability. Our results support the work by Assländer and Streuber (2020), showing that a state-of-the-art VR device with photorealistic and lifelike VR scenarios provides visual conditions that equal real environments and consequently evoke body sway behaviour that is similar to real life. Menzies et al. (2016) also support the notion that better technical devices reduce spontaneous sway and dedicated their findings to higher fidelity of the visual surround. However, we observed greater postural instability after the 3-min familiarization (VR_{Fam}) compared to the initial VR condition and $REAL_{Pre}$. Work by Robert et al. (2016), who also used a recent VR device, might explain this finding. For their VR content, the authors used a 3D filmed visual representation that showed the laboratory room. They reported no difference for VR and the real environment for static balance. However, dynamic balance tasks were more perturbed in the VR compared to the real environment. They conclude that dynamic or more challenging balance tasks are impaired in VR because of sensory conflicts due to for example latency of the HMD. The 3-min walking familiarization in our study might have triggered this phenomenon as well. Consequently, we referenced postural stability during the optic flow perturbations to VR_{Fam}, to ensure that increased instability during the perturbations is a consequence of the perturbation itself.

This postural instability during the perturbation trials was solely observed for First_T2 compared to VR_{Fam}; for First_T1 mean sway speed was not increased compared to VR_{Fam}. In an experiment with support-surface perturbations, Horak et al. (1989a) observed overreacting postural responses when a small platform perturbation was preceded by a series of larger perturbations. This phenomenon was also shown for optic flow perturbations (O'Connor et al., 2008). Similarly, in our study, subjects may have shown an overreacting postural response in First_T2 because they were expecting a larger optic flow stimulus. Mean sway speed then adapts, with Last_T2 resembling VR_{Fam}.

Besides this, the only difference between T1 and T2 is of a temporal nature. Although speculative, the additional time spent in the VR may have increased visual reliance on the virtual scenery, thus becoming more prone to optic flow perturbations in T2 than in T1. The 3-min static break between T1 and T2 may have further facilitated this effect. The fact that the optic flow perturbation did not induce significant postural instability in First_T1 contrasts with the literature (Chiarovano et al., 2015; Heidner et al., 2020). To account for this incoherent and unexpected finding, we more closely inspected the first block in T1 and T2, respectively, and conducted a perturbation-to-perturbation analysis. Striking here is that the first perturbation cycle of each trial elicited greater mean sway speed compared to VR_{Fam} and that mean sway speed during the first perturbation cycle was also greater than during the remaining cycles of the first blocks in T1 and T2, respectively. This suggests that our optic flow perturbation paradigm can only initially trigger a sensory conflict, which is sufficient to cause postural instability. This finding is supported by Nashner and Berthoz (1978) and Bronstein (1986), who also showed this rapid habituation of sway response to visual scene movement. Especially the first presentation of optic flow perturbation induced larger instability compared to the following presentations (Bronstein, 1986). In a protocol with 45 s exposure to sinusoidal optic flow perturbations O'Connor et al. (2008) observed substantial reduction in sway speed in the first 10 s. The authors suggest changes in sensory reweighting as a possible mechanism. That subjects have the greatest reduction after first trial was also demonstrated by Sundermier et al. (1996) who exposed subjects to successive forward translations of the visual field. This observation may have been induced by the unexpected incongruence of vision with other balance-relevant inputs that is destabilizing (Sundermier et al., 1996). If optic flow perturbations are continuous and expectable, which is also true for our sinusoidal application, it is plausible that subjects were able to anticipate the perturbation after the first exposure and subsequently resisted to instability. This is in line with the findings of Chander et al. (2019) or Guerraz et al. (2001), who showed that expected optic flow perturbations resulted in smaller effects on postural control compared to unexpected optic flow perturbations. Moreover, even with pseudorandom and therefore unpredictable stimuli, the results by Peterka (2002) did not show evidence of adaption or habituation in the sway response after the first perturbation cycle. In our study, the continuous optic flow perturbations may have caused the subjects to adapt using a predictive strategy (Thompson and Franz, 2017) and thereby exhibit anticipatory behaviour (Amiri et al., 2019) that resists balance perturbation. In contrast, the first perturbation cycle in each trial differed by its abrupt start and the associated discrete characteristic from the following continuous cycles. The abrupt start results from the characteristic of sinusoidal stimuli. In contrast to a cosine or smoothstep function, a sine (with no phase shift) has its peak in the speed profile directly at stimulation onset. Abrupt optic flow perturbations are unpredictable and require a greater reactive and compensatory response that is more reflective of the direct effect of the optic flow perturbation (Kajrolkar et al., 2014).

This impact of the first perturbation cycle can be explained by the feedback model-based interpretation of balance control where feedback is provided by a weighted combination of sensory inputs, which are dynamically adjusted to maintain stability after changes in environmental conditions (Peterka, 2002). This has functional consequences for postural stability. In situations where environmental conditions suddenly change, i.e. perturbation onset, the availability of sensory orientation cues is altered, which causes a transient period of either under- or over-generation of corrective torque due to inappropriate (slow) adjustments of sensory weights for the new environmental condition (Peterka and Loughlin, 2004). Ultimately, the ability to quickly react to sudden changes is an important function of the balance control mechanism, as sudden environmental changes initiate dynamic postural adjustments that are centrally mediated and thus cannot occur instantaneously (Peterka, 2018). This poses a threat to stability (Polastri et al., 2012) and leaves subjects vulnerable to transient instability. Deficits in sensory reweighting are therefore considered one of the most critical factors for balance control in various patient groups and elderly (Horak et al., 1989b).

In the existing literature, these transient responses are commonly masked by whole-trial analyses (Reed et al., 2020) or the first perturbation cycle is discarded to avoid transient responses in order to get more reliable results for describing steady-state balance as a function of sensory input (Carpenter et al, 2001). However, in the perturbation-to-perturbation analysis, the transient instability is prominent and relatable to the enhanced integration of the visual afferents into the postural system as revealed by the increased gain values for the first perturbation cycle. The optic flow perturbation therefore has greater impact for postural control. Consequently, the VR-induced optic flow perturbation provides an initial destabilizing period preceding more stable postural control highlighting the role played by the dynamic regulation of sensorimotor integration. One interpretation of our findings is that the adaptation within trials may be explained by the decreased gain values and therefore the relative downregulation of visual feedback to reduce instability in the system for the following perturbation cycles. Reciprocally, subjects might have increased their awareness of reliable, available sensory information (Jeka et al., 2010). This can have implications for the rehabilitation of subjects with strong visual dependency for balance control, for example fall-prone elderly (Lord and Webster, 1990; Jeka et al., 2010). Incorporating optic flow perturbations during rehabilitation exercises may reweight sensory neural processing towards an upregulation of proprioceptive or vestibular inputs and reduce the dependency on vision to guide postural control.

Based on EEG data from Peterson et al. (2018) the time-dependent changes we report here may be simultaneously accompanied by changes in cortical activation. The authors used transient optic flow perturbations in young adults walking on a balance beam. They report increased electrocortical activation in parietal, occipital, and cingulate areas due to conflicting sensory information during balance. This suggests that such perturbations promote motor learning of a balance task in brain areas associated with integrating visual information and may thus reflect the brain's ability to adapt to variations in sensory input (Peterson et al., 2018).

Limitations

We observed no difference in postural stability between reality and VR. After the dynamic familiarization period, however, subjects standing balance was decreased. Due to lacking a control group who spent the familiarization statically, it remains questionable whether this increase in postural instability is a result of the walking interaction or an aspect of time or visual fatigue. To account for this, we referenced the perturbation trials to VR_{Fam} to ensure that postural instability is a consequence of the optic flow perturbation and not of the VR itself. After the first perturbation cycle, postural instability is no longer induced. Thus, it is not clear, whether the optic flow perturbations paradigm used in this study was not challenging enough for the participants. However, numerous studies have shown evidence for sway responses to sinusoidally optic flow perturbations, even after repeated exposure (van Asten et al., 1988; Peterka and Benolken, 1995). Another important limitation of our study design is that the results do not allow separating whether the decrease of postural instability after the first perturbation cycle is due to the continuous or the predictable nature of the stimulus. This hampers the development of optimal metrics for VR based rehabilitation paradigms.

Perspective

Our findings come with potential implications within the area of VR-based training and rehabilitation of balance. Many older adults may fall not because they are too weak or too stiff to respond, but as the results by Peterka and Loughlin (2004) predict, their risk of falling increases when environmental conditions change due to the too slow regulation of sensory weights. This is leveraged by the fact that, for instance, elderly succumb to a loss of reliability of the sensory feedback and rely more on vision than somatosensory and vestibular systems to maintain their balance (Lord and Webster, 1990; Jeka et al., 2010). This inaccurate perception may lead to compensatory responses that are inappropriate to correct for the loss of stability (Anson and Jeka, 2010), because somatosensation is the most important system for postural control, as it provides the fastest information processing (Campbell, 2007). Against this background, optic flow perturbations might be a helpful tool that should be considered when developing rehabilitation programs, as they could help patients to decrease reliance on visual information during balance control and upregulate reliance on somatosensory information for motor programming (Lee et al., 2021; Han et al., 2022). The implementation of this novel approach may enhance the activity of the somatosensory pathways to the postural system due to limited visual information input (Kim et al., 2021). Future research is needed to investigate the effects of training programs that include optic flow perturbations on postural control in individuals with altered somatosensory input due to musculoskeletal injury or aging. Furthermore, VR research and rehabilitation lacks of perturbation paradigms for creating effective and repeatable sensory conflicts. These sensory conflicts are needed to induce sensory reweighting and to improve the dynamic regulation of sensorimotor integration. Amiri et al. (2019) recently suggested using transient optic flow perturbations in random directions, as these stimuli are unpredictable and abrupt. Empirical evidence for this suggestion needs to be established yet.

CONCLUSION

Continuous sinusoidal optic flow perturbations appear not to be suitable to provide persisting sensory conflicts and hence to challenge the postural system sustainably. Therefore, it seems questionable to use these predictable perturbation paradigms as a tool for balance training. However, particularly the first perturbation cycle with its discrete characteristic is suitable for triggering instability. The application of discrete perturbations may elicit separate, distinct corrections that may be less easy to adapt to. Consequently, this sort of optic flow perturbation appears to be promising for balance training and balance rehabilitation.

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 Ethics committee of the University of Freiburg. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

All authors contributed substantially to this study; all contributed to the conception and design of the study. JK performed the experiments and analysed the data. JK, SR and DG discussed the results, and drafted the manuscript. All authors revised it critically. All authors gave final approval of the submission.

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