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Special Issue Reprint

Applied Biomechanics in Sports Performance, Injury Prevention and Rehabilitation

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Alfonso Penichet-Tomás

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Guest Editor

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Contents

About the Editor

Alfonso Penichet-Tomás

Alfonso Penichet Tomás is a Professor at the University of Alicante, with a PhD in Sports Science, Master's Degree in Educational Research, and is a second-cycle graduate in Physical Education and Sports Sciences and first-cycle graduate in Physical Education (University of Alicante). He is an Associate professor in Sports Science in the Department of General and Specific Didactics of the Faculty of Education at the University of Alicante and Sports Director of the Rowing Section of the S.E. Universitat d'Alacant.

As a member of the research group Health, Physical Activity, and Sports Technology (HEALTH-TECH), he leads the research line in rowing, and he has directed and directs both competitive R+D+i projects (CIGE/2022/15), as well as emerging R+D+i projects (GRE20-21A) and private projects (FEDERACIONREMO1-19TPA), all of them related to this sport. He has also participated in competitive R+D+i projects on technology and instrumentation applied to sports science, both funded by the Generalitat (GV/2021/098) and emerging R+D+i projects (GRE18-09). He has also participated in projects on sports performance analysis in different sports, funded by the Consejo Superior de Deportes (111/UPB10/12) (099/UPB10/12) and privately (COSTABLANCA1-13T) (AITEX6-13T) (HERCULES1-13T). At the same time, he has accumulated more than twenty teaching innovation projects and more than sixty publications indexed in reference databases.

Finally, he has more than fifty research papers published in *JCR* and *SJR* journals, as well as more than forty participations in national and international congresses. In addition, he is a reviewer of different journals indexed in *JCR* and *SJR*, highlighting his participation in the journal *Applied Sciences* as a guest editor in several Special Issues.

Preface

Dear colleagues,

I am pleased to present this Special Issue dedicated to the fascinating field of applied biomechanics in sports performance, injury prevention, and rehabilitation. Continuous advancements in modern technology have allowed sports scientists to gather increasingly precise and detailed information on movement and performance in the sports field. With the help of cutting-edge tools and devices, we now have access to a wider range of kinetic and kinematic parameters, enabling significant improvements in our ability to analyze various aspects of sports performance. The ability to monitor these parameters in real-time during actual practice and competition provides valuable insights that were previously difficult to obtain.

Biomechanical studies are fundamental in helping us understand how athletes move, perform, and interact with their environment. By examining key factors such as joint angles, force production, and movement efficiency, these studies guide the development of improved movement patterns, better postural habits, and strategies for energy conservation through the economy of movement. In particular, identifying inefficient or incorrect movement patterns is crucial for optimizing athletic performance. Small adjustments in technique, for example, can lead to noticeable improvements in both performance and injury prevention.

Moreover, biomechanical analysis plays a key role in injury prevention. Through a detailed understanding of movement dynamics, we are able to detect potential issues, such as improper alignment or muscle imbalances, that could lead to overuse injuries, strains, or sprains. By identifying these problems early, athletes can make conscious changes to their technique or training routine, reducing the likelihood of injury.

The main objective of this Special Issue is to highlight the latest advances in the field of biomechanics applied to sports. Our goal is to explore how these advancements can improve sports performance, prevent the risk of injury, and facilitate effective rehabilitation. This Special Issue brings together innovative research addressing the application of biomechanical principles in various sporting contexts, and we hope that the findings shared will inspire future work and collaboration in this rapidly evolving field.

Alfonso Penichet-Tomás
Guest Editor

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Editorial

Applied Biomechanics in Sports Performance, Injury Prevention, and Rehabilitation

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1. Introduction

Biomechanics has become an integral discipline in the sports field, enabling the optimization of performance, injury prevention, and athlete rehabilitation. Thanks to the development of new technologies and analytical methods, advances in biomechanics have enhanced the understanding of human movements and their implications for health [1]. This discipline combines the principles of mechanics with anatomy and physiology to examine how external and internal forces affect the body during exercise and sports [2]. Through its application, sports professionals not only improve the effectiveness and safety of training but also develop more effective injury prevention strategies and rehabilitation programs [3]. The study of the kinematics and dynamics of sports movements provides crucial information on how athletes generate and control forces [4]. The use of motion capture systems, accelerometers, pressure platforms, and force sensors has allowed for more precise quantification of athletes' performance, optimizing their techniques and reducing unnecessary effort [5]. In high-performance sports, biomechanical advances allow for the analysis of how small variations in movement, such as takeoff angles during jumping or the trajectory of a kick, influence performance [6]. Research has shown that biomechanically efficient techniques not only enhance the athlete's capability but also reduce the negative impact of external forces [7].

Biomechanics plays a key role in injury prevention, identifying and correcting movement patterns that increase the risk of harm [8]. Common injuries, such as sprains, muscle tears, tendinitis, or knee injuries, are often the result of inefficient or incorrect movements [9]. Research has shown that poor alignment during high-impact activities, such as running or jumping, can generate excessive stress on joints and soft tissues, predisposing athletes to chronic injuries [10]. In this regard, the integration of real-time monitoring devices has improved the ability to detect changes in movement mechanics that may precede an injury [11]. Advances in imaging technology and motion analysis have allowed physical therapists and doctors to closely monitor the recovery process of athletes, adjusting rehabilitation protocols according to the specific biomechanical needs of the patient. In several sports disciplines where knee injuries are common, biomechanical studies have enabled the design of more personalized protocols that accelerate recovery without increasing the risk of re-injury [12].

This Special Issue presents the recent advancements relating to the role of biomechanics in sports performance, injury prevention, and rehabilitation. Compiling results from various investigations, the findings underscore the importance of biomechanics in designing adaptive training programs that meet the specific demands of each sport, thereby optimizing athletic performance and reducing injury risks. This comprehensive approach not only improves performance but also contributes to effective injury prevention and rehabilitation strategies, making biomechanics an essential field in sports science.

2. An Overview of the Published Articles

Biomechanical analysis plays a crucial role in understanding and optimizing sports performance. This Special Issue compiles research addressing various aspects of the application of biomechanics to sports, covering topics ranging from movement variability in skiers to the integration of concurrent training into the development of swimmers and young soccer players. In the study on slalom skiing, Pérez-Chirinos Buxadé et al. (contribution 6) assessed how slope steepness and gate configurations influence movement variability (MV) and performance in elite skiers. By measuring MV through entropy and by analyzing total race time, the researchers found that steeper slopes and curved courses significantly increased MV, while better race times were correlated with lower MV. These findings underscore the role of motor control in technical skiing performance. On the other hand, Arsoniadis et al. (contribution 8) explored the effects of a concurrent training program combining maximum strength or muscular endurance with sprint interval training (SIT) in swimming. After six weeks, improvements in stroke rate (SR) were observed in groups incorporating dryland training, while no significant changes were noted in stroke length or stroke index. These findings suggest that concurrent training can enhance specific biomechanical components, particularly stroke rate, improving efficiency in front crawl swimming. In their study on the interaction between strength training and running performance, Pérez-Castilla et al. (contribution 9) compared velocity-based training (VBT) squat protocols to evaluate their acute effect on the maximal aerobic speed (MAS). All protocols reduced MAS compared to the control condition, with the greatest decrease observed with higher velocity loss during sets. These results indicate that the immediate application of VBT before endurance tests can negatively affect aerobic performance due to the induction of fatigue. Ioannou et al. (contribution 12) examined the impact of a combined balance and multidirectional plyometric training program in U-13 soccer players. Although no significant changes in postural stability were recorded after six weeks, a trend towards improvement was identified with regard to lower-limb explosive performance. However, visual restrictions impaired postural control, highlighting the importance of evaluating challenging conditions during training.

Two recent studies addressing biomechanical aspects related to muscle activation and isometric strength assessment offer new insights with utility in rehabilitation and training. In the first study, Pinho et al. (contribution 2) investigated antagonist muscle coactivation in thighs and ankles during the sit-to-stand task in post-stroke and healthy subjects. Dysfunctional coactivation was more pronounced in ankle muscles than in thigh muscles, suggesting that the distal segment may better reflect central nervous system dysfunction in post-stroke patients. These findings highlight the importance of conducting further studies to clarify the spatiotemporal variability in coactivation levels in this population. In the second study, Mellemkjar et al. (contribution 5) assessed the between-session reliability and concurrent validity of using the FysioMeter H-station to measure isometric quadriceps, hamstring strength, and the hamstring-to-quadriceps (H/Q) ratio compared to the values obtained with an isokinetic dynamometer (ID). The H-station demonstrated excellent relative reliability and moderate to strong concurrent validity for hamstring and quadriceps strength compared to the ID. These results indicate that the H-station is a reliable tool for tracking relative changes in isometric strength, providing a practical alternative to the ID for clinical and sports environments.

Biomechanics is also a crucial discipline in terms of understanding how athletes perform and how they can reduce the risk of injuries. Recent studies have highlighted the importance of biomechanical analysis in various sports and contexts, shedding light on the impact of training loads, injuries, and postural strategies on performance and health. The first study by Lehnert et al. (contribution 3) explored the impact of competitive match play and subsequent training on landing biomechanics. The results indicated that the competitive load, including matches and training, did not negatively affect the landing mechanics of the players or increase the risk of ACL injuries. This suggests that, within the context of the competitive microcycle, there may not be a significant biomechanical impact on lower limb movements during landing, thus alleviating concerns about the risk of injury during regular training and competition cycles in young female athletes. In a different context, Gräf et al. (contribution 7) examined the upper-body postures of nurses during manual patient handling, a physically demanding task that places significant strain on the body. The findings revealed considerable differences in posture angles and movement accelerations across nurses and between different repositioning sequences. These results highlight the importance of ergonomic training in healthcare settings, emphasizing that nurses who engage in manual handling may benefit from targeted interventions aimed at reducing physical strain. A study by Kirkham et al. (contribution 10) further expands the scope of biomechanics by investigating the effects of concussions on static postural stability. The results demonstrated that individuals with a history of concussions exhibited greater postural sway compared to those without such a history. Furthermore, individuals with multiple or recent concussions showed more pronounced postural instability. These findings underline the long-term effects of concussions on balance and stability, which can have serious implications for athletes' performance and increase the risk of further injuries. Lastly, León-Guereño et al. (contribution 11) examined the injury incidence in "Traineras" rowing, a traditional competitive rowing modality in northern Spain. The study found significant differences in injury patterns between male and female rowers, with men reporting a higher injury incidence. This research is vital for developing sport-specific injury prevention programs that can help rowers to maintain performance levels while reducing injury risks in traditional rowing disciplines.

Biomechanics plays a key role in rehabilitation, particularly in improving movement quality and restoring function. Two recent studies explored the integration of biomechanics into rehabilitation practices, focusing on smartphone-based video analysis for shoulder exercises and robotic-assisted rehabilitation for lower-limb osteoarthritis patients. Lopes et al. (contribution 1) evaluated the use of smartphone-based 2D video analysis to assess movement quality during shoulder exercises, comparing it with the gold-standard 3D optoelectronic system. The results showed that while the smartphone application was generally in agreement with the 3D optoelectronic system, it demonstrated lower sensitivity in terms of detecting poor-quality movements. This suggests that smartphones can be a useful tool for home-based rehabilitation, enhancing patient engagement and autonomy, although they may be less effective in assessing low-quality exercises. In a different rehabilitation context, Castelli et al. (contribution 4) investigated robotic-assisted rehabilitation for elderly patients' post-hip or knee replacement surgery. The study compared robotic therapy with conventional treatment in 24 patients, assessing balance, walking, fatigue, and quality of life. The robotic-assisted group showed significant improvements in dynamic balance and walking, as well as reductions in motor and cognitive fatigue. These findings suggest that robotic systems provide more targeted interventions than conventional methods, improving balance and mobility in elderly patients, thereby reducing the risk of falls.

3. Conclusions

By incorporating biomechanical principles into training, rehabilitation, and injury prevention strategies, athletes and professionals in various fields can improve performance and reduce the likelihood of injuries. Collectively, these studies highlight the relevance of biomechanics in designing adaptive training strategies that address the specific demands of each sport and competitive level. This multidimensional approach contributes to the development of effective programs to optimize athletic performance and prevent injuries. Future research should continue to explore the dynamic relationship between biomechanics, performance, and injury risk, offering practical recommendations for optimizing training regimens and enhancing safety across a range of athletic contexts.

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List of Contributions

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Article

Influence of the Slope and Gate Offset on Movement Variability and Performance in Slalom Skiing

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Abstract: Adaptability to all types of terrain changes, slopes, and course settings is a key aspect related to the coordinative ability that elite skiers possess. In recent years, several studies have analyzed coordinative aspects of different motor actions via the assessment of movement variability (MV), an indicator of the motor control that assesses movement regularity. The aims of this study were (a) to evaluate the influence of different slopes and slalom (SL) gate offsets on MV and performance and (b) to assess the relationship between MV and performance. Four SL courses were set: a flat-turned (FT), a steep-turned (ST), a flat-straighter (FS), and a steep-straighter (SS). Five elite alpine skiers (21.2 ± 3.3 years, 180.2 ± 5.6 cm, 72.8 ± 6.6 kg) completed several runs at maximum speed for each SL course. A total of 77 runs were obtained. The use of an IMU accelerometer attached to the lower back of skiers measured MV through entropy. The skiers' performance was evaluated with the total time of each run. The one-way repeated measures analysis revealed that the steepness of the slope significantly increases skiers' MV, concretely between FS and ST courses ($p = 0.004$). Differences at the 10% level have been found between FS and SS and FT and ST courses ($p = 0.055$ and $p = 0.078$, respectively). For a given slope, turned courses (FT and ST) tend to produce a higher MV. In addition, faster times correlate with lower MV ($r = 0.587$, $p = 0.01$). It has been observed that both steeper and turned courses produce greater MV and that the best performing skiers have lower MV. Determining MV through entropy can be used to assess skiers' expertise regarding different types of slopes and gate offsets.

Keywords: steepness of the slope; gate offset; slalom course setting; movement variability; inertial measurement unit; entropy; performance; alpine skiing; elite alpine skiers; elite athletes

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1. Introduction

Alpine ski racing is a technical sport, which requires great coordinative ability from the skiers [1]. One of the main challenges of alpine skiing is the unpredictability of the descents. This is due to the high degree of uncertainty that arises when interacting with the environment at high speed [2]. When competing, the skier must be prepared to deal with variations in the terrain such as the steepness of the slope [3–5], and with variations in the course setting, such as the horizontal gate offset, that will regulate the amount that a course will be turned [6,7]. Therefore, in order to adjust to all types of terrain changes and course configurations, the central and peripheral nervous system of alpine skiers must work quickly and efficiently at every moment of the race [2,8]. Adaptability to the unpredictable and changing environments is clearly a key aspect that elite skiers possess.

In recent years, several studies have analyzed the coordinative aspects of different motor actions via the assessment of human movement variability (MV) [9–13]. MV is inherent within all biological systems and can be defined as the natural variations that typically occur during the execution of motor tasks when repeated multiple times [14–18], also known as execution variability [19] that assesses movement reproducibility [17].

In that context, research suggests that MV might be reduced as a function of practice [20,21] and experience [22,23]. The degree of mastery or adaptation to a task decreases the system's degrees of freedom and makes it highly predictable. Thus, experienced athletes are able to reproduce more consistently the same technical gesture, even in changing circumstances [10,17,24,25]. However, when athletes face new or complex situations, a technical gesture becomes less reproducible. The central nervous system of athletes is forced to find an optimized motor solution and must explore the full range of movement possibilities, which causes an increase in the degrees of freedom and thus in MV [26]. This increase in MV is what enables motor adaptation to a changing environment. Thus, MV should be considered a tool to stimulate learning [27–29].

In this regard, to ensure that exercise continues to provide an optimal stimulus for athlete development, it would be desirable to introduce different variations to the tasks that induce an increase in MV [10]. These variations become important in sports, especially in situation sports, such as alpine skiing, characterized by unpredictable and changing environments [30]. These athletes must continuously perceive and interpret external and internal information to adjust their actions and obtain effective and efficient solutions [31]. In addition, the application of different stimulus on-snow training such as a different steepness of the slope and variations in course settings, such as a different gate offset between gates, could produce a destabilizing effect on the body that would cause an increase in MV.

MV could be measured through the use of linear measurements such as standard deviation in the time evolution of a variable [32]. The standard deviation is useful to characterize the amount of variability present in the perceptual-motor system, as it quantifies the mean amount of dispersion around an averaged value. However, in some cases, different time series can exhibit the same standard deviation even if they do not share the same structure in time [33,34]. This limitation can be addressed with the use of non-linear tools, such as entropy, since it quantifies the amount of regularity and unpredictability of point-to-point fluctuations in large sets of time-series data and it is suitable for dealing with the complexity of biological systems [34–37]. Sample entropy (SampEn), Multiscale entropy (MSE), and Approximate entropy (ApEn) are the most popular methods for assessing data regularity in health and sports sciences [9,35,38,39]. SampEn is more robust to variations in time series length and can provide meaningful insights even with shorter data [40], and is one of the most widely used entropies in sports and health sciences [9–12,35,41,42].

In recent years, MV has been calculated through SampEn from the acceleration time series collected at the lower back of athletes using an inertial measurement unit (IMU) [9–13]. In alpine skiing, IMUs can offer a number of advantages that go beyond the limitations of traditional or laboratory methods [43–49]: compact and lightweight, easily portable and positionable, capable of wireless operation, and with the ability to capture a substantial amount of data (i.e., several runs), instantaneous data acquisition, and the optimization of data collection due to automatic synchronization among all the built-in sensors. However, to the best of our knowledge, no research has been conducted yet to evaluate MV through IMUs in the domain of alpine skiing, nor has it been evaluated regarding how the characteristics of the terrain (flatter vs. steeper slopes) and course setting (gate offset: i.e., turned vs. straighter courses) influence skier MV. Additionally, no article has been found that relates skier performance to MV. Therefore, the aims of this study were (a) to evaluate the influence of different slopes and slalom gate offsets on MV through an IMU and (b) to assess the relationship between MV and performance of elite alpine skiers. It was hypothesized that (a) steep and turned courses would increase MV and that (b) best-performing skiers would exhibit a lower MV.

2. Materials and Methods

This section is similar to that reported by Pérez-Chirinos Buxadé et al. [47].

2.1. Participants

Five elite alpine skiers belonging to the highest national level in Andorra with less than 40 FIS points in SL (aged 21.2 ± 3.3 years, weighing 72.8 ± 6.6 kg, with a height of 180.2 ± 5.6 cm) participated in this study. Written informed consent was obtained from all participants. All the procedures were conducted in accordance with the Declaration of Helsinki and were approved by the Ethics Committee for Clinical Sport Research of Catalonia (Study Number: 27/CEICGC/2020).

2.2. Procedures

2.2.1. IMU Device and Location

An IMU device (WIMU, Realtrack Systems, Almeria, Spain; weight: 70 g; size: 81 mm × 45 mm × 15 mm) was used in this study. Signals from the 3-axial accelerometer (range: ±400 G; sampling frequency: 1000 Hz) and the 3-axial magnetometer (range: ±8 gauss; sampling frequency: 100 Hz) were used. Before data collection, the inertial measurement units (IMUs) underwent calibration on a level and uniform surface in accordance with the manufacturer's guidelines. The IMU was affixed to the lower back of the skiers, adhering to the instructions provided by Pérez-Chirinos Buxadé et al. [50]. Measurement system axis orientation and calibration were established as shown in Figure 1.

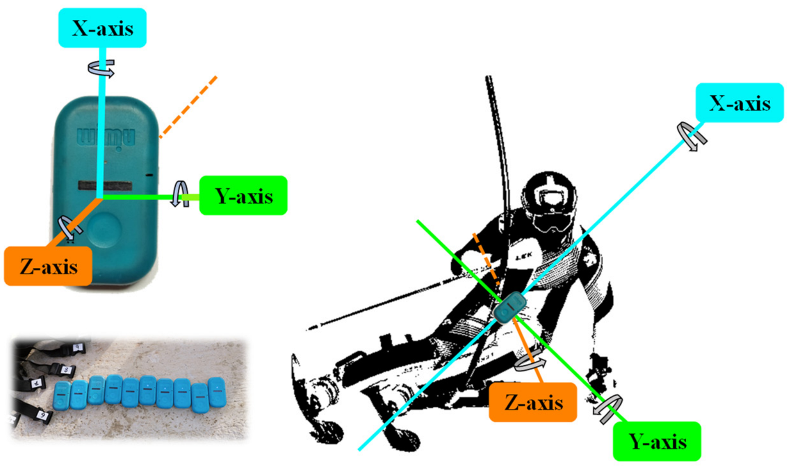

Figure 1. Calibration process and axis orientation. In the lower-left corner, the calibration procedure is depicted, with the *Z*-axis perpendicular to the surface. The *X*-axis (blue) represents the vertical axis, the *Y*-axis (green) corresponds to the lateral axis, and the *Z*-axis (orange) indicates the antero-posterior axis. The photo has been reproduced with permission from Nacho Casares/Comité Olímpico Español and has been edited by the authors.

2.2.2. Course Setting

In order to simulate different competition scenarios, four SL course settings (Figure 2) were designed in accordance with the rules set by the International Ski Federation (FIS) [51]. The slope (12° and 21°) and gate offset (4 m and 3.25 m) were varied to replicate the conditions encountered in various competition sections. For ease of reference, each course was assigned a code based on its slope and gate offset: FT (flat-turned), ST (steep-turned), FS (flat-straighter), and SS (steep-straighter). In these codes, the first letter denotes the slope, and the second letter indicates the skier's trajectory based on the gate offset. A detailed gate analysis was performed on ten consecutive gates for each SL course. Gate placements were measured precisely using tape measures to ensure consistency across all courses. For the flatter slope courses (FT, FS), the analysis focused on gates 6 to 15, excluding gate 16. Similarly, for the steeper slope courses (ST, SS), the analysis was conducted on gates 19 to 28, excluding gate 29. The initial gate consistently required a right turn (involving the left outer leg). This approach allowed for a comprehensive analysis of gate sequences while maintaining consistency in gate numbers across course configurations. Bar magnets (measuring D33 mm × 267 mm, ND35, A.C. magnets 98, Barcelona, Spain) were positioned on ten successive gates for each SL course, adhering to the guidelines provided by Pérez-Chirinos Buxadé, et al. [50]. Prior to the trial runs, skiers were given a reconnaissance run to familiarize themselves with the SL courses. They were then instructed to perform 3 to 5 runs at maximum speed for each SL course. In total, 77 runs were collected, with 22 runs conducted on the FT course, 24 runs on the ST course, 16 runs on the FS course, and 15 runs on the SS course. Runs were distributed among five skiers: skier 1 (with counts of 4 FT, 4 ST, 0 FS, 0 SS), skier 2 (n = 4, 5, 4, 4), skier 3 (n = 5, 5, 4, 4), skier 4 (n = 4, 5, 4, 4), and skier 5 (n = 5, 5, 4, 3). Notably, skier 1 faced difficulty completing descents on the FS and SS courses due to lower back pain. To synchronize the accelerometry data with the video recordings of each run for visual validation, a portable Full HD camera (Panasonic HC-V700) recording at 30 Hz was employed. Throughout the data collection process, the snow surface was maintained in hard-packed and groomed conditions. To ensure standardization, coaches and the experimental team meticulously smoothed the course prior to each run. Air temperatures progressively increased from 1.1 °C to 1.9 °C over the course of the experiment. Relative humidity slightly decreased from 71% to 61%. The maximum recorded wind speed was 6.3 km/h, blowing from a south-westerly direction, perpendicular to the course.

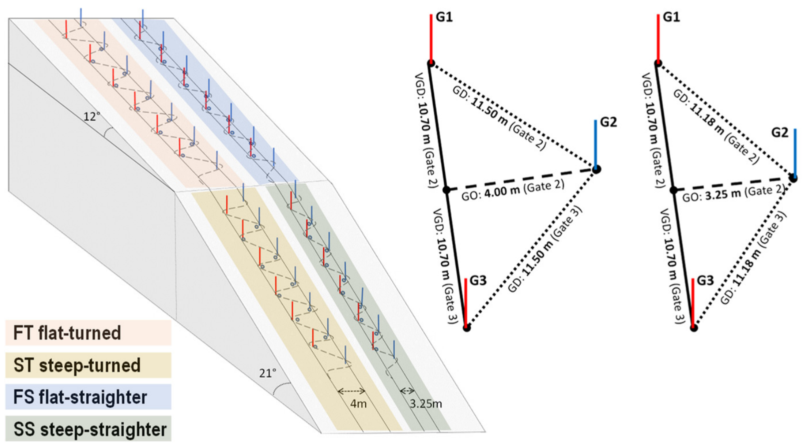

Figure 2. On-hill measurements. Flat-turned course (pink): 12° slope, 4 m gate offset (GO); flat-straighter course (blue): 12° slope, 3.25 m GO; steep-turned course (yellow): 21° slope, 4 m GO; steep-straighter course (green): 21° slope, 3.25 m GO.

Terrain characteristics, such as the slope, were determined by generating a digital elevation model using high-resolution images obtained through a drone (DJI Mavic Air, SZ DJI Technology Co., Hong Kong, China) and several photogrammetry software (Agisoft Metashape Professional® 1.5.2 version, Agisoft LLC., St. Petersburg, Russia, and ArcGIS free software version 10.3, Environmental Systems Research Institute, Inc., Redlands, CA, USA).

2.3. Data Analysis

2.3.1. Skiers' Performance

The skiers' performance was evaluated with the descent time of each run defined as the time elapsed between the first and tenth gates and was calculated through a validated Magnet-Based Timing System (M-BTS) [50].

2.3.2. Skier's MV

MV was evaluated through SampEn from the acceleration time series collected with an IMU at the lower back of the skiers. The acceleration was the magnitude of the vector formed by adding the acceleration vectors in each axis (x, y, z). Taking advantage of the synchronization of all the sensors, the peaks recorded in the magnetometer signal were used to delimit the start and end of each run [50]. The SPRO software (version, 987, Realtrack Systems, Almeria, Spain) was used to segment the acceleration signals of each run and export the data to Excel. Then, the calculation of SampEn entropy was performed according to Goldberger et al. [52], following a routine programmed in MatLab® (version R2020a, The MathWorks, Natick, MA, USA).

2.3.3. Statistical Analysis

The analysis encompassed a total of 77 runs for a sample of five skiers. In this sense, the runs from each skier were considered as repeated measures. To assess the influence of different slopes and slalom gate offsets on MV and total descent time, a one-way repeated measures analysis of variance (ANOVA) was implemented. If a significant result was obtained, post hoc comparisons were performed and *p*-values were adjusted considering Tuckey correction for multiplicity of contrasts. To show the differences more visually, bar plots depicting the mean SampEn for each SL course and the distribution of skiers were used. Additionally, the total descent times for each SL were also plotted. Additionally, a two-way ANOVA was employed to determine the significance of the explanatory variables, specifically the steepness of the slope and gate offset. Partial eta squared was used to determine the effect size as small ($\eta^2p = 0.01$), medium ($\eta^2p = 0.06$), and large ($\eta^2p = 0.14$) [53]. To assess the relationship between MV and performance, a Pearson's linear correlation coefficient was used to calculate the correlation between SampEn and total descent time. A scatter plot was created with each point representing the average of each skier's runs for each SL course. All database management tasks and statistical analyses were performed using R v4.0.4 software [54]. The significance level for all statistical tests was set at 5% ($p < 0.05$) and 10% ($p < 0.1$) [55].

3. Results

The effect of different slopes and gate offsets on MV was investigated using SampEn, as shown in Figure 3. SampEn was higher on the steeper slope (21°: ST and SS courses) than on the flatter slope (12°: FT and FS courses). Concretely, differences at the 5% level were found between ST and FS courses (ST: 0.094 ± 0.015 a.u. vs. FS: 0.074 ± 0.008 a.u.). Differences at the 10% level were found between ST and FT (ST: 0.094 ± 0.015 a.u. vs. FT: 0.082 ± 0.007 a.u.) and between SS and FS courses (SS: 0.088 ± 0.008 a.u. vs. FS: 0.074 ± 0.008 a.u.) (Table 1). Additionally, a trend was observed in which turned courses had greater SampEn values than straight courses, consistent across all comparisons, i.e., ST had higher entropy values than SS and FT had higher entropy values than FS. Figure 3 also shows the deviation of each skier from the mean SampEn in each SL course. On the FT course, skiers 1, 4, and 5 exceeded the mean entropy (0.082) by 1.83%, 1.62%, and 1.97%, respectively. The mean SampEn on the FS course was 0.074, and all skiers had a lower entropy on this run except skier 4, who had exceeded this value by 6.55%. The ST course had the highest mean SampEn (0.094), and skiers 2 and 5 had even higher entropies on this course, exceeding the mean by 8.29% and 7.85%, respectively. The mean SampEn on the SS course was 0.088, and skier 5 exceeded the mean on this SL by 4.72%.

Differences in descent times between SLs were also plotted (Figure 4). Turned courses (gate offset of 4 m: FT and ST) had the longest descent times (7.712 ± 0.245 s and 7.857 ± 0.146 s, respectively). The FS course had been the one with the shortest descent time (6.934 ± 0.168 s), followed by the SS course (7.335 ± 0.145 s). There were differences at the 5% level between all the SL courses, except for FT vs. ST courses (Table 1). Figure 4 also illustrates the deviation of each skier's descent time from the mean for each SL course. Skiers 4 and 5 consistently exceeded the mean descent time on all course settings. On the

FT course, skiers 4 and 5 were 4.39% and 1.58% slower than the mean group, respectively. Similarly, on the FS course, skier 4 exceeded the mean descent time by 3.13%, and skier 5 by 0.32%. This trend repeated on the ST course, where skier 4 exceeded the mean time by 2.30% and skier 5 by 1.64%. Finally, on the SS course, skier 4 surpassed the mean descent time by 1.78%, and skier 5 by 1.52%.

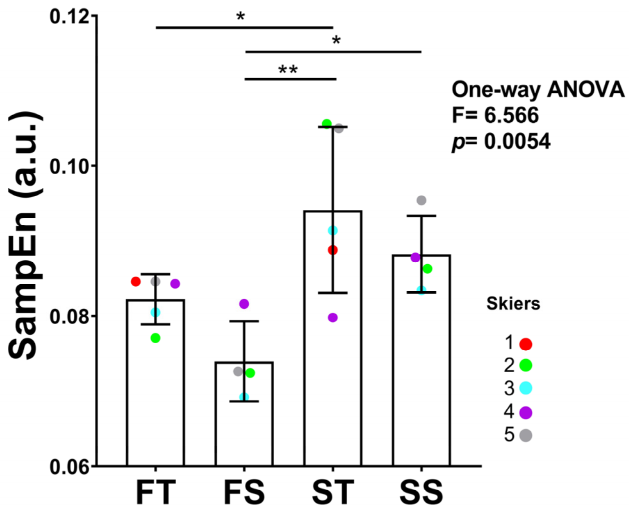

Figure 3. Bar plots illustrating the Sample entropy (SampEn) for individual skiers on each SL. The bar values indicate group means ± SD. Statistical comparisons were conducted using a one-way ANOVA for repeated measures. The course abbreviations are as follows: FT (flat-turned course), FS (flat-straighter course), ST (steep-turned course), and SS (steep-straighter course). Different colors on the graphs correspond to different skiers. Significance levels are denoted as ** $p < 0.05$ and * $p < 0.1$.

Table 1. Differences between SL course settings in SampEn and Time.

	Conditions		Mean Differences	p	95% CI	
					Lower	**Upper**
SampEn (a.u.)	FT	FS	0.008	0.339	−0.006	0.022
		ST	−0.012	0.078 *	−0.025	0.001
		SS	−0.006	0.598	−0.020	0.008
	FS	ST	−0.020	0.004 **	−0.034	−0.006
		SS	−0.014	0.055 *	−0.029	0.000
	ST	SS	0.006	0.611	−0.008	0.020
Time (s)	FT	FS	0.778	0.0001 **	0.419	1.136
		ST	−0.146	0.605	−0.484	0.192
		SS	0.377	0.038 **	0.018	0.735
	FS	ST	−0.923	<0.0001 **	−1.282	−0.565
		SS	−0.401	0.036 **	−0.779	−0.023
	ST	SS	0.522	0.004 **	0.164	0.880

FT, flat-turned course; FS, flat-straighter course; ST, steep-turned course; SS, steep-straighter course. ** $p < 0.05$; * $p < 0.1$. CI, 95% confidence interval. Comparisons used a one-way ANOVA for repeated measures.

Additionally, in Table 2, a two-way ANOVA was employed to evaluate the significance of two key explanatory variables: the steepness of the slope and the gate offset. Both had a significant impact on MV and performance of the skiers with large ($\eta^2p > 0.14$) effects. There was no interaction between the explanatory variables in any of the response variables.

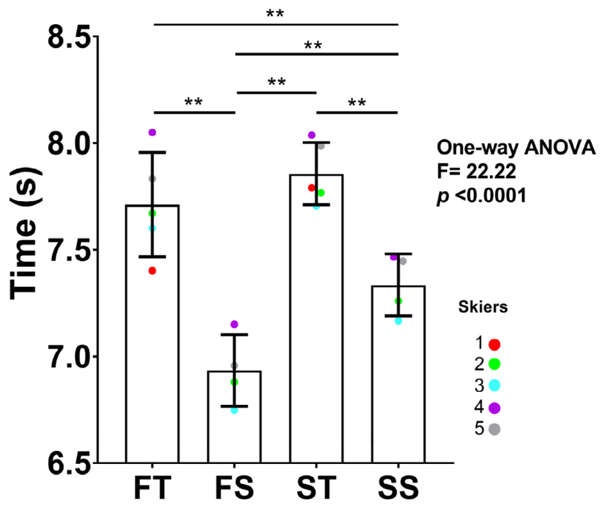

Figure 4. Bar plots illustrating the descent time for individual skiers on each SL. The bar values indicate group means ± SD. Statistical comparisons were conducted using a one-way ANOVA for repeated measures. The course abbreviations are as follows: FT (flat-turned course), FS (flat-straighter course), ST (steep-turned course), and SS (steep-straighter course). Different colors on the graphs correspond to different skiers. Significance levels are denoted as ** $p < 0.05$.

Table 2. Influence of the steepness of the slope and gate offset on SampEn and descent time.

	Steepness of the Slope (12°)		Steepness of the Slope (21°)		Steepness Effect	η^2p	Gate Offset Effect	η^2p	Steepness x Gate Offset	η^2p
	Gate Offset (4 m)	Gate Offset (3.25 m)	Gate Offset (4 m)	Gate Offset (3.25 m)						
SampEn (a.u.)	0.082 ± 0.003	0.074 ± 0.005	0.094 ± 0.011	0.088 ± 0.005	0.002 **	0.521	0.053 **	0.242	0.729	0.009
Time (s)	7.712 ± 0.245	6.934 ± 0.168	7.857 ± 0.146	7.335 ± 0.145	0.007 **	0.413	<0.0001 **	0.799	0.165	0.133

Values are means ± SD. ** $p < 0.05$. Comparisons used a two-way ANOVA. η^2p, partial eta squared.

To assess the relationship between MV and performance of elite alpine skiers, a Pearson's linear correlation coefficient was used (Figure 5). A moderate positive correlation was observed between SampEn and descent time; thus, as SampEn decreases, descent time also decreases (r = 0.587, $p < 0.01$).

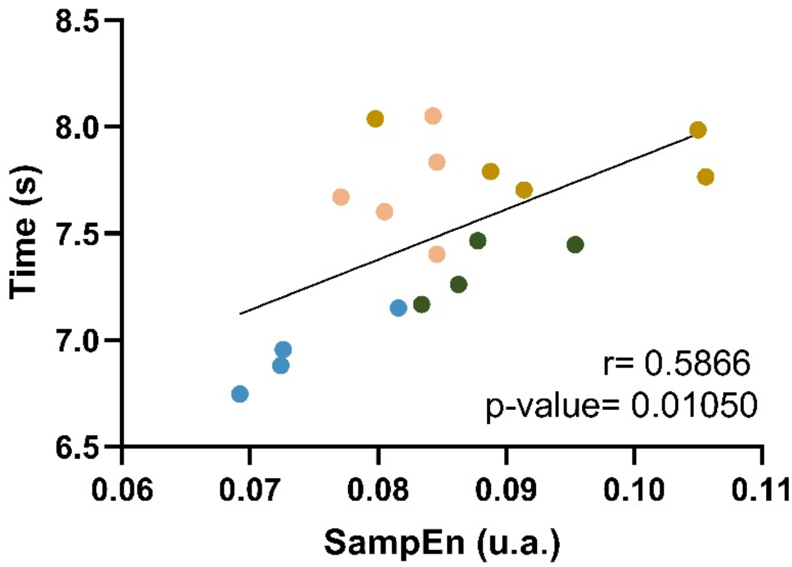

Figure 5. Correlation between descent time and SampEn. Different colors on the graphs represent the SL courses: flat-turned course (pink), flat-straighter course (blue), steep-turned course (yellow), steep-straighter course (green); r, coefficient of correlation. Significance levels are denoted as $p < 0.05$.

4. Discussion and Conclusions

This study aimed to (a) evaluate the influence of different slopes and slalom gate offsets on MV through an IMU and to (b) assess the relationship between MV and performance of elite alpine skiers. To the best of our understanding, this study represents the initial attempt to investigate the impact of terrain features, such as the steepness of the slope (flatter and steeper), and course settings, involving gate offset variations (turned and straighter courses), on skier's MV. Also, it is the first that relates skiers' performance to MV in the field of alpine skiing. The main findings corroborated the hypotheses and pointed out that (a) steep and turned courses increased skier's MV and that (b) best-performing skiers exhibited a lower MV.

The results of this study indicated that skier's MV increased with slope steepness. Thus, the steepness of the slope can be considered a conditioning factor that increases the difficulty of the task of descending an SL [56]. This is supported from a biomechanical perspective. Supej et al. [3] reported significant kinematic and kinetic variations in ski turn execution when comparing turns performed on slopes with different inclinations: 25.2° (≈47%) and 19.8° (≈36%). On the steeper slope, skiers initiated their turns earlier, with the apex located before the gate; turns were sharper with a tighter radius and lateral knee and hip angles were more pronounced, suggesting a heightened focus on controlling speed and edge engagement. The increased potential energy available on steeper slopes demands greater energy dissipation and, consequently, higher accelerations and forces [57]. This dynamic necessitates alterations in the skier's coordination patterns to maintain equilibrium throughout the turn, reducing speed across all turn phases. This observation aligns with our findings since skiers exhibited the longest descent times on steeper slopes, further supporting the notion of increased task difficulty. Furthermore, slope inclination has been shown to induce a greater activation of lower limb muscles [58], which play a crucial role in powering turns and maintaining stability. These findings collectively indicate that slope variations significantly modulate task difficulty, prompting skiers to adapt their movement patterns and muscle activation strategies to successfully navigate the varying terrain.

Concerning course settings, changing the distance between gates modifies the characteristics of the course and, therefore, the trajectories that skiers must follow. Although the FIS establishes regulations [51], considerable flexibility remains in determining course setting features. For instance, manipulating the horizontal distance between gates, known as the gate offset, can alter the turning demands of the course. In the present study, two different gate offsets (3.25 m and 4 m) were set, and it was observed that there was a tendency for skier MV to increase with the gate offset, particularly on courses with higher turning demands (FT and ST). These findings align with those of previous biomechanical studies that have examined the impact of increasing the gate offset in giant slalom and super-giant slalom [59,60]. Spörri et al. [59] demonstrated that increasing the horizontal gate offset in giant slalom led to a decrease in speed during the steering out of the turn, a prolonged centripetal force duration, which significantly doubled at turn completion, and a predominance of backward skier positions (weight shifted toward the ski tails) and lateral inclinations (toward turn, inwards). These factors contributed to the adoption of more critical postures that may heighten the risk of out-of-balance situations. Notably, high standard deviations of speed, centripetal force, and front–back positions were observed at the end of the turn on the more turned course, indicating that skiers exhibited increased variability in their movement patterns when the horizontal gate offset was widened. Similarly, Gilgien et al. [60] reported that increasing the horizontal gate offset in super-giant slalom reduced turn speed, forced skiers to execute sharper turns with a decreased radius of curvature, and increased peak snow reaction forces and their duration (greater impulse) [61,62], consequently increasing physical exertion and fatigue [7]. In this context, studies investigating the impact of the horizontal gate offset on kinematic and kinetic variables have primarily focused on giant slalom and super-giant events [59,60,63]. Limited research has been conducted on the effects of the gate offset in slalom [8,47]. Pérez-Chirinos Buxadé et al. [47] reported a decrease in turn initiation time and an increase in turn steering time with an increasing gate offset. Reid [8] modified both the gate offset and distance between gates, making it difficult to isolate the influence of each factor on ski turn biomechanics. While the distance between gates is the distance regulated by the FIS [51] and the most practical for coaches to use on a daily basis, it is linearly independent of the gate offset. Therefore, the gate offset and vertical distances accurately represent course configuration, although they are also the most complex to determine in practice [64].

The present study also investigated the relationship between MV and performance and it was found that skiers that obtained better descent times, hereinafter skilled skiers, had a lower MV than skiers with greater descent times. MV would be expected to be lower in the skilled skiers, due to their greater ability to control the degrees of freedom of the task [21,23,27]. Learning is explained through synaptic reorganization, which makes movement patterns more stable and reduces MV [21,27]. As a result, experienced athletes can reproduce the same technical gesture more similarly, even when circumstances change [10,11,17,24,25]. An example of this is a study investigating the MV of volleyball players during the spike. The MV of the spike decreased as players progressed from the youth to the cadet category, demonstrating a reduction in MV among more experienced athletes. Similarly, spikers, who use this technical gesture most frequently, exhibited a lower MV than other players [65]. Another similar example to highlight this idea is that of Fernández-Valdés et al. [11], who conducted a study in a rugby league to examine changes in MV between positions (forwards vs. backs) during cumulative tackle event training. Their findings revealed that forwards consistently exhibited lower MV values compared to backs across all training blocks. Previous research suggests that MV can be reduced by various factors including practice and experience [66]. Given that forwards engage in more collisions throughout a match, this suggests that they may possess a heightened ability to adapt to tackle actions effectively. Additionally, the study observed a progressive reduction in MV with increasing tackle events, particularly among backs and during the defensive role.

The closest approach to this idea in the field of alpine skiing was a study conducted by Müller et al. in 1998 [67], in which the turn technique was compared between expert and intermediate skiers. They found that intermediate skiers had much higher standard deviations for all biomechanical variables analyzed, while they were reduced in experts. Along the same lines, Yamagiwa et al. [48] also analyzed variability during the skiers' descents. In this case, they developed a simple system based on a single IMU mounted on the trunk of a skier to assess skiing quality based on turning tempo (turn frequency). The algorithm assessed the turning tempo during a run to differentiate between high- and low-skill skiers. They found that high-skill skiers were those who showed greater regularity, demonstrating the ability to maintain a constant tempo during descents. In contrast, the low-skill skiers had greater tempo variability, who could be considered as athletes who are improving their motor adaptations, as they explore more of the task space [27]. Unlike previous research that employed linear statistical measures such as standard deviation and tempo calculation, this study introduces non-linear tools like SampEn in the field of alpine skiing, offering a novel approach to explore the nature of human movement and its connection to coordination development [18]. In this sense, a skier's MV must be understood and perceived during the execution of the task, within the other biomechanical variables of the process [68].

The conclusion of this study is that determining MV through entropy can be used to assess the skier's ability to adapt to changing conditions. Introducing variations in on-snow training, such as different slopes and gate offsets, can increase MV, suggesting that skiers are adapting to the new stimuli. Best-performing skiers exhibit lower MV, indicating greater consistency of movement patterns. Skiers with higher MV may be in the process of developing their motor adaptations, as they explore new movement options.

5. Practical Application

The availability of good training conditions (snow) will determine the number of on-snow training days in a competitive season. A study conducted in 2018 on the training of Olympic skiers [2] found that, in a competitive season of 130–150 days, the effective training time does not exceed 9.2 h in the case of technical disciplines (slalom and giant slalom), and does not exceed 7.1 h in the case of speed disciplines (super-giant slalom and downhill). This is due to the nature of the sport, in which descents have a very short duration (between 30 and 60 s), what represents a 6 min training for a morning's training where an average of eight runs are made. In this context, it is necessary to carry out a thorough analysis of the skiers' coordinative level to make informed decisions about which aspects need to be prioritized in on-snow training. To optimize this training time, it is necessary to know which stimuli represent a greater challenge for our group of skiers (i.e., higher entropies) to give them preference in our training. As an example, in the current study, with the use of a single IMU device placed on the lower back of the skiers, it is possible to analyze that the FT course is more challenging for skiers 1, 4, and 5, as they are the skiers who have presented an above-average entropy value. In the same way, the FS course has been more challenging for skier 4. The ST course has been more difficult for skiers 2 and 5 and the SS course has been a greater challenge for skier 5. This study demonstrates that if skiers consistently train under the same conditions (slopes and course settings), their coordination may stagnate. It is essential to carry out tests to obtain objective data on our skiers' coordination level to propose sufficiently stimulating training sessions and to optimize the very limited training time available on the slopes.

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Institutional Review Board Statement: This study was conducted according to the guidelines of the Declaration of Helsinki and approved by Ethics Committee for Clinical Sport Research of Catalonia (Study Number: 27/CEICGC/2020).

Informed Consent Statement: Written informed consent was obtained from all subjects involved in this study.

Data Availability Statement: The raw data supporting the conclusions of this article will be made available by the authors on request.

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Concurrent Sprint Swimming Interval and Dryland Training: Performance and Biomechanical Variable Changes within a Mesocycle

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Abstract: The aim of this study was to examine the effects of concurrent dryland and sprint swimming interval training (SIT), and of SIT only, on swimmers' performance and biomechanical variables before, during, and following 6 weeks of training. Twenty-four swimmers (age: 16.5 ± 2.9 years) were assigned to three groups of equal performance level and applied concurrent dryland and SIT three times per week, as follows: (i) maximum strength (three sets × four repetitions, load 90% of one-repetition maximum) [1RM]) prior to SIT (group: G-MS); (ii) muscular endurance (2 sets × 20 repetitions, load 55% of 1RM) prior to SIT (group: G-ME); and (iii) SIT only (consisting of 2 series of 4 × 50 m sprints (group: G-CON)). Performance time, stroke rate (SR), stroke length (SL), and stroke index (SI) were measured during 4 × 50 m sprints. For pre- vs. post-performance time, SR, SL, and SI were similar between groups ($p > 0.05$). SR increased in G-MS and G-ME in week 6 vs. week 1 ($p = 0.02$), while SL and SI were similar between groups ($p > 0.05$). Concurrent dryland compared with sprint interval swimming training on the same day may progressively increase SR within a 6-week period, and all types of training improved front crawl efficiency following a mesocycle of training.

Keywords: dryland maximum strength; dryland muscular endurance; sprint swimming training; biomechanical variables

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1. Introduction

Competitive swimmers may apply maximum strength (3–5 sets, 3–5 repetitions, >85% of one-repetition maximum [1RM]) or muscular endurance in dryland training (2–4 sets, >12 repetitions, 40–60% of 1RM) prior to swimming training [1,2]. Following dryland training, the swimmers participate in swimming training to improve endurance [3] or sprint interval swimming training (SIT) with maximum effort to improve anaerobic potential [4–6]. Within a training microcycle, coaches may plan more than two dryland strength training sessions prior to in-water training, and this is regularly repeated during a mesocycle or longer periods of training [7].

There is evidence that the long-term concurrent application of dryland strength and endurance swimming training may improve performance compared with swimming training only, and this has been extensively reviewed and supported with experimental findings [2,8–10]. However, no study in swimming has examined the possible effects of concurrent dryland maximum strength or muscular endurance training and SIT on swimmers' performance. On the contrary, it has been well documented that a long-term application of SIT only may improve swimmers' performance in race distances ranging from 50 to 400 m [11–13].

In addition, alterations in biomechanical variables such as stroke rate (SR), stroke length (SL), and stroke index (SI) may explain swimming performance [14]. However, controversial findings have been reported for biomechanical variables from a combination of dryland training (80–90% of maximal load) with endurance training [10,15,16]. Previous findings indicated increments in SR and SL after 4 weeks [16] but not after 6 to 12 weeks [10,15]. However, high-intensity swimming training applied during 4 weeks of intervention increased SR during maximal efforts of the 100 and 400 m front crawl [17]. It is possible that the biomechanical alterations observed during a training period depend on the swimmers' level as well as the characteristics of the training [18,19].

To our knowledge, no study in swimming has examined the effects of concurrent dryland maximum strength or muscular endurance training and SIT applied on the same day on SR, SL, and SI during and after a training period. In addition, there is limited information available concerning the progression of swimmers' SR, SL, and SI during a training period when different concurrent training plans have been applied. The aim of this study was to examine the effects of concurrent dryland maximum strength and SIT, as well as muscular endurance and SIT, and SIT only, on swimmers' performance and biomechanical variables before, during, and following 6 weeks of training. We hypothesized that swimmers will improve their performance and biomechanical characteristics irrespective of the training combination.

2. Materials and Methods

2.1. Participants

Twenty-four national-level competitive swimmers (twelve males and twelve females) volunteered to participate in this study. All swimmers had participated in the national championship of the previous year. As inclusion criteria, each swimmer needed to meet the following: (i) be free from injury; (ii) indicate no use of medication prior to or during the training period; (iii) have at least 5 years of experience in competitive swimming; and (iv) participate in six swimming training sessions and two to three dryland sessions per week. After a thorough explanation of this study's procedures, all swimmers or their legal guardians signed a consent form accepting their participation in this study. The local institutional review board approved the experimental protocol (approved number: 1111), which was according to the Helsinki Declaration.

2.2. Study Design

A 3-group repeated-measure design was applied with pre-training and post-training period measurements. Following baseline testing, swimmers were divided into three groups of equal performance levels according to their 100 m swimming performance, and then completed a 6-week training mesocycle. Swimmers' characteristics in each group are shown in Table 1.

Table 1. Anthropometric and performance characteristics of the participants in each group.

Variables	G-MS (n = 8)	G-ME (n = 8)	G-CON (n = 8)
Age (years)	17.0 ± 2.6	15.9 ± 2.0	16.7 ± 4.2
Body mass (kg)	60.8 ± 8.0	59.4 ± 8.5	60.3 ± 12.5
Body height (cm)	170.1 ± 5.3	171.0 ± 8.2	168.5 ± 12.1
Body fat (%)	15.5 ± 4.5	15.3 ± 3.4	17.6 ± 3.6
Body mass index ($kg \cdot m^{-2}$)	20.9 ± 1.9	20.3 ± 1.9	20.8 ± 2.1
100 m front crawl performance time (s)	64.9 ± 7.4	66.3 ± 6.8	67.3 ± 7.7
WA points (100 m front crawl)	457.5 ± 95.8	425.0 ± 75.6	411.8 ± 104.9
Competitive training experience (years)	8.0 ± 1.5	7.9 ± 1.4	7.6 ± 1.7

WA: World Aquatics, G-MS: group of maximum strength, G-ME: group of muscular endurance, G-CON: control group.

During the 6-week period, swimmers of the G-MS group (n = 8) performed a maximum strength dryland training session prior to SIT. Swimmers in the G-ME group (n = 8) performed a muscular endurance dryland training session prior to SIT, while G-CON (n = 8) performed the SIT only. All groups applied the concurrent session three times per week and 20 min after the dryland session. G-CON performed easy stretching and arm-swing exercises prior to SIT during the intervention days and no dryland training was applied within the mesocycle of intervention. All the swimming training sessions were the same for all groups. Measurements were conducted during the specific preparation period of the second seasonal cycle of the year-round training plan. All tests as well as training sessions were completed at the same time of the day (17:00 to 19:00 p.m.) in a 50 m outdoor swimming pool with a water temperature of 27 °C. Ambient temperature during testing ranged between 20 and 25 °C. All SIT testing procedures during, as well as prior to and post the 6-week period were carried out by experienced and certified personnel. The experimental design of the study is shown in Figure 1.

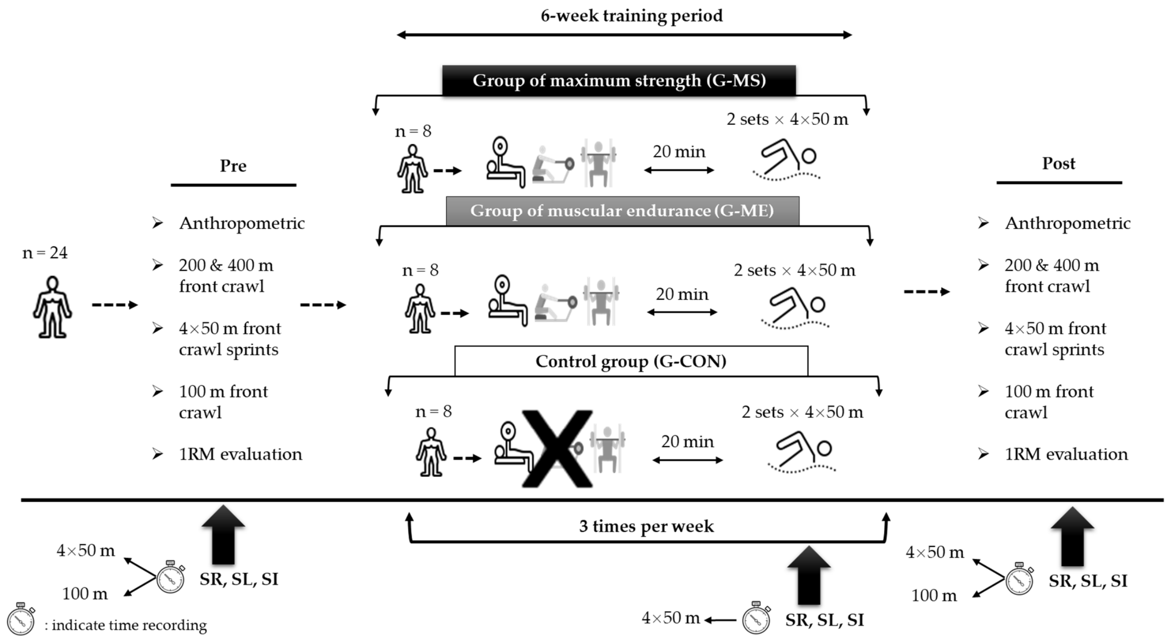

Figure 1. Experimental design of the study: 1RM: one-repetition maximum; SR: stroke rate; SL: stroke length; SI: stroke index.

2.3. Testing Procedures

All swimmers were evaluated before (pre) and after (post) the 6-week training period. On day 1, body mass and body height (Seca, Hamburg, Germany) were measured and body mass index was calculated. Body fat percentage was estimated according to Jackson and Pollock's method [20] and lean body mass (LBM) was calculated according to Boer's method [21]. On day 2, the swimmers performed 200 and 400 m front crawl tests, applying maximum effort. The recovery period between 200 and 400 m was 30 min, including a 5 to 10 min period of active recovery. From the two timed distances (200 and 400 m), the linear relationship of time vs. distance was drawn and the critical speed (CS) was determined as the slope of the regression line [3]. On day 3, the swimmers completed four repetitions of 50 m front crawl sprints (4 × 50 m) using a push-off start from within the water and starting every 2 min. The mean swimming performance time was used for the statistical analysis. Moreover, swimming time of each repetition was used to calculate the decrement score (DS) [22]. On day 4, performance time in a 100 m front crawl test with maximum effort was recorded. In all testing sessions, the SR was calculated by the time to complete 3 stroke cycles, and SL was calculated by the ratio of swimming speed to SR. SI was calculated by the product of SL and swimming speed. All biomechanical variables were measured at every 50 m during the 4 × 50 m sprints and the 100 m test and were averaged to obtain one value for each test, which was used for the statistical analysis. On day 5, the individual 1RM was evaluated in bench press (ICC = 0.99), seated pulley rowing (swimmers were allowed to move their torso during the pull; ICC = 0.98), and half squat exercises (knee angle 90°; ICC = 0.99) using standard procedures [23]. Prior to each swimming testing procedure, the swimmers performed an 800 m standardized warm-up (400 m slow front crawl swimming, 4 × 50 m front crawl drills, and 4 × 50 m front crawl swimming with progressively increasing speed).

2.4. Training Content and Testing

Both maximum strength and muscular endurance dryland sessions consisted of sit-ups and back extension exercises (3 sets × 15 repetitions and 30 s resting interval) and three resistance training exercises that have been previously included in dryland sessions for competitive swimmers [15]. The dryland sessions' characteristics are shown in Figure 2. The training volume of both dryland sessions were equalized by manipulating the number of sets, repetitions, load/intensity, and movement tempo as it is shown in Equation (1) [24]:

$$\text{Trainingvolume} = \text{Sets} \times \text{Repetitions} \times \%1\text{RM} \times \text{MT} \tag{1}$$

where %1RM (repetition maximum) is the training load/intensity and MT is the movement tempo during a repetition in bench press, seated pulley rowing, or half squat.

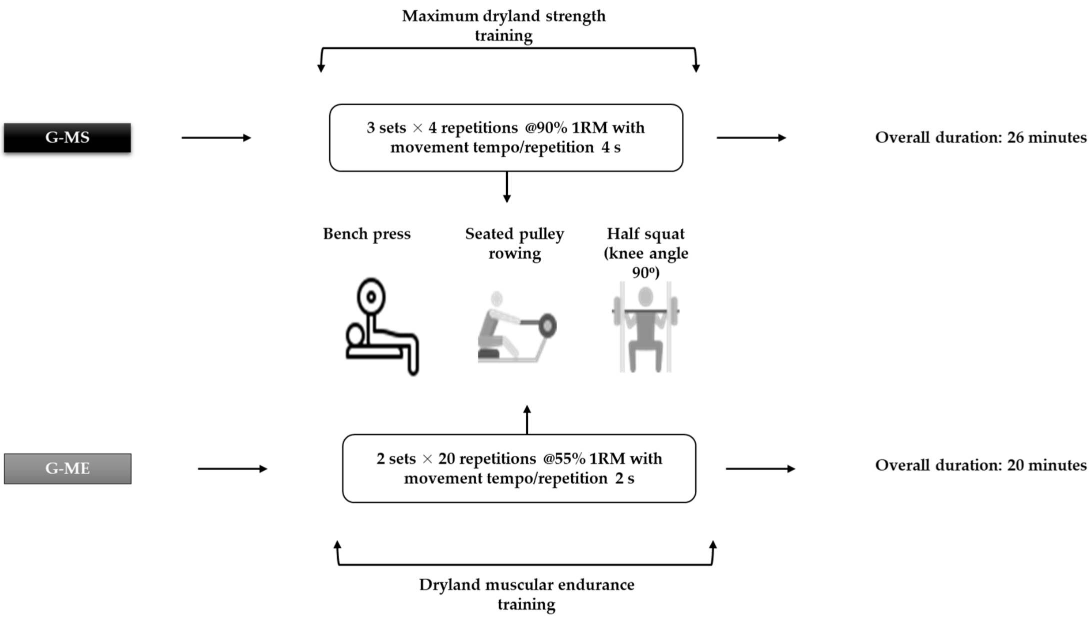

Figure 2. A graphic representation of the maximum strength and muscular endurance dryland training sessions applied by the swimmers in the group of maximum strength (G-MS) and in the group of muscular endurance (G-ME) prior to sprint swimming training during the 6-week training period; 1RM: one-repetition maximum.

2.5. Sprint Swimming Interval Training and Decrement Score

The SIT session was the same in all experimental groups and it was applied after an 800 m standardized warm-up (400 m slow front crawl swimming, 4 × 50 m front crawl drills, and 4 × 50 m front crawl swimming with progressively increasing speed), including two sets of 4 repetitions for 50 m sprints. The first set was performed in front crawl and the second set in the personally preferred swimming stroke. Both sets were applied using a push-off start and starting every 2 min. A five-minute passive resting interval was allowed between the two sets. The daily training volume during the days when SIT applied, was 3000 m and the swimming training during the remaining 3 days of the week ranged from 3200 to 5000 m. The training intensity was adjusted according to CS and applied in three training zones: (i) zone 1 corresponding to 95–97% of CS, (ii) zone 2 corresponding to 99–101% of CS, and (iii) zone 3 corresponding to 104–107% of CS [3]. Performance time of the first 4 × 50 m front crawl training set in the first SIT session of each week was recorded by experienced timekeepers and the mean time as well as the calculated DS were used for the statistical analysis. Moreover, the SR, SL, and SI were calculated during the first set of 4 × 50 m sprints and the mean values from each set in each week were used for the statistical analysis. The internal training load of daily swimming training was estimated by calculating the session rating of perceived exertions (session-RPE) and using a 10-point Borg scale [25]. The swimming training volume was recorded daily and was stored for subsequent analysis.

2.6. Statistical Analysis

Normal distribution of the data was tested using Kolmogorov–Smirnov test and sphericity was verified using a Mauchly test. When the assumption of sphericity was not met, the significance of *F* ratios was adjusted according to the Greenhouse–Geisser correction. Analysis of variance on repeated measures in two factors (3 groups × time points) was used for all dependent variables (anthropometric characteristics, performance time in the 4 × 50 m, 100 m, SR, SL, SI, and 1RM). A Tukey honest significant difference as a post hoc test was used to compare the means when significant F ratios were found. In addition, analysis of variance in two factors (3 groups × repeated measures) was used for all dependent variables as well as training volume and training load during the 6-week training period. The *Δ* values were estimated from post- to pre-measurements and from week 6 to week 1 for the performance time, the SR, SL, and SI. Furthermore, one-way analysis of variance between groups was used for percentage differences (%Δ*)*. To estimate the size of the main effects and interaction, the partial eta-squared ($\eta_p{}^2$) values from the analysis of variance were used. The $\eta_p{}^2$ was considered small if the value was ≤0.01, medium if it was ≤0.06, and large if it was ≥0.14. The $\eta_p{}^2$ for the sample size in the present study (n = 24) separated by three equal groups with sample (n = 8) resulted in a power of analysis corresponding to 0.71 [26]. Pearson correlation was used to examine relationships between variables and was qualitatively interpretated as small (r = 0.1–0.3), moderate (r = 0.3–0.5), large (r = 0.5–0.7), very large (r = 0.7–0.9), and nearly perfect (r > 0.9) [27]. The ICC using 1-way random effects was used to test the reliability. Data are presented as mean ± SD. Statistical significance was set at $p \leq 0.05$.

3. Results

3.1. Anthropometric Characteristics

Swimmers' body weight, body height, body fat, and LBM were similar between groups ($F_{(2,21)}$ = 0.86, p = 0.43, $\eta_p{}^2$ = 0.08 [medium], Table 2). In addition, the body height increased ($F_{(2,21)}$ = 8.83, p = 0.01, $\eta_p{}^2$ = 0.30 [large]), while body fat decreased in all groups ($F_{(2,21)}$ = 30.39, p = 0.01, $\eta_p{}^2$ = 0.49 [large]) after the 6-week training period (Table 2).

Table 2. Swimmers' anthropometric characteristics in pre- vs. post-training period. The group of maximum strength (G-MS), the group of muscular endurance (G-ME), and the control group (G-CON).

Variables	Time	G-MS	G-ME	G-CON
Body weight (kg)	Pre	60.8 ± 8.0	59.4 ± 8.5	60.3 ± 12.5
	Post	59.1 ± 7.5	59.5 ± 8.5	59.9 ± 11.9
	%Δ	−2.6 ± 2.1	0.2 ± 2.5	−0.2 ± 4.7
Body height (cm)	Pre	170.1 ± 5.3	170.9 ± 8.2	168.5 ± 12.0
	Post	170.3 ± 5.2 *	172.0 ± 8.9 *	169.4 ± 12.4 *
	%Δ	0.1 ± 0.2	0.6 ± 0.8	0.5 ± 0.8
Body fat (%)	Pre	15.5 ± 4.5	15.3 ± 3.4	17.6 ± 3.6
	Post	14.8 ± 4.7 *	14.7 ± 3.4 *	16.6 ± 3.4 *
	%Δ	−5.4 ± 3.8	−3.6 ± 4.3	−5.6 ± 3.9
LBM (kg)	Pre	49.2 ± 5.6	49.0 ± 6.8	48.3 ± 9.0
	Post	48.7 ± 5.2	49.4 ± 7.1	48.4 ± 9.0
	%Δ	−1.0 ± 1.0	0.9 ± 1.6	0.5 ± 2.8

%Δ: Post- vs. pre-measurements, LBM: lean body mass; * p < 0.05, post- vs. pre-measurements.

3.2. Training Load and Training Volume

The mean training volume was similar among G-MS, G-ME, and G-CON ($F_{(2,21)}$ = 0.88, p = 0.43, η_p^2 = 0.07 [medium]), along with training load during the 6-week intervention period ($F_{(2,21)}$ = 3.17, p = 0.06, η_p^2 = 0.23 [large], Table 3).

Table 3. Mean training volume and training load during the 6-week period for the three groups of swimmers. The group of maximum strength (G-MS), the group of muscular endurance (G-ME), and the control group (G-CON).

	G-MS	G-ME	G-CON
Training volume (m)	42.563 ± 2.613	43.794 ± 2.608	42.087 ± 2.739
Training load (a.u.)	3694 ± 185	3858 ± 232	3694 ± 250

3.3. Performance in the 4 × 50 m and 100 m Tests

Performance time of the 4 × 50 m sprints was similar among G-MS, G-ME, and G-CON (group effect, $F_{(2,21)}$ = 0.89, p = 0.42, η_p^2 = 0.07 [medium]) and decreased (indicating improvement) in all groups after the 6-week training period (effect of time, $F_{(1,2)}$ = 11.86, p = 0.01, η_p^2 = 0.36 [large], Figure 3a). Accordingly, the calculated DS during the 4 × 50 m sprints was similar between groups ($F_{(2,21)}$ = 0.17, p = 0.85, η_p^2 = 0.01 [small]) and decreased following the 6-week training period (G-MS, pre: 2.5 ± 1.9 vs. post: 1.6 ± 0.1%, G-ME, pre: 2.7 ± 1.7 vs. post: 1.5 ± 0.1%, G-CON, pre: 3.0 ± 1.5 vs. post: 1.5 ± 0.1%, p = 0.01).

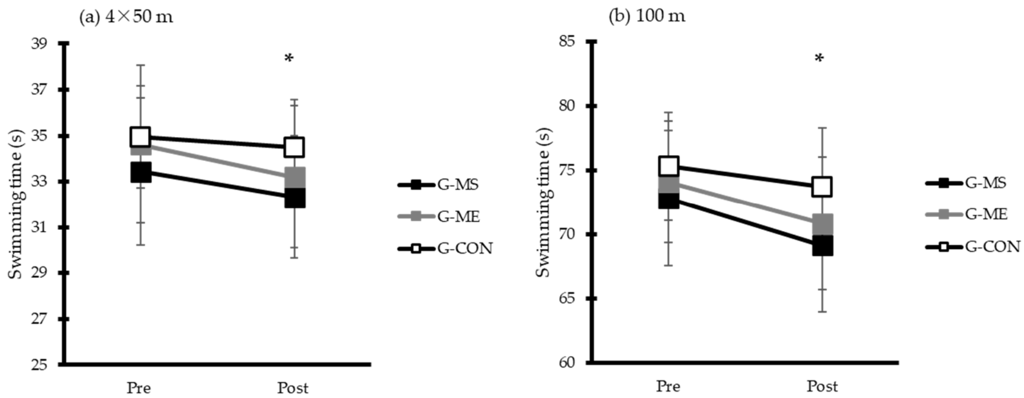

Figure 3. Performance time changes in the 4 × 50 m sprints (panel (**a**)) and 100 m (panel (**b**)) prior to and post the 6-week training period for the three groups of swimmers participating in the study. G-MS: group of maximum strength, G-ME: group of muscular endurance, and G-CON: control group. * $p < 0.05$, between pre- and post-measurements.

Performance time in the 100 m test was similar between groups ($F_{(2,21)}$ = 1.11, p = 0.35, η_p^2 = 0.09 [medium]) and decreased (indicating improvement) after the 6-week training period (G-MS: 5.0 ± 2.8%, G-ME: 4.4 ± 2.2%, and G-CON: 2.1 ± 3.2%, p = 0.01, Figure 3b).

3.4. Biomechanical Variables in the 4 × 50 m and 100 m Tests

The biomechanical variables during the 4 × 50 m test (SR, SL, SI) were similar between groups (p = 0.39–0.49). Moreover, SR and SL were unchanged in all groups (p = 0.11–0.45, Table 4); however, the SI increased in the G-MS, G-ME, and G-CON after the 6-week training period ($F_{(1,21)}$ = 10.03, p = 0.01, η_p^2 = 0.32 [large], Table 4). All of the biomechanical variables during the 100 m test were similar between groups (effect of group, $F_{(2,21)}$ = 0.47, p = 0.63, η_p^2 = 0.04 [medium]). The SR increased in all groups (G-MS: 8.7 ± 14.8%, G-ME: 5.3 ± 3.6%, and G-CON: 1.9 ± 3.2%, p = 0.01) but the SL and SI were unchanged in all groups after the 6-week training period (p = 0.25–0.57, Table 4).

Table 4. The biomechanical variable changes in the 4 × 50 sprints and 100 m front crawl test prior to and post the 6-week training mesocycle for the three groups of swimmers participating in the study. G-MS: group of maximum strength, G-ME: group of muscular endurance, and G-CON: control group.

		4 × 50 m sprints		
Variables	Time Points of measurment	G-MS	G-ME	G-CON
SR (cycles·min^{-1})	Pre	43.06 ± 3.98	42.81 ± 5.61	41.10 ± 3.17
	Post	42.55 ± 2.23	44.75 ± 5.57	41.00 ± 4.86
	%Δ	−0.61 ± 8.68	4.69 ± 5.48	−0.49 ± 5.77
SL (m·cycle^{-1})	Pre	2.11 ± 0.18	2.06 ± 0.20	2.11 ± 0.18
	Post	2.20 ± 0.15	2.05 ± 0.18	2.15 ± 0.20
	%Δ	4.52 ± 6.94	−0.17 ± 5.62	1.98 ± 4.02
SI (m^2·s^{-1}·cycle^{-1})	Pre	3.19 ± 0.52	3.00 ± 0.41	3.04 ± 0.39
	Post	3.43 ± 0.49 *	3.11 ± 0.38 *	3.12 ± 0.28 *
	%Δ	8.03 ± 8.31	4.03 ± 7.56	3.41 ± 7.02
		100 m test		
Variables	Time points of measurement	G-MS	G-ME	G-CON
SR (cycles·min^{-1})	Pre	38.81 ± 3.38	39.66 ± 5.28	38.47 ± 3.86
	Post	41.82 ± 2.78 *	41.72 ± 5.29 *	39.23 ± 4.64 *
	%Δ	8.67 ± 14.80	5.32 ± 3.58	1.90 ± 5.19
SL (m·cycle^{-1})	Pre	2.15 ± 0.33	2.07 ± 0.23	2.09 ± 0.19
	Post	2.09 ± 0.16	2.06 ± 0.20	2.10 ± 0.24
	%Δ	−1.71 ± 11.70	−0.56 ± 4.12	0.58 ± 6.03
SI (m^2·s^{-1}·cycle^{-1})	Pre	3.00 ± 0.69	2.80 ± 0.33	2.79 ± 0.19
	Post	3.04 ± 0.40	2.91 ± 0.31	2.86 ± 0.39
	%Δ	3.68 ± 13.90	4.09 ± 6.22	2.96 ± 8.65

%*Δ*: Post- vs. pre-measurements, SR: stroke rate, SL: stroke length, SI: stroke index; * $p < 0.05$, post vs. pre-measurements.

The %*Δ* of performance time was negatively correlated with %*Δ* of SI in G-ME and G-CON ($r = -0.75$ and $r = -0.82$, respectively, $p < 0.05$, Figure 4), while no correlation was observed in G-MS ($r = -0.51$, $p > 0.05$). Moreover, in G-CON, the %*Δ* of performance time was negatively correlated with %*Δ* of SR ($r = -0.74$, $p < 0.05$), while no correlation was observed in G-ME and G-MS ($r = -0.13$, r = 0.61, respectively, $p > 0.05$). Moreover, no correlation was observed in %Δ of performance time with the %*Δ* in the SL of all groups ($r = -0.55$–0.14, $p > 0.05$, Figure 4).

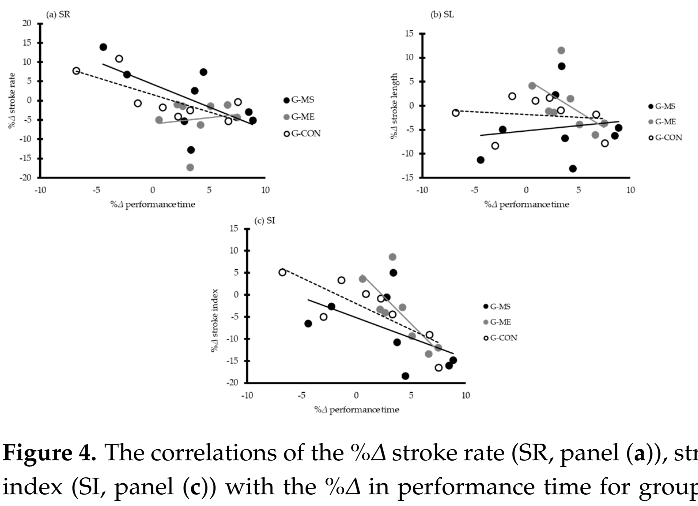

Figure 4. The correlations of the %*Δ* stroke rate (SR, panel (**a**)), stroke length (SL, panel (**b**)), stroke index (SI, panel (**c**)) with the %*Δ* in performance time for groups of maximum strength (G-MS), muscular endurance (G-ME), and control group (G-CON).

3.5. Performance and Biomechanical Variable Percentage Changes in Post- vs. Pre-Measurements

Performance time percentage changes, as well as the corresponding changes on the SR, SL, and SI measured in the 4 × 50 m sprints post and prior to the 6-week mesocycle, were not different between groups (performance time, $F_{(2,21)} = 0.97$, $p = 0.39$, $\eta_p^2 = 0.09$ [medium], biomechanical variables, $p > 0.05$, Table 5).

Table 5. Performance time, stroke rate (SR), stroke length (SL), and stroke index (SI) percentage changes (%Δ) in the 4 × 50 m sprints at post and prior to the 6-week training period. G-MS: group of maximum strength, G-ME: group of muscular endurance, and G-CON: control group.

Variables	G-MS	G-ME	G-CON
Performance time (%)	−3.1 ± 4.6	−4.0 ± 2.3	−1.2 ± 4.8
SR (%)	−0.6 ± 8.7	4.7 ± 5.5	−0.5 ± 5.8
SL (%)	4.5 ± 6.9	−0.2 ± 5.6	2.0 ± 4.0
SI (%)	8.0 ± 8.3	4.0 ± 7.6	3.4 ± 7.0

3.6. One-Repetition Maximum Strength

The maximum strength in bench press was similar between groups ($p = 0.43$, Table 6). However, G-MS and G-ME increased their 1RM in the seated pulley rowing ($F_{(1,2)} = 45.99$, $p = 0.01$, $\eta_p^2 = 0.69$ [large], Table 6) and the half squat ($F_{(1,2)} = 32.94$, $p = 0.01$, $\eta_p^2 = 0.61$ [large], Table 6) compared with G-CON.

Table 6. Post vs. prior to one-repetition maximum (1RM) strength in the bench press, the seated pulley rowing, and the half squat in the group of maximum strength (G-MS), the group of muscular enduarance (G-ME), and the control group (G-CON).

	Bench Press (kg)		
	G-MS	G-ME	G-CON
Pre	52.50 ± 17.11	48.13 ± 12.80	45.63 ± 14.00
Post	59.81 ± 19.94 *	55.31 ± 14.79 *	46.50 ± 14.90
%Δ	15.08 ± 18.92	16.31 ± 17.00	1.85 ± 7.80
	Seated pulley rowing (kg)		
	G-MS	G-ME	G-CON
Pre	57.50 ± 18.90	50.63 ± 9.03	53.75 ± 15.06
Post	70.44 ± 18.34 *#	66.87 ± 9.23 *#	54.69 ± 15.61
%Δ	24.42 ± 12.28	34.72 ± 25.60	1.80 ± 6.68
	Half squat (90°) kg		
	G-MS	G-ME	G-CON
Pre	73.13 ± 30.93	67.50 ± 14.39	61.25 ± 18.66
Post	90.00 ± 30.24 *#	78.75 ± 16.85 *#	60.94 ± 18.51
%Δ	25.62 ± 13.50	17.76 ± 14.85	0.01 ± 8.64

%Δ: Post- vs. pre-measurements, * $p < 0.05$; post- vs. pre-measurements, # $p < 0.05$, between groups.

3.7. Performance and Biomechanical Variables during the 6-Week Training Period

3.7.1. Performance

Performance time that was recorded in the first session of each week during the 4 × 50 m training set was similar among G-MS, G-ME, and G-CON during the 6-week training period ($F_{(2,21)} = 1.05$, $p = 0.37$, $\eta_p^2 = 0.09$ [medium], Figure 5). Furthermore, all groups decreased (indicating improvement) their performance time in the 4 × 50 m training set in weeks 4, 5, and 6 compared with week 1 ($F_{(5,105)} = 11.73$, $p = 0.01$, $\eta_p^2 = 0.36$ [large], Figure 5). In addition, the calculated DS was similar among G-MS, G-ME, and G-CON during the 6-week training period ($F_{(2,21)} = 0.09$, $p = 0.90$, $\eta_p^2 = 0.00$ [small], Figure 5). Furthermore, all groups decreased their DS in week 4 compared with weeks 1 and 2 ($F_{(5,105)} = 2.78$, $p = 0.02$, $\eta_p^2 = 0.12$ [medium], Figure 5).

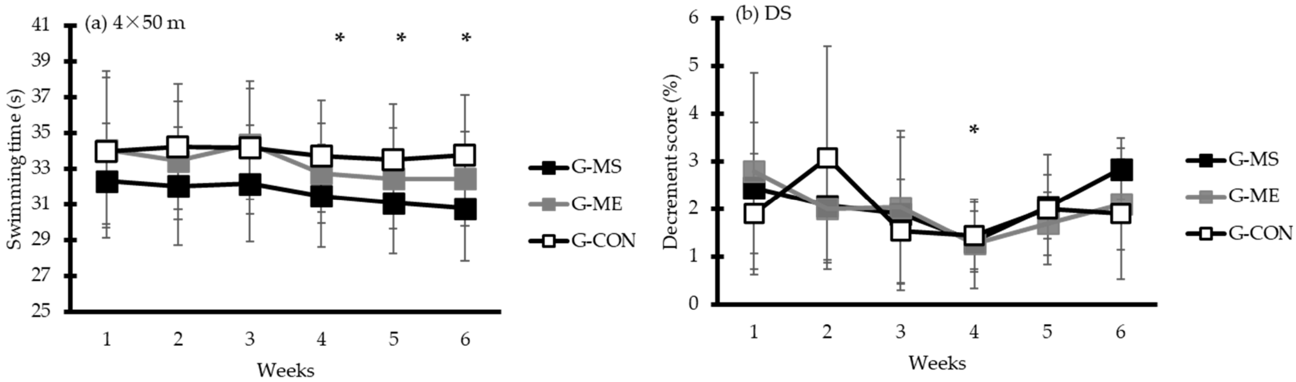

Figure 5. Performance time changes in the 4 × 50 m training set (panel (**a**)) and decrement score (DS, panel (**b**)) during the 6-week training period. G-MS: group of maximum strength, G-ME: group of muscular endurance, and G-CON: control group. * p <0.05, performance changes in weeks 4, 5, and 6 compared with week 1 (panel (**a**)), and decrement score changes in week 4 compared with weeks 1 and 2 (panel (**b**)).

3.7.2. Biomechanical Variables

The SR in G-MS and G-ME increased, while SR in G-CON decreased in week 6 compared with week 1 (group × time interaction, $F_{(10,105)} = 2.21$, $p = 0.02$, $\eta_p{}^2 = 0.18$ [large], Figure 6). Moreover, the swimmers in G-ME managed to increase their SR to a higher extent in week 6 ($F_{(14,147)} = 1.80$, $p = 0.04$, $\eta_p{}^2 = 0.15$ [large], Figure 6).

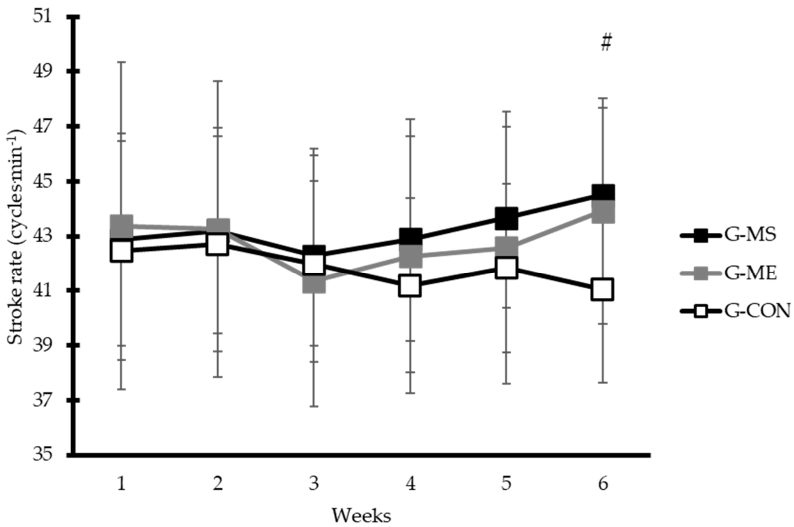

Figure 6. Stroke rate changes in the 4 × 50 m training set during the 6 weeks of training period in the three groups of swimmers participating in the study. G-MS: group of maximum strength, G-ME: group of muscular endurance, and G-CON: control group. # $p < 0.05$, between G-MS and G-ME compared with G-CON.

The SL was similar among the G-MS, G-ME, and G-CON during the 6-week training period (Figure 7). In addition, all groups increased their SL in weeks 4 and 5 compared with week 1 ($F_{(5,105)} = 5.01$, $p = 0.01$, $\eta_p{}^2 = 0.19$ [large], Figure 7). The G-MS increased the SL between week 4 and week 1 by 2.4 ± 4.5%, compared with 5.4 ± 6.6% and 3.1 ± 6.6% increments in the G-ME and G-CON, respectively (Figure 7). The G-MS increased the SL by 1.8 ± 4.1% between week 5 and week 1 compared with 5.6 ± 6.8%, and 2.1 ± 5.9% increments in the G-ME and G-CON, respectively (Figure 7).

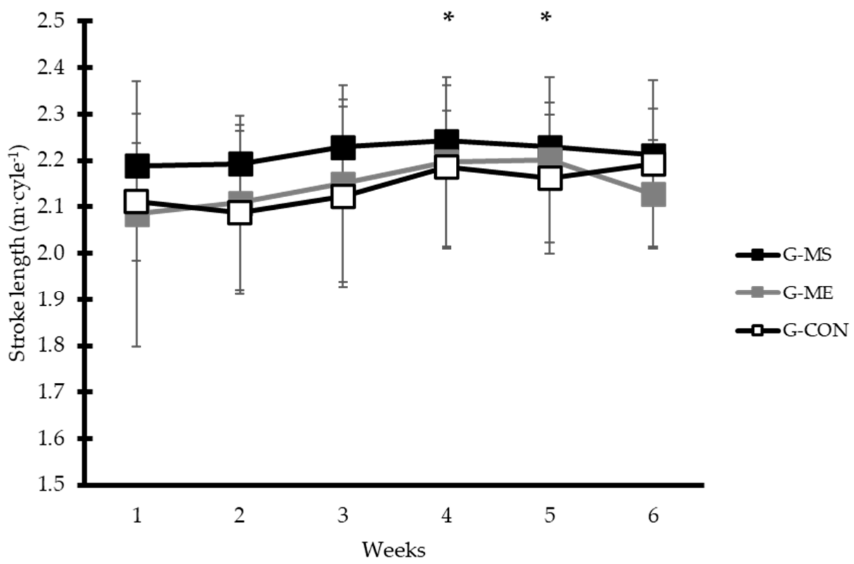

Figure 7. Stroke length changes in the 4 × 50 m training set during the 6-week training period in the three groups of swimmers participating in the study. G-MS: group of maximum strength, G-ME: group of muscular endurance, and G-CON: control group. * $p < 0.05$, weeks 4 and 5 compared with week 1 for all groups.

The SI was similar among the G-MS, G-ME, and G-CON during the 6-week training period (Figure 8). In addition, in all groups, the SI increased in weeks 4, 5, and 6 compared with weeks 1, 2, and 3 ($F_{(5,105)} = 10.03$, $p = 0.01$, $\eta_p^2 = 0.32$ [large], Figure 8).

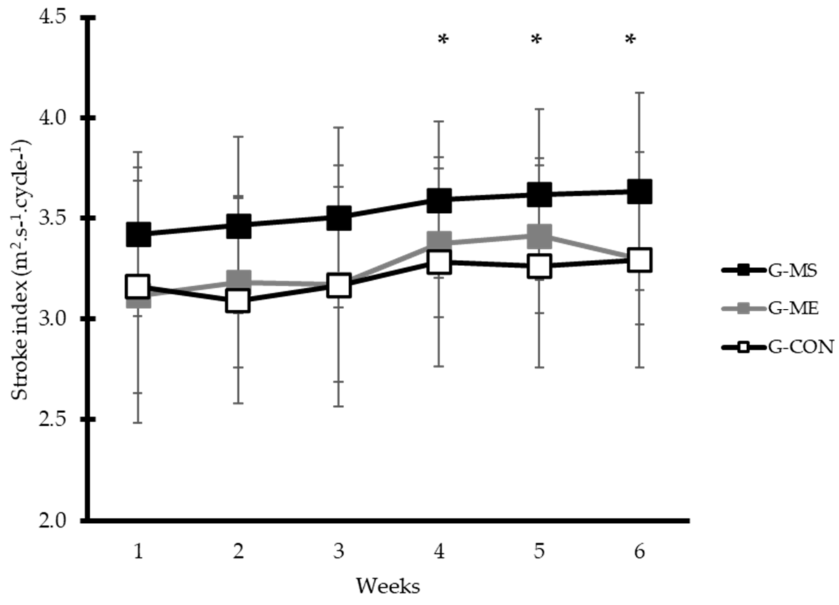

Figure 8. Stroke index changes in the 4 × 50 m training set during the 6 weeks of training period in the three groups of swimmers participating in the study. G-MS: group of maximum strength, G-ME: group of muscular endurance, and G-CON: control group * $p < 0.05$, weeks 4, 5, and 6 compared with weeks 1, 2, and 3 for all groups.

The G-MS showed higher %Δ values of the performance time compared with G-CON between week 6 and week 1 ($F_{(2,21)} = 3.66$, $p = 0.04$, $\eta_p^2 = 0.26$ [large], Table 7). However, the corresponding %Δ values of the SR, SL, and SI were similar between groups ($p > 0.05$, Table 7).

Table 7. Performance time, stroke rate (SR), stroke length (SL), and stroke index (SI) percentage differences (%Δ) in the 4 × 50 m training set in week 6 vs. week 1 for the three groups of swimmers participating in the study. G-MS: group of maximum strength, G-ME: group of muscular endurance, and G-CON: control group.

Variables	G-MS	G-ME	G-CON
performance time (%)	−4.7 ± 2.8	−4.3 ± 4.7	−0.5 ± 2.5 #
SR (%)	4.0 ± 4.3	1.9 ± 7.1	−3.1 ± 6.4
SL (%)	1.2 ± 4.3	3.2 ± 10.9	4.0 ± 7.2
SI (%)	6.3 ± 6.6	8.5 ± 17.4	4.6 ± 8.6

# $p < 0.05$, between G-MS and G-CON.

4. Discussion

The study examined the effects of concurrent maximum strength or muscular endurance dryland and SIT, as well as SIT only, on swimming performance and biomechanical variables before, during, and following a 6-week mesocycle of training. Swimming performance time showed a decrease (indicating improvement) in the 4 × 50 m sprints and the 100 m test in all groups after the training period compared with pre-training. Moreover, the SR and SL remained unchanged, while SI increased in all groups. Considering the progression of biomechanical variables within the 6 weeks of training, the present findings indicate that swimmers in the G-MS and G-ME increased their SR compared with G-CON, while the SL and SI were maintained in all groups.

4.1. Pre- vs. Post-Training Changes

Swimmers in all groups managed to improve swimming performance after the 6-week training period. Previous studies have reported that the concurrent training applied two to four times per week improved performance by 2% to 4% in distances of 25 to 400 m after 6 to 12 weeks of training [10,28,29]. However, in previous studies, the swimmers did not perform concurrent training on the same training session. In particular, they followed a concurrent application under different periodization models; for example, they applied dryland and swimming sessions separated by 7 h or on different days [10,15,16].

Moreover, SIT training only may mask swimming performance improvement during the concurrent dryland and SIT training, since such a type of training may by itself improve performance [13,15,30]. Probably, the concurrent dryland and SIT, or SIT alone, improved the aerobic and anaerobic metabolism during the 6 weeks of training period [31–33]. Furthermore, irrespective of concurrent resistance and SIT, or SIT only, the expected increments in glycolytic, oxidative enzyme activity, muscle buffering capacity, and ionic regulation is apparent [31–33]. Notwithstanding, swimmers in the current study managed to improve the performance after 18 sessions compared with a previous study, where 27 sessions were applied [13].

On the contrary, the SR and SL remained unaltered after the 6-week training period and this is in agreement with previous findings [10,15]. However, the swimmers increased their SI after 6 weeks of training in all groups. Irrespective of the training group, all swimmers were more efficient (higher SI) during the 4 × 50 m sprints after the 6-week mesocycle [14]. Possibly, neuromuscular and mechanical adaptations occur after the application of the concurrent maximum strength or muscular endurance dryland training and SIT, or SIT only. It is well known that the SI is a biomechanical parameter that relates with swimming speed [14] and it is possible that SI increments may be more closely connected to speed improvements.

The correlation of performance changes with SI changes indicates that the swimmers in the G-ME and G-CON who improved their SI were also able to improve performance in the 4 × 50 m training set. The increased SI with unchanged SL during the 4 × 50 m sprints may indicate that the swimmers applied more propulsive force during this set, but without affecting their swimming economy. The last was also observed in a previous study where the increased SR without a decreased SL helped in adjusting stroke mechanics and maintain economy [33]. The performance change in G-CON was correlated with SR change, indicating that swimmers in this group were able to improve SR concomitantly with their performance. It is possible that SIT during the 6-week period improved specific fitness aspects (i.e., buffering capacity, aerobic power) concomitantly with swimming efficiency [34]. However, we did not measure any physiological variables, and further studies need to be conducted for safer conclusions.

Despite the fact that no %Δ difference was found between groups (Table 5), the swimmers in G-MS and G-ME presented a 3 to 4% performance improvement compared with a ~1% decrement in G-CON, both in performance time of the 4 × 50 m sprints and the 100 m test, and in the SR, SL, and SI (see Table 5). These findings, however, indicate a trend for a beneficial effect of a concurrent application of MS or ME and SIT compared with SIT only. Moreover, G-MS and G-ME groups managed to increase the maximum strength in upper and lower body muscles, which may be translated to as a facilitated transfer of land strength gains in water, and this may persist to a subsequent mesocycle of training. However, this was not possible to be tested in the present study.

The increased muscular strength may allow swimmers to activate or recruit more muscle fibers during testing with maximum efforts as it is reflected by the higher SR during the 100 m test following the training period. This is also supported by the unchanged LBM after 6 weeks of training. Then, maximum strength gains in the G-MS and G-ME may be attributed to neural adaptations that may occur in this period of training [35–37]. However, the lack of significant findings in the biomechanical variables between groups may be explained by the fact that SIT only training may have induced similar neuromuscular adjustments [31–33].

4.2. Progression of Performance and Biomechanical Variables during the 6-Week Training Period

The swimmers in all groups improved their performance within the 4th week of training (see Figure 4). Other studies have reported similar findings, either with the concurrent training [10] or with a sprint interval training only [38]. However, the present study is the first that examined the progression of swimming performance during a training period and including different dryland training content. Possibly, these types of training and the specific characteristics (intensity, rest, duration) facilitated cardiovascular and neural adaptations in a short period of 4 weeks of swimming training [38]. Despite the fact that we did not measure any physiological variables in the current study, we found that DS (indicating fatigability) was similar between groups throughout the 6 weeks of training period. In addition, the calculated DS was decreased in all groups in the 4th compared with 1st and 2nd weeks. It is possible that physiological and neural adjustments occurred in the 4th week in agreement with performance improvements at the same time point of intervention (see Figure 4). This finding in the current study may indicate that the swimmers perceived less effort (decreased DS) during the 4 × 50 m training set, because of the training-induced adaptations (see Figure 4).

It has been shown that the biomechanical variables may explain any performance progression within this short period of training [39]. We found that all groups increased their SL and SI during the 6-week training period. These SL and SI increments may be connected with strength gains and related variables such as motor unit recruitment [37]. In addition, the SL and SI increased (see Figures 6 and 7) while DS was decreased (see Figure 4) at week 4 in all groups. It is likely that swimmers in all groups adjusted their technique to be more efficient (decreased DS) during the 4 × 50 m training set. However, the G-MS and G-ME groups increased their SR during the 4 × 50 m training set. This may indicate that the concurrent application of MS or ME with the SIT session facilitated a progressive increase in muscle strength and adequate neural adaptations, increasing SR and SL. However, any fatigue induced by MS and ME sessions may have forced swimmers to adjust to a higher SR during the subsequent swimming training session [40]. Thus, the increase in SR in the MS and ME groups in week 6 may indicate accumulated fatigue.

Whatever the case, swimmers in each group may have applied different adjustments in biomechanical variables that progressively increased the speed in the MS and ME groups.

Possibly, the concurrent application of dryland training with the characteristics of maximum strength or muscular endurance and SIT is a promising type of training compared with SIT only, in improving the biomechanical variables in swimmers. Then, swimming coaches may construct a training session which includes the concurrent MS and SIT or ME and SIT. There are some limitations of the present study that should be mentioned. Both male and female swimmers participated in the study and the SIT session included short duration efforts and resting intervals (approximately 35 s and 90 s, respectively) between each 50 m sprint.

5. Conclusions

The swimmers improved performance irrespective of the training intervention during and following the training period. In addition, the swimmers in all groups increased the SL and SI during the 6-week training period. This finding reflects a better efficiency during the 4×50 m sprint interval training set. However, only G-MS and G-ME groups improved their SR in the last week of training period compared with G-CON. It is likely that dryland training when applied concurrently with swimming SIT facilitates an improvement in biomechanical variables. The swimmers may perform maximum strength or muscle endurance dryland training concurrently with swimming SIT. Such an approach is equally effective in performance enhancement as SIT alone. However, the concurrent application may be more promising in enhancing biomechanical variables and progressively increase stroke efficiency in swimmers.

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Acute Effect of Velocity-Based Resistance Training on Subsequent Endurance Running Performance: Volume and Intensity Relevance

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Abstract: This study aimed to compare the acute effect of four back squat velocity-based training (VBT) protocols in terms of intensity (60% vs. 80% of the one repetition maximum [1RM]) and volume (10% vs. 30% threshold for velocity loss in the set) on the maximal aerobic speed (MAS) estimated from a running track test (RTT) in recreationally trained young adult men and women. Twenty participants (eleven men and nine women) undertook five randomized protocols in separate occasions: (i) RTT alone (control condition); (ii) VBT with 60% 1RM and a 10% velocity loss followed by RTT ($VBT_{60–10}$ + RTT); (iii) VBT with 60% 1RM and a 30% velocity loss followed by RTT ($VBT_{60–30}$ + RTT); (iv) VBT with 80% 1RM and 10% velocity loss followed by RTT ($VBT_{80–10}$ + RTT); (v) VBT with 80% 1RM and 30% velocity loss followed by RTT ($VBT_{80–30}$ + RTT). All VBT protocols involved three sets with three minutes of rest. The MAS was higher for RTT (control) than $VBT_{60–30}$ + RTT ($p < 0.001$; $\Delta = 3.8\%$), $VBT_{60–10}$ + RTT ($p = 0.006$; $\Delta = 2.8\%$), $VBT_{80–10}$ + RTT ($p = 0.008$; $\Delta = 2.7\%$), and $VBT_{80–30}$ + RTT ($p = 0.019$; $\Delta = 1.9\%$). No protocol × sex interaction was noted ($p = 0.422$). Therefore, regardless of sex, MAS is acutely impaired after VBT, especially if the training sets are performed with a low relative load and a high velocity loss threshold.

Keywords: endurance training; human physical conditioning; musculoskeletal and neural physiological phenomena; resistance training

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1. Introduction

Running performance depends on the complex interaction of several factors, particularly physiological, as well as biomechanical and psychological [1,2]. From a physiological point of view, it is well documented that the ventilatory/lactate threshold, maximal oxygen uptake (VO_2max), and running economy are strong performance indicators, especially when the latter are combined, and these indicators determine the maximal aerobic speed (MAS) [2–4]. Therefore, running performance could be improved through central adaptations, but also through the ability of athletes to produce more mechanical work for a given energy cost [2,5]. It is therefore not surprising that increased running performance has been reported when resistance and endurance training are incorporated simultaneously within the same program (e.g., "*concurrent training*") [1,3,6,7]. However, if exercise variables such as intensity and volume are not adequately prescribed in a concurrent training session, resistance training-induced fatigue may acutely impair the quality of subsequent endurance training sessions and induce an interference effect on long-term cardiorespiratory adaptations in a phenomenon referred to as "*resistance training-induced suboptimization on endurance performance*" (RT-SEP) [8,9]. For example, Doma et al. [10] reported impaired running time-to-exhaustion at 110% of the second ventilatory threshold six hours after a resistance training session with heavy loads (six repetitions at ~80% of the one-repetition maximum [1RM]) compared to light loads (total work equated with 20 repetitions) in trained male runners. Relatedly, it has been suggested that heavy loads ($\geq$80% 1RM) may increase susceptibility to RT-SEP [9].

Velocity-based training (VBT) may help to assess optimization (e.g., auto-regulation) and individualization of resistance training intensity and volume according to the training readiness of athletes [11,12], thus reducing chances of RT-SEP [9]. For example, using VBT, Nájera-Ferrer et al. [13] found that compared to a moderate (20%) magnitude of velocity loss during resistance training (three full-squat sets at 60% 1RM), a high (40%) velocity loss resulted in higher metabolic (e.g., greater blood lactate, higher ventilatory equivalents) and mechanical stress (e.g., impaired vertical jump and squat velocity), as well as impaired running performance (e.g., unable to run 10 min at 90% MAS). Sánchez-Moreno et al. [14] observed higher running performance (i.e., MAS) following an eight-week concurrent training program with a moderate rather than high velocity loss (15% > 30%) in the resistance training bouts. Further, 2000 m rowing ergometer time-trial performance was compromised by greater velocity loss in the set (30% vs. 10%), but not by the loading magnitude (60% = 80% of 1RM) [15]. However, further research is needed to gain a deeper understanding of the acute effects of different concurrent VBT protocols, in terms of loading magnitude (60% vs. 80% of 1RM) and velocity loss in the set (10% vs. 30%), on running performance.

It has been shown that men reported higher velocities than women for the same %1RM during a variety of resistance training exercises and, consequently, the load-velocity relationship should be sex-specific for a better adjustment of the training intensity [16]. Similarly, it has been reported that recreationally trained men and women can achieve similar increases in strength and power performance following an eight-week VBT program with either 20% or 40% velocity loss, although some results (1RM strength and velocity attained to low/moderate loads) have indicated that strength and power gains favor using 40% rather than 20% velocity loss in women [17]. Therefore, it seems that women require a greater within-set fatigue than men to maximize strength and power development. These authors also observed that men were more susceptible to acute neuromuscular fatigue than women, but these differences in fatigability were reduced after the VBT program [18]. Likewise, Taipale et al. [19] generally observed greater fatigue in terms of decreased maximal and explosive strength in men than in women after a concurrent training session composed of multiple sets of different maximal and explosive strength exercises focused primarily on the leg extensors muscles, along with 10 min of running at ~80% of VO_2max. However, although these results are encouraging in addressing the sex gap observed in the scientific literature, there is scarce evidence on how sex could mediate the RT-SEP phenomenon, particularly for the VBT prescription variables (loading magnitude and velocity loss in the set), and its effect on MAS while running.

Therefore, this study aimed to examine the acute effect of four different VBT protocols, in terms of loading magnitude (60% vs. 80% 1RM) and velocity loss in the set (10% vs. 30%), on MAS performance estimated from a running track test (RTT) in recreationally trained men and women. We hypothesized that MAS performance would be compromised when the RTT is preceded by the different VBT protocols [9]. Specifically, greater impairment in MAS performance would be expected with (i) a high relative load along with a high velocity loss threshold in the set [10,15] and (ii) men [18,19].

2. Materials and Methods

2.1. Subjects

Twenty recreationally trained young adults, 11 men (age = 28.4 ± 6.4 years [range: 19–38]; body mass = 78.9 ± 11.2 kg; body height = 176.4 ± 6.0 cm; back squat 1RM relative to body mass = 1.8 ± 0.4 kg·kg^{-1}; VO_2max = 46.0 ± 7.8 mL·kg^{-1}·min^{-1}) and nine women (age = 23.6 ± 2.2 years [range: 21–28]; body mass = 56.1 ± 6.6 kg; body height = 161.7 ± 8.1 cm; back squat 1RM relative to body mass = 1.6 ± 0.3 kg·kg^{-1}; VO_2max = 37.2 ± 5.1 mL·kg^{-1}·min^{-1}), volunteered to participate in this study. All subjects had at least one year of resistance and endurance training experience (7.3 ± 5.9 and 10.3 ± 5.9 years for men, and 2.2 ± 1.2 and 9.3 ± 3.4 years for women, respectively) and were familiar with the back-squat and running exercises. No physical limitations, health problems, or musculoskeletal injuries that could compromise testing were reported. In addition, none of the subjects were taking drugs, medications, or dietary supplements to influence physical performance. All subjects were informed about the research purpose and procedures of the study before signing a written informed consent form. The study protocol adhered to the tenets of the Declaration of Helsinki and was approved by the Institutional Review Board.

2.2. Design

A randomized-controlled crossover design was used to compare the acute effect between control condition (i.e., RTT) and four different VBT protocols followed by the RTT ($VBT_{60–10}$ + RTT, $VBT_{60–30}$ + RTT, $VBT_{80–10}$ + RTT, and $VBT_{80–30}$ + RTT) on MAS performance between recreationally trained men and women. Subjects completed the five randomized protocols in sessions separated by 48–72 h (Figure 1). The *Test VAM-HPSS* application (version 3.3, University of Murcia, Murcia, Spain) was installed on a Samsung Galaxy A71 smartphone (Samsung, Suwon, South Korean) to estimate VO_2max and MAS during each RTT (see below for further details). Both VO_2max and MAS estimated from the RTT protocol were very similar to those observed during the laboratory test and gas exchange methods (bias = 0.2 mL·kg^{-1}·min^{-1} and <0.1 km·h$^{-1,}$ respectively [20]. Subjects were required to avoid any strenuous exercise throughout the study. All sessions were conducted at the university's running track, at the same time of the day for each subject (±3 h), and under similar environmental conditions (temperature: 6–15 °C; wind: <8 km·h^{-1}).

2.3. Procedures

Body mass and body height were measured at the beginning of the first session using a contact electrode foot-to-foot body fat analyzer system (TBF-300A; Tanita Corp of America Inc., Arlington Heights, IL, USA) and a wall-mounted stadiometer (Seca 202; Seca Ltd., Hamburg, Germany), respectively. Each protocol began with the same general warm-up, which consisted of five minutes of running at a self-selected pace, dynamic stretching, and joint mobility exercises. The specific warm-up consisted of two sets of ten air squats and five sub-maximal countermovement jumps, followed by one set of six, four, and two repetitions at 40%, 60%, and 80% of the subjects' self-perceived back squat 1RM with 3 min of inter-set rest, respectively. After warming up, subjects rested passively for three minutes before beginning each protocol (see Figure 1).

2.3.1. VBT Protocols

Two different relative loads (60% vs. 80% 1RM) and two different magnitudes of velocity loss during the set (10% vs. 30%) were used. Specifically, the configuration of the four VBT protocols was as follows: (i) 60% 1RM with a velocity loss in the set of 10% ($VBT_{60–10}$), (ii) 60% 1RM with a velocity loss in the set of 30% ($VBT_{60–30}$), (iii) 80% 1RM with a velocity loss in the set of 10% ($VBT_{80–10}$), and (iv) 80% 1RM with a velocity loss in the set of 30% ($VBT_{80–30}$). The relative load of each testing session was determined from the individualized load-velocity relationship using the specific warm-up sets and a minimal velocity threshold of 0.33 m·s^{-1} [21]. Sets were terminated when the subjects were unable to complete two consecutive repetitions above the velocity loss limit or with the full range of motion. The fastest repetition from the first set was used to define the target velocity loss limit (e.g., if the fastest velocity is 0.75 m·s^{-1}, the target velocity used to finish a set would be 0.68 m·s^{-1} for the 10% velocity loss). The same exercise (back squat), number of sets (three), and inter-set rest (three minutes) were used in all VBT protocols. A validated linear velocity transducer (T-Force system; Ergotech, Murcia, Spain) was used to automatically calculate the mean velocity and provide auditory mean velocity feedback after each repetition [22]. The VBT performance indicators were: (i) the number of repetitions completed in the set, (ii) the fastest velocity of the set, and (iii) the average velocity of the set.

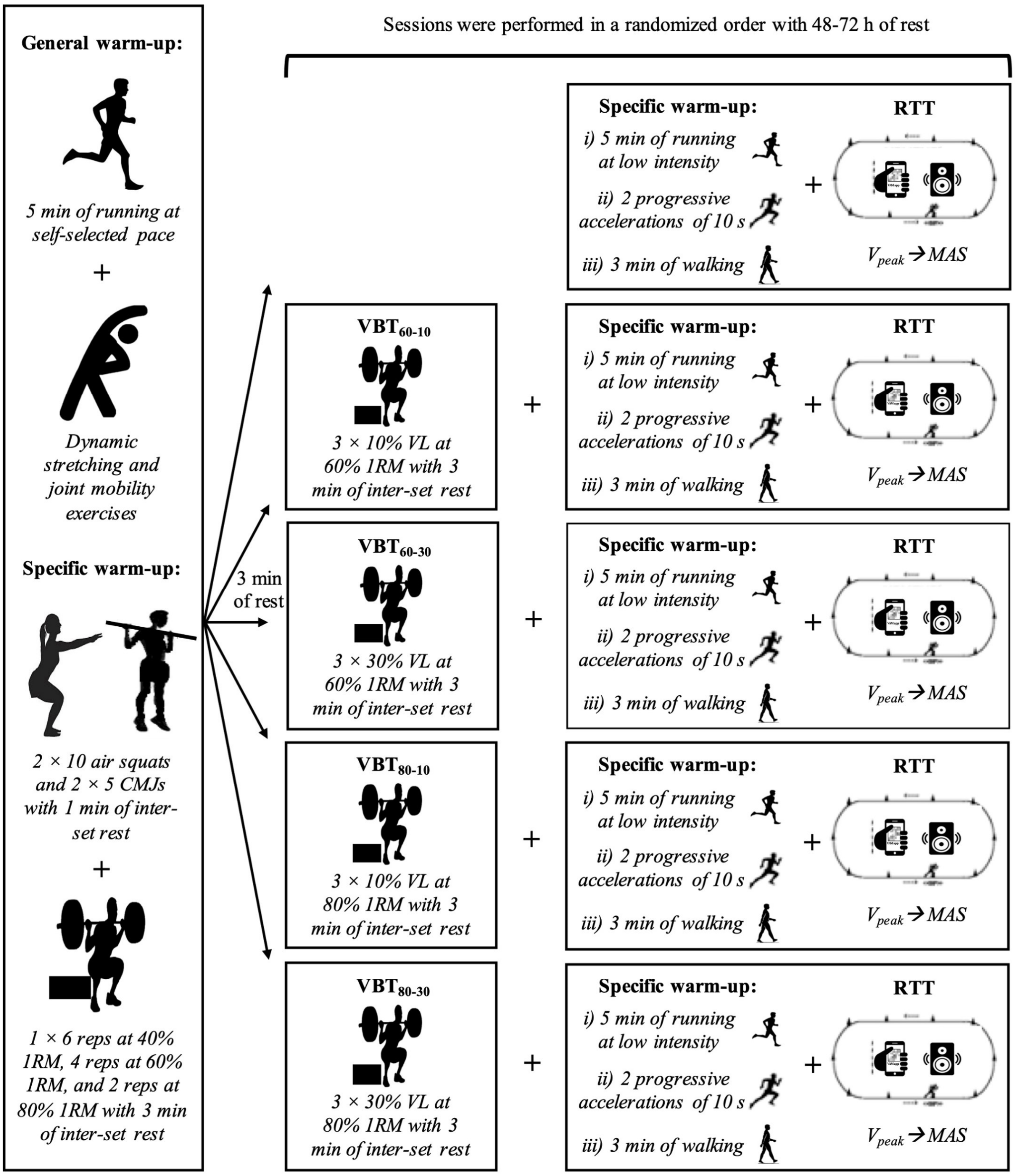

Figure 1. Overview of the experimental design. MAS, maximal aerobic speed; CMJ, countermovement jump; 1RM, one-repetition maximum; RTT, running track test; V_{peak}, peak velocity; $VBT_{60–10}$, velocity-based training (VBT) with 60% of 1RM and a velocity loss (VL) in the set of 10%; $VBT_{60–30}$, VBT with 60% of 1RM and a VL in the set of 30%; $VBT_{80–10}$, VBT with 80% of 1RM and a VL in the set of 10%; $VBT_{80–30}$, VBT with 80% of 1RM and a VL in the set of 30%.

The back-squat technique involved subjects standing with the knees and hips fully extended, feet approximately shoulder-width apart, and the barbell held across the top of the shoulders and upper back. From this position, they were required to descend in a continuous motion until their buttocks made contact with a wooden box and, immediately after, return to the initial position as fast as possible. The height of the wooden box was individually set at 90° of knee flexion with a manual goniometer (Goniómetro Rulong, Fisaude, Spain).

2.3.2. RTT Protocol

The *Test VAM-HPSS* application was used to determine running performance (MAS and VO_2max) following the manufacturer's instructions. First, the RTT protocol was selected based on the subjects' self-reported peak velocity: (i) <17.0 km·h^{-1} (>41.0 min in a 10-km race), 17.0–19.0 km·h^{-1} (36.5–41.0 min in a 10-km race), and >19.0 km·h^{-1} (<36.0 min in a 10-km race). Second, subjects completed five minutes of running at low intensity, two ten-second progressive runs, and three minutes of walking as part of the specific warm-up. Third, subjects received an auditory "*ready, set*" cue before beginning the RTT protocol with a beep signal. After pressing the start button, the stopwatch, distance, and velocity fields were launched in the *Test VAM-HPSS* application. Subjects were previously instructed to reach each cone located every 25 m around a running track while they regulated their running pace according to the beep signals. The frequency of the beep signal was automatically set according to the peak velocity selected for each RTT protocol. All auditory cues and beep signals were provided by the *Test VAM-HPSS* application connected to a loudspeaker. The RTT protocol ended when the subjects were unable to reach the cone at the time of the beep signal on two consecutive occasions, or they voluntarily decided to stop running after perceiving maximal exertion. The peak heart rate (HR) was recorded with a Polar H10 chest strap (Polar Electro Oy, Kempele, Finland) during the RTT, and the Borg's category-ratio 10 scale (CR-10) was reported after the test. The HR and CR-10 were used as maximal effort criteria [23]. The *Test VAM-HPSS* application automatically estimated the MAS and VO2max from the peak velocity achieved in each RTT [20].

2.4. Statistical Analyses

Descriptive data are presented as mean ± SDs. The Shapiro-Wilk test confirmed the normal distribution of all variables ($p > 0.05$), except for CR-10. A one-way repeated-measures analysis of variance (ANOVA) and Friedman test were used to compare peak HR and CR-10 between protocols, respectively. A mixed model ANOVA was conducted on each VBT performance indicator (numbers of repetitions, fastest velocity, and average velocity) with the protocol and set as within-subject factor and sex as between-subject factor. A mixed model ANOVA was applied to the MAS, with the protocol as a within-subject factor and sex as a between-subject factor. The Greenhouse-Geisser correction was used when Mauchly's sphericity test was violated and pairwise comparisons were identified using Bonferroni post hoc corrections. The magnitude of the differences was quantified through the standardized mean differences (Cohen's d effect size [ES]). The following scale was used to interpret the magnitude of the ES: *trivial* (<0.20), *small* (0.20–0.59), *moderate* (0.60–1.19), *large* (1.20–2.00), and *extremely large* (>2.00) [24]. All statistical analyses were performed using the software package SPSS (IBM SPSS version 25.0, Chicago, IL, USA) and statistical significance was set at an alpha level of 0.05. *Post hoc* statistical power was conducted using G*Power (Version 3.1) with an ES of 0.30 and α of 0.05, and this revealed a 0.93 statistic power.

3. Results

3.1. Descriptive Characteristics of the VBT Protocols

The main effect of protocol was significant for the number of repetitions, fastest velocity, and average velocity ($F_{(3,54)} \geq 46.9$; $p < 0.001$). A significant main effect of set was only reported for the fastest velocity ($F_{(2,36)} = 4.3$; $p = 0.021$). Finally, the protocol × sex interaction for the fastest and average velocity ($F_{(3,54)} = 5.8$ and 4.9; $p = 0.002$ and 0.005, respectively) and protocol × set interaction for the average velocity ($F_{(6,108)} = 3.9$; $p = 0.002$) also reached statistical significance (Table 1).

Table 1. Comparison of the number of repetitions, fastest velocity, and average velocity between protocols (Pr), set numbers, and sexes.

Variable	Set Number	Sex	$VBT_{60–10}$	$VBT_{60–30}$	$VBT_{80–10}$	$VBT_{80–30}$	ANOVA Main Effects	ANOVA Interactions
Number of repetitions	1	Men	9.5 ± 4.0	18.6 ± 5.4	5.9 ± 1.2	9.4 ± 3.6	Pr: $F_{(3,54)}$ = 49.6; p < 0.001	Pr × Set: $F_{(6,108)}$ = 3.2; p = 0.028
		Women	7.4 ± 2.6	17.1 ± 2.6	7.3 ± 3.1	9.6 ± 3.6	Set: $F_{(2,36)}$ = 0.9; p = 0.426	Pr × Sex: $F_{(3,54)}$ = 0.2; p = 0.906
	2	Men	9.8 ± 3.2	17.3 ± 4.9	5.9 ± 2.3	8.9 ± 3.5	Sex: $F_{(1,19)}$ = 0.1; p = 0.755	Set × Sex: $F_{(2,36)}$ = 1.8; p = 0.181
		Women	10.0 ± 6.3	19.7 ± 7.2	5.1 ± 2.7	7.7 ± 2.2		Pr × Set × Sex: $F_{(6,108)}$ = 1.8; p = 0.116
	3	Men	10.1 ± 3.9	15.4 ± 4.8	4.5 ± 1.8	7.8 ± 3.2		
		Women	12.0 ± 8.2	18.0 ± 8.2	5.1 ± 2.8	8.0 ± 3.3		
Fastest velocity ($m·s^{-1}$)	1	Men	0.76 ± 0.07	0.73 ± 0.07	0.54 ± 0.06	0.55 ± 0.07	Pr: $F_{(3,54)}$ = 80.9; p < 0.001	Pr × Set: $F_{(6,108)}$ = 1.7; p = 0.129
		Women	0.67 ± 0.03	0.67 ± 0.05	0.57 ± 0.07	0.52 ± 0.05	Set: $F_{(2,36)}$ = 4.3; p = 0.021	Pr × Sex: $F_{(3,54)}$ = 5.8; p = 0.002
	2	Men	0.77 ± 0.07	0.70 ± 0.06	0.55 ± 0.04	0.53 ± 0.07	Sex: $F_{(1,19)}$ = 4.4; p = 0.051	Set × Sex: $F_{(2,36)}$ = 0.9; p = 0.432
		Women	0.67 ± 0.07	0.66 ± 0.05	0.56 ± 0.07	0.50 ± 0.05		Pr × Set × Sex: $F_{(6,108)}$ = 0.8; p = 0.584
	3	Men	0.76 ± 0.06	0.69 ± 0.06	0.53 ± 0.06	0.53 ± 0.05		
		Women	0.67 ± 0.06	0.64 ± 0.04	0.57 ± 0.05	0.52 ± 0.04		
Average velocity ($m·s^{-1}$)	1	Men	0.70 ± 0.06	0.63 ± 0.06	0.48 ± 0.05	0.47 ± 0.06	Pr: $F_{(3,54)}$ = 84.5; p < 0.001	Pr × Set: $F_{(6,108)}$ = 3.9; p = 0.002
		Women	0.62 ± 0.02	0.59 ± 0.05	0.52 ± 0.07	0.45 ± 0.04	Set: $F_{(2,36)}$ = 1.4; p = 0.260	Pr × Sex: $F_{(3,54)}$ = 4.9; p = 0.005
	2	Men	0.71 ± 0.06	0.62 ± 0.06	0.49 ± 0.04	0.45 ± 0.05	Sex: $F_{(1,19)}$ = 2.5; p = 0.132	Set × Sex: $F_{(2,36)}$ = 0.2; p = 0.791
		Women	0.63 ± 0.05	0.58 ± 0.05	0.51 ± 0.07	0.45 ± 0.04		Pr × Set × Sex: $F_{(6,108)}$ = 1.1; p = 0.348
	3	Men	0.71 ± 0.06	0.61 ± 0.06	0.48 ± 0.05	0.45 ± 0.04		
		Women	0.64 ± 0.05	0.56 ± 0.06	0.51 ± 0.06	0.46 ± 0.05		

Data are presented as means ± standard deviations. $VBT_{60–10}$, velocity-based training (VBT) protocol with 60% of one-repetition maximum (1RM) and a velocity loss in the set of 10%; $VBT_{60–30}$, VBT protocol with 60% 1RM and a velocity loss in the set of 30%; $VBT_{80–10}$, VBT protocol with 80% 1RM and a velocity loss in the set of 10%; $VBT_{80–30}$, VBT protocol with 80% 1RM and a velocity loss in the set of 30%; ANOVA, analysis of variance; F = Snedecor's F.

The pairwise comparisons revealed that (i) the number of repetitions grew higher as the relative load was reduced and the velocity loss threshold was increased ($p \leq 0.008$; ES ≥ 0.75), although no differences were reported between $VBT_{60–10}$ and $VBT_{80–30}$ ($p = 0.100$; ES = 0.23), (ii) the fastest velocity was higher for $VBT_{60–10}$ and $VBT_{60–30}$ than for $VBT_{80–10}$ and $VBT_{80–30}$ ($p < 0.001$; ES ≥ 1.49), with no significant difference for the same relative loads ($p \leq 0.096$; ES ≤ 0.58), (iii) the average velocity was higher for $VBT_{60–10}$, followed by $VBT_{60–30}$, $VBT_{80–10}$, and $VBT_{80–30}$ ($p \leq 0.001$; ES ≥ 1.04), although no differences were reported between $VBT_{80–10}$ and $VBT_{80–30}$ ($p = 0.081$; ES = 0.56), (iv) the fastest velocity was higher for the first set than the third set ($p = 0.031$; ES = 0.34), with no significant differences with respect to the second set ($p \geq 0.233$; ES ≤ 0.20), (v) the fastest velocity was significantly higher for $VBT_{60–10}$, followed by $VBT_{60–30}$, $VBT_{80–10}$, and $VBT_{80–30}$ ($p \leq 0.009$; ES ≥ 0.68), although no differences were reported between $VBT_{80–10}$ and $VBT_{80–30}$ for men ($p = 0.081$; ES = 0.02) and $VBT_{60–10}$ and $VBT_{60–30}$ for women ($p = 0.772$; ES = 0.30), (vi) the average velocity was significantly higher for $VBT_{60–10}$, followed by $VBT_{60–30}$, $VBT_{80–10}$, and $VBT_{80–30}$ ($p \leq 0.013$; ES ≥ 0.66), although no differences were reported between $VBT_{80–10}$ and $VBT_{80–30}$ for men ($p = 0.329$; ES = 0.35), and (vii) the average velocity was comparable between sets for each protocol ($p \geq 0.135$; ES ≤ 0.48), except for $VBT_{60–10}$ where it was significantly lower for the third set than the first and second sets ($p = 0.007$; ES = 0.79).

3.2. MAS Performance

No significant differences were reported for peak HR ($F_{(4,76)} = 1.7$; $p = 0.164$) and CR-10 ($\chi^2_{(4,N=20)} = 4.2$; $p = 0.378$) between protocols (RTT = 188 ± 13 bpm and 8.9 ± 0.5; $VBT_{60–10}$ + RTT = 188 ± 14 bpm and 8.7 ± 0.5; $VBT_{60–30}$ + RTT = 186 ± 10 bpm and 8.8 ± 0.6; $VBT_{80–10}$ + RTT = 190 ± 10 bpm and 8.7 ± 0.4; $VBT_{80–30}$ + RTT = 187 ± 11 bpm and 8.9 ± 0.6, respectively). A significant main effect of protocol ($F_{(4,72)} = 7.3$; $p < 0.001$) and sex ($F_{(1,18)} = 8.3$; $p = 0.010$) was observed for MAS performance. The protocol × sex interaction did not reach statistical significance ($F_{(4,72)} = 1.0$; $p = 0.422$). The main effects revealed that the MAS was significantly higher (i) for RTT than $VBT_{60–30}$ + RTT ($p < 0.001$; ES = 1.71; Δ = 3.8%), $VBT_{60–10}$ + RTT ($p = 0.006$; ES = 0.93; Δ = 2.8%), $VBT_{80–10}$ + RTT ($p = 0.008$; ES = 0.93; Δ = 2.7%), and $VBT_{80–30}$ + RTT ($p = 0.019$; ES = 0.77; Δ = 1.9%) (Figure 2), and (ii) for men than women ($p = 0.010$; ES = 1.34; Δ = 13.0%).

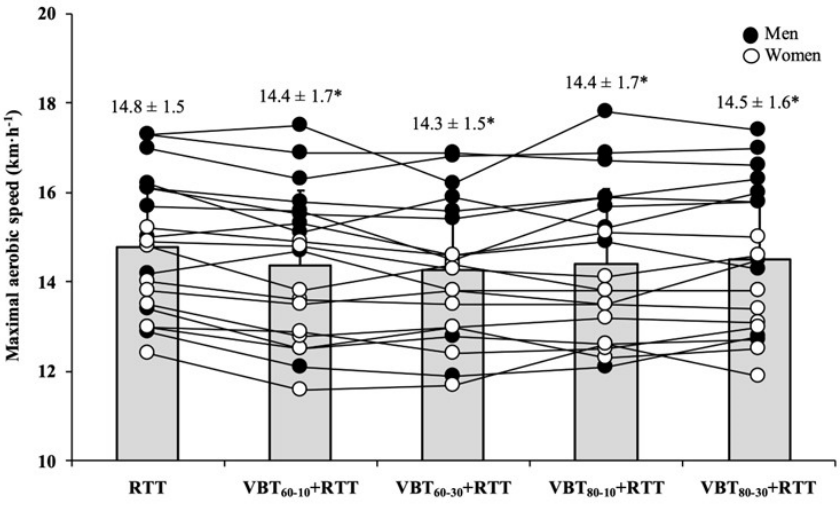

Figure 2. Comparison of the maximal aerobic speed between protocols. Data are depicted as means and standard deviation, whereas each point represents the individual data of each man (black circles) and woman (white circles). RTT, running track test; $VBT_{60–10}$ + RTT, velocity-based training (VBT) with 60% of one-repetition maximum (1RM) and a velocity loss in the set of 10% followed by RTT; $VBT_{60–30}$ + RTT, VBT with 60% of 1RM and a velocity loss in the set of 30% followed by RTT; $VBT_{80–10}$ + RTT, VBT with 80% of 1RM and a velocity loss in the set of 10% followed by RTT; $VBT_{80–30}$ + RTT VBT with 80% of 1RM and a velocity loss in the set of 30% followed by RTT. *, significantly lower than RTT ($p < 0.05$; analysis of variance with Bonferroni correction).

4. Discussion

This study was designed to examine the acute effect of four different VBT protocols ($VBT_{60–10}$ + RTT, $VBT_{60–30}$ + RTT, $VBT_{80–10}$ + RTT, and $VBT_{80–30}$ + RTT) on the MAS estimated from an RTT in recreationally trained men and women. Results revealed that, when compared to the control condition (RTT alone), the MAS was acutely compromised after the four different VBT protocols. Regardless of sex, the $VBT_{60–30}$ + RTT impaired the MAS to a greater extent than the $VBT_{60–10}$ + RTT, $VBT_{80–10}$ + RTT, and $VBT_{80–30}$ + RTT protocols. These results suggest that, regardless of sex, running performance (MAS) is impaired when preceded by VBT, especially if the training sets are performed with a low relative load (60% 1RM) and a high velocity loss threshold (30%).

The main strength of this study has been the implementation of the VBT methodology during resistance training sessions. First, the subjects received velocity performance feedback immediately after each repetition. This is of paramount importance, since the provision of velocity performance feedback has been proposed as an effective strategy to increase the quality (i.e., movement velocity) of the strength-oriented resistance training sessions [25]. Second, the individualized load-velocity relationships were used to match the intensity of load to individuals' daily readiness to train. In line with previous research [15], the fastest velocity significantly differed between relative loads, but not for the same relative load. Note that, while the accuracy of traditional methods may be affected by normal daily fluctuations in strength levels [26], the individualized load-velocity profiles provided high stability to prescribe resistance training intensity [27]. Third, velocity loss thresholds have been proposed as a more objective and homogeneous alternative to control proximity to failure during non-failure resistance training sets (e.g., subjects can complete ~60% of the maximum possible number of repetitions when reaching 30% velocity loss during the back-squat sets performed against 60% 1RM) [28]. Of note, while the target velocity loss is commonly determined from the fastest velocity achieved in each training set [13,14], the fastest repetition from the first set was used in the present study to guide set termination. Like Pérez-Castilla et al. [15], we have observed that the fastest velocity was higher for the first set than the third set. Therefore, if the fastest velocity of each set had been taken as the criterion, the subjects would have been closer and closer to muscular failure in the successive sets. Practitioners must keep this methodological aspect in mind when comparing different VBT studies.

The RT-SEP phenomenon suggests that neural and metabolic fatigue derived from previous resistance training sessions may compromise the quality of subsequent endurance training sessions and, consequently, induce sub-optimal endurance adaptations [8]. In line with previous research [10,29,30], our results provide further evidence that running performance (i.e., MAS) is compromised when preceded by a resistance training bout. This phenomenon may be related to various mechanisms, including (but not limited to) (i) impaired neural recruitment patterns, (ii) attenuated movement efficiency, (iii) increased muscle damage and soreness, and (iv) reduced muscle glycogen [8,9]. However, RT-SEP is a complex phenomenon conditioned by multiple training variables, including resistance training intensity and volume as acute interference modulators [9]. Indeed, supporting our hypothesis, the MAS was compromised to a greater extent when a low relative load (60% 1RM) along with a high velocity loss threshold (30%) was used in the set. This finding partially concurs with the results of Pérez-Castilla et al. [15] who reported, regardless of intensity (60% = 80% 1RM), a greater impairment in rowing ergometer performance when preceded by VBT protocols with high velocity loss in the set (30% > 10%). Such discrepancies regarding the effect of the relative load with the Pérez-Castilla's study [15] could be partly explained by the endurance performance indicator (MAS estimated from RTT vs. 2000 m rowing ergometer time trial), or the level of the study sample (recreationally trained adults vs. competitive rowers). More specifically, the muscles groups involved (back squat in our study vs. prone bench pull in Pérez-Castilla's study [15]) may explain potential differences between studies. Indeed, our results are in line with Nájera-Ferrer et al. [13] who reported a higher detriment to running performance when subjects completed three full-squat sets at 60% 1RM with a magnitude of 40% but not 20% velocity loss. It is therefore not surprising that, due to the higher fatigue levels and slower rates of recovery, a higher MAS has been reported following a concurrent running and VBT program with a lower velocity loss threshold in the set (15% vs. 45%) [14].

It has been shown that men have a higher muscle mass and proportion of type I fibbers than women [31]. Such sex differences might explain the greater muscle fatigability in women compared to men [31]. Rissanen et al. [17] recently observed that women require a greater velocity loss (40%) than men to maximize strength and power gains after an eight-week resistance training program. More specifically, Taipale et al. [19] reported a greater amount of fatigue, in terms of decreased maximal and explosive strength, in men than in women after a concurrent training session. In disagreement with those studies [17,19], our hypothesis was rejected since men not only reported a comparable number of repetitions during VBT protocols, but also a comparable MAS performance deterioration after VBT protocols than women. It is possible to speculate that the greater muscular fatigability in the women may be offset by the greater muscular strength (back squat 1RM relative to body mass = 1.8 vs. 1.6) and VO_2max (46 vs. 37 mL·kg^{-1}·min^{-1}) reported in the men. Of note, research has reported that resistance-untrained individuals present a greater magnitude of muscle damage and attenuation in muscle function than resistance-trained individuals [32]. Taken together, our results suggest that there is a comparable impairment of running performance (MAS) immediately after VBT between recreationally trained men and women.

Several issues need to be acknowledged when interpreting the findings from the present study. First, it should be taken into account the training status and history of our sample before generalizing the results of this crossover study. For example, Walker et al. [18] revealed that men were more susceptible to acute loss in force production capacity after different VBT protocols, but there were no signs of females being less fatigable after the eight-week velocity-based intervention. Second, we have examined the impact of resistance training-induced fatigue on endurance performance following a single bout of resistance training. In this regard, Doma et al. [33] reported that the magnitude of the increase in muscle damage markers was attenuated following a second resistance training bout in resistance-untrained runners and, therefore, it has been speculated that repeated resistance training bouts during concurrent training could minimize RT-SEP [8]. Finally, it is important to highlight that the subjects performed both training modalities (VBT and RTT) in the same session. Note that previous research [34] has reported that the magnitude of increase in peak oxygen consumption was greater when resistance and endurance training was performed on alternate days with 24 h of recovery compared to both training modalities performed in the same session with and without six hours of recovery, suggesting that the interference effect depends on the recovery period, although the influence of training variables must be also kept in mind [8].

5. Conclusions

Resistance training can improve running performance. However, several prescription variables should be considered to minimize the RT-SEP phenomenon during concurrent training programs. Our results revealed impaired running performance (MAS estimated from RTT) when preceded by different VBT protocols in loading magnitude (60% vs. 80% 1RM) and velocity loss in the set (10% vs. 30%). This acute interference effect was comparable between recreationally trained men and women. Additionally, the greatest MAS detriment was reported after VBT performed with a low relative load (60% 1RM) together with a high velocity loss threshold (30%). Therefore, practitioners who wish to optimize running performance while simultaneously incorporating resistance training into endurance athletes' training programs should avoid high repetition volumes to reduce susceptibility to RT-SEP. Indeed, a recent systematic review with meta-analysis [35] indicated that high-repetition strength training may not result in improved performance in competitive endurance athletes over a four- to twelve-week period, although a high-repetition strength training session induces high physiological (blood lactate concentration > 8.8 mmol·L^{-1})

and perceptual (rating of perceived exertion $\geq$ 17) demands. Hence, regardless of sex, running endurance athletes may consider a cautious approach when implementing resistance training sessions of low load (e.g., 60% 1RM) and high volume (e.g., 30% threshold for velocity loss in the set), as running endurance performance can be acutely reduced, which might affect long-term adaptations. Instead, VBT sessions involving greater load (e.g., 80% 1RM) and controlled volume (e.g., 10% threshold for velocity loss in the set), might offer better results in male and female endurance runners' performance.

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Data Availability Statement: The data presented in this study are available on request from the corresponding author due to privacy.

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Effect of a Short-Term Combined Balance and Multidirectional Plyometric Training on Postural Balance and Explosive Performance in U-13 Male and Female Soccer Athletes

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Abstract: This study's aim is to examine the effect of a combined balance and multidirectional plyometric training intervention on postural balance ability and lower limb explosive performance in U-13 male and female soccer athletes. Twenty pre-adolescent (age: 12.6 ± 1.6 years) soccer athletes followed a 6-week training intervention combining balance exercises, dynamic stabilization tasks and multidirectional plyometric exercises at a frequency of twice/week for 20–25 min, based on a progressive increase in exercise difficulty from phase A (week 1–3) to phase B (week 4–6). Pre- and post-training measurements were carried out to assess the following: (a) static balance performance in single (left, right)-legged and two-legged quiet stance trials with eyes open and eyes closed (two trials per stance and vision condition of 30 s duration) and (b) lower limb explosive performance in countermovement and squat jumps without arm swing (three trials/jump). The vertical GRF was recorded by a customized force plate (Wii, 1.000 Hz, Biovision) and offline, CoP and explosive performance parameters were calculated. The overall results showed that the static balance ability of athletes remained unaffected, while restricting their vision deteriorated their postural control. The lower limb explosive performance showed a trend for improvement; however, inter-individual variations in athletes' responses might have obscured any effect.

Keywords: static balance; plyometrics; biomechanical analysis; jumping performance; preadolescence; soccer

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1. Introduction

Soccer is listed as the top activity in the preferences of children and adolescents with regards to their participation in sport- and leisure-time physical activities [1]. Muscle strength and power constitute critical parameters that relate to performance in soccer [2], since due to the sport's intermittent nature, along with the high physical and energetic demands [3], substantial parts of the total covered distance are performed at high or very high intensity [4]. Moreover, within the context of the athletes' technical and tactical choices, a major part of their actions is based on rapid and sudden change-of-direction movements [5]. To enhance and optimize muscle strength and power in athletes, soccer training applies plyometric training [6]. As is well known, plyometric training is based on the concept of training specificity by advocating the use of vertical and horizontal jumps and displacements in order to improve performance by the transference of gains in strength and power to performance parameters related to vertical and horizontal force production (for example, vertical–horizontal jumps, sprints) [7]. A recent meta-analysis that compared the effect between vertically and horizontally orientated plyometric training on physical performance showed that both methods were effective at enhancing respective outcome measures [8]. It was also shown that horizontally orientated plyometric training might be a more efficient method for enhancing multi-vector performance-related measures. Soccer's movement patterns are predominantly multidirectional, and, indeed, previous studies have found greater benefits of multidirectional plyometric training [9–12] compared to single-vector (vertically or horizontally directed) plyometric training on soccer performance. Despite the relative homogeneity of those studies in training frequency and intervention duration [9–12], there is, however, a paucity of studies focusing on investigating female athletes and especially of preadolescent age, since most of the existing evidence is derived from young or adult male athletes.

Balance, on the other hand, is a fundamental skill that contributes both to the efficient technical execution of a sport's movements and to injury prevention, while an improvement in postural control will enhance athletic performance [13]. In soccer, the complexity of technical actions, for which unilateral stance and/or body weight transfer is frequently required in combination with unpredictable situations during training or the game, which demand postural regulation of the body (e.g., unpredicted changes of ball direction, interference with the opponent while being in motion, strenuous physical contact in an offensive or defensive action, etc.), constitute the necessary development of postural skills in soccer players [14,15]. Previous short-term training balance interventions in young and adult soccer athletes have shown improvements in static and dynamic balance as well as in technical features [16–18]. Moreover, the implementation of a balance training program throughout the competitive season (~40 weeks) resulted in a significant enhancement in dynamic balance performance and a respective decrease in the percentage occurrence of lower extremity injuries in young soccer players [19]. Further, the effect of balance training and plyometric training interventions has been investigated in young soccer athletes, either separately [20,21] or with a variation in training sequence [22] or with alternating training schedules within the total intervention duration [23]. Findings have not been conclusive up to now with regards to the stronger influence of one training intervention or training sequence over another, as some researchers have found greater adaptations in performance measures related to balance, jumping ability and velocity, when 4 weeks of plyometric training were preceded by balance training of equivalent duration [22].

Vision is the main structure involved in postural control as it provides input, enabling the body to actively control its alignment for the purposes of orientation and stability [24]. The contribution of visual information has been studied in athletic activities, where regulation of the body's posture and/or perception of its orientation with regards to the external environment by means of sensory input is critical for successful performance [25,26]. Expert soccer players were found to have smaller dependence on vision for postural control compared to less experienced players, and they used vision for processing game-related information, like cooperation with other team members or interaction with the opponents [15].

Therefore, in the present study, we were interested in the implementation of a combined balance and plyometric training intervention in male and female preadolescent soccer athletes, aiming at the collection of evidence to be used by coaches and strength and conditioning trainers to design training programs for performance optimization. The study's primary purpose was to examine the possible effect of short-term balance and multidirectional plyometric training on balance and lower limb explosive performance in male and female preadolescent soccer athletes. It was hypothesized that the short-term combined balance and multidirectional plyometric training would improve the balance and the lower limb explosive ability in the athletes. A secondary purpose was to examine possible differences between male and female soccer athletes resulting from the possible effect of the combined training intervention. The study's final purpose was to investigate the interaction between the training intervention and a restriction of vision. Considering previous findings on the visual contribution in athletic activities, we hypothesized that the short-term combined balance and multidirectional plyometric training would assist the athletes in relying on the other sources of available sensory information, when vision was restricted, to regulate their postural control.

2. Materials and Methods

2.1. Participants

Twenty (ten male and ten female) pre-adolescent (aged 12.6 ± 1.6 years) soccer athletes with a training experience of 4.5 ± 2.7 years volunteered to participate in this study. Inclusion criteria required that the athletes had participated in a minimum of 80% of the team's training during the last 3 months prior to the start of the study and did not report any ligamentous or musculoskeletal injury in the last 6 months. The athletes trained systematically at a frequency of 3–4 sessions per week for 90 min and competed in one weekly game at the regional championship of the Greek Association of Soccer Clubs in the U-13 category. All athletes and their parents or legal guardians gave their written informed consent to participate in this study in accordance with the Helsinki Declaration after being thoroughly informed about the study's procedures. The study was approved by the Ethics Committee of the School of Physical Education and Sport Science, National and Kapodistrian University of Athens (approval number: 1520/2023).

2.2. Study Design and Intervention Protocol

A one-group repeated-measures design was applied with pre-training (PRE) and post-training (POST) measurements. Soccer's demanding nature with regards to physical and neuromuscular requirements necessitates multimodal training approaches. To that aim, based on previous work that implemented solely plyometric [9–12] or balance training [16,18,21] or a combination of both [22,23,27], we designed a 6-week training intervention (2 sessions/week, 20–25 min/session), comprising static balance tasks, dynamic stabilization tasks and multidirectional plyometric exercises. The intervention was divided into two phases of 3 weeks duration each, which were characterized by a progressive increase in task/exercise difficulty with regards to the base of support (two-legged vs. single-legged base of support), visual information (eyes open vs. eyes closed), movement direction and execution velocity [28,29]. Table 1 presents, in detail, the implemented training intervention. Prior to the main part of their afternoon typical soccer training, the athletes followed a standardized warm-up of 5–7 min in duration, involving jogging and 2–3 running drills at light intensity followed by dynamic stretching of the major muscle groups of the upper and lower body, and then carried on with the training intervention. Caution was taken so that a 48 h break period apart from training sessions occurred to avoid fatigue and to provide full recovery. Athletes participated in two familiarization sessions before the start of the training intervention. In one of those sessions, anthropometric measurements of body mass, height, sitting height and left and right lower limb length (measured distance from the top iliac crest to medial malleolus in the supine position) were carried out. Biological maturation was assessed based on the equations, as previously proposed [30].

Table 1. Combined balance and multidirectional plyometric training intervention of the study.

Phase ^{A}	Balance Tasks	Set × Exercise Duration/Rest	Dynamic Stabilization Tasks	Set × Exercise Duration/Rest	Multidirectional Plyometric Exercises	Set × Exercise Duration/Rest
	Two-legged stance-EC *	1 × 30 s/ 15 s	Forward Walking Lunge - Execution rate: Slow Distance: short	1 × 15 s per leg/ 15 s	Standing vertical jump with alternating unipedal landing	1 × 30 s/ 15 s
	Single-legged stance-EO *	1 × 30 s/ 15 s	Diagonal Forward Lunge – Execution rate: slow Distance: short	1 × 15 s per leg/ 15 s	Side jumps	1 × 30 s/ 15 s

Table 1. *Cont.*

Phase [A]	Balance Tasks	Set × Exercise Duration/Rest	Dynamic Stabilization Tasks	Set × Exercise Duration/Rest	Multidirectional Plyometric Exercises	Set × Exercise Duration/Rest
A′ Weeks: 1–2–3 Warm up [B]: ~5′–7′	Two-legged stance on bosu ball-EO	1 × 30 s/ 15 s	Ski Jumps - Distance: short	1 × 30 s/ 15 s	Lateral Step-Up jumps	1 × 30 s/ 15 s
	Single-legged stance on Balance Disk-EO	1 × 30 s per leg/ 15 s	Single Leg Ski Jumps - Distance: short	1 × 15 s per leg/ 15 s	Drop jumps from a bosu ball to the ground	1 × 30 s/ 15 s
	Single-legged stance with hip rotation-EO	1 × 15 s per leg / 15 s	-	-	Vertical jump from a bosu ball to a box of 20 cm height	1 × 30 s/ 15 s
	Plank with alternating shoulder rotation	1 × 30 s per side/ 15 s	-	-	Static jumping lunges - Execution rate: slow	1 × 30 s/ 15 s
	-	-	-	-	Single-legged Rise Dead Lifts	1 × 15 s per leg/ 15 s
B′ Weeks: 4–5–6 Warm up [B]: ~5′–7′	Single-legged stance-EO	1 × 30 s/ 15 s	Forward walking lunge and core rotation with a ball – Execution rate: fast Distance: long	1 × 15 s per leg/ 15 s	Horizontal Jump, Hand on the hips and landing on the toes - Distance: Long	1 × 60 s/ 60 s
	Single-legged stance-EC	1 × 30 s/ 15 s	Diagonal forward lunge - Execution rate: fast Distance: long	1 × 15 s per leg/ 15 s	Repeated horizontal jumps with hands on the hips and landing on the toes - Distance: long	1 × 75 s/ 60 s
	Single-legged stance on Balance Disk-EO	1 × 30 s per leg / 15 s	Ski Jumps - Distance: long	1 × 30 s/ 15 s	-	-
	Single-legged stance on ground with hip rotation-EO	1 × 15 s per leg/ 15 s	Single Leg Ski Jumps - Distance: long	1 × 15 s per leg/ 15 s	-	-
	Plank with alternating shoulder rotation	1 × 15 s per side/ 15 s	-	-	-	-

[A] Frequency of training in phase A and B was two sessions/week, 20–25 min duration. [B] Warm-up consisted of jogging and 2–3 running drills at light intensity followed by dynamic stretching of the major muscle groups of the upper and lower body. * EC–EO: eyes closed, eyes open.

2.3. Static Balance Assessment

From a biomechanical perspective, the two-legged quiet stance is treated as the most conventional condition of balance assessment in order to gather evidence about a subject's ability to regulate one's center-of-mass oscillations within the limits of the base of support [31,32]. This test is typically accompanied by single-legged stance trials, for they are considered the basic mode of postural challenge with regards to manipulating the base of support [31]. This method of static balance assessment has been previously used either in children or adolescents [29,33,34] as well as in athletes [35,36]. Therefore, the assessment of static balance took place before (PRE) and after the 6-week period (POST) of the training intervention during single-legged and two-legged quiet stance trials in a quiet and spacious room of the sport club's facilities. During assessment of the single-legged stance trials, the athletes were instructed to assume a straight body posture with their arms hanging relaxed by their sides, to flex their hip joint and their knee joint at 90 degrees and to stand as motionless as possible on the force plate with either left or right leg. For the two-legged trials, the athletes assumed the same posture as described above, with their feet hip width apart. During the trials with eyes open, the athletes were instructed to fix their gaze on an imaginary point on the wall 2–3 m in front of them, while keeping their heads in a neutral position parallel to ground level, while during the trials with eyes closed, care was taken so that one researcher was always situated behind the athletes for safety reasons. Two successful trials per stance and visual condition were performed in a randomized order of 30 s duration each with 15 s of rest across trials and 1 min between testing conditions. Static balance performance was determined by the recording of center-of-pressure (CoP) data with a vertical force plate (Wii, A/D converter, 24-bit resolution, 1.000 Hz, Biovision, Wehrheim, Germany). Offline, the data were filtered using a 2nd bi-directional-order digital low-pass Butterworth filter with a 15 Hz cut-off frequency and analyzed with MATLAB custom-made scripts (R2012a, 64 Bit; Mathworks, Natick, MA, USA) from the 5th to the 25th second ($\Delta t = 20$ s) of each 30 s trial duration. Based on the CoP displacement, whose derived values represent the geometrical location of the vertical ground reaction force vector on the platform during quiet standing [37], the following parameters were determined: (a) CoP path length, defined as the sum of Euclidean distances between adjacent measurement points, and (b) CoP sway range, defined as the range (i.e., from minimum to maximum) of the CoP values in the anteroposterior and mediolateral directions. To assess performance in the two-legged quiet stance trials, the average values of the two trials were used, whereas in the single-legged trials, the average value of the mean left and right leg trials was used for analysis, since no statistically significant differences were observed between sides (paired t test: $p > 0.05$) in any parameter.

2.4. Explosive Performance Assessment

The assessment of lower limb explosive performance before (PRE) and after the 6-week training period (POST) was based on a protocol of countermovement (CMJ) and squat jumps (SJ) [38,39] by recording the ground reaction force with the use of a vertical force plate, as described above. Following a short familiarization period, during which the athletes performed 2–3 submaximal CMJs and SJs, where they were instructed to focus on (a) starting the propulsion phase from a position of 90-degrees knee flexion along with no countermovement for the SJ and from an erect position for the CMJ, (b) having a depth of the downwards movement that would allow for an unobstructed propulsion phase and (c) reaching full lower limb extension at the apex of the jump, they performed three successful CMJs and three SJs for maximum height without arm swing. A rest interval of 2 min between familiarization and measurements and 30 s between jump trials was provided to minimize fatigue. The trial with the highest height achieved was selected for further analysis, and explosive performance was determined by the parameters of jump height (m), maximum force (N), maximum impulse (N·s), and mean and maximum mechanical power (Watt) for the CMJ and SJ.

2.5. Statistical Analysis

For the statistical analysis, we first checked for the normal distribution of the CoP data using the Kolmogorov–Smirnov test with Lilliefors correction. Two-way ANOVAs for repeated measures were performed with training intervention (PRE vs. POST measures) as the within-subjects factor, and (a) sex (male vs. female) and (b) vision restriction (eyes open vs. eyes closed) as the between-subjects factors to check for possible differences in the dependent variables of CoP path length, CoP anteroposterior and CoP mediolateral sway during the single-legged and two-legged static balance assessments. In the event of a significant main or interaction effect, a Bonferroni-corrected pairwise analysis was conducted. Similarly, two-way ANOVA for repeated measures with training intervention and sex as fixed factors was performed to test for possible differences in the anthropometric and lower limb explosive performance outcome measures with post hoc Bonferroni-corrected pairwise analysis. The percentage change due to training intervention was calculated ((%Δ: (POST − PRE)/POST × 100%), and Cohen's d with values of $0.2 < d \leq 0.5$, $0.5 \leq d < 0.8$ and $d \geq 0.8$ determining a small, medium or large effect size due to training intervention was calculated [40]. All statistical analyses were performed using SPSS IBM v.21, and the significance level was set at a = 0.05.

3. Results

3.1. Anthropometric Characteristics

Due to the non-statistically significant training intervention by the sex interaction effect in any anthropometric parameter, Table 2 presents the anthropometric characteristics of all the athletes and the effect size of training intervention, based on the results of paired-sample *t* test analysis. After the 6-week training period, age (t (19) = 71.1), body mass (t (19) = 4.1), body height (t (19) = 4.7) and lower limb length (t (19) = 3.6) were significantly increased. There was no statistically significant difference in maturity offset (t (19) = 1.9, $p = 0.605$) and body mass index (t (19) = 1.7, $p = 0.098$) (Table 2).

Table 2. Male and female soccer athletes' anthropometric characteristics before (PRE) and after (POST) the combined balance and multidirectional plyometric training intervention.

Parameter	PRE	POST	%Δ	Cohen's d
Age (years)	12.6 ± 1.6	12.7 ± 1.6 ***	+0.8%	0.10
Maturity offset (years) [a]	−0.76 ± 1.46	−0.75 ± 1.49 [ns]	+1.8%	0.01
Body mass (kg)	47.1 ± 10.8	48.2 ± 11.1 ***	+2.3%	0.09
Body height (cm)	157.1 ± 13.4	158.0 ± 13.1 ***	+0.6%	0.07
Body mass index (kg/m^2)	18.8 ± 1.9	19.0 ± 2.1 [ns]	+1.1%	0.10
[b] Lower limb length (cm)	83.5 ± 6.8	84.3 ± 7.2 **	+1.0%	0.10

[a] Years to the age at peak height velocity. [b] Average value of the left and right lower limb length. %Δ: Post- vs. pre-measurements. Statistically significant difference between pre–post measurements: *** $p < 0.001$, ** $p < 0.01$, [ns] $p > 0.05$. Data are mean ± SD.

3.2. Static Two-Legged Balance Performance

With regard to results on two-legged static balance performance, due to the non-statistically significant training intervention by the sex interaction effect in any CoP parameter, the results refer to the main effect of training intervention and vision and to the training intervention considering the vision interaction effect. Specifically, there was a significant main effect of training intervention only in the anteroposterior CoP sway range ($F_{1,38} = 7.52$, $p = 0.009$), with higher values post-training (PRE vs POST measurement: 1.8 ± 0.4 vs. 2.5 ± 0.3 cm) (Figure 1B). There were no significant differences due to training intervention for the CoP path length ($F_{1,38} = 0.35$, $p = 0.564$, Figure 1A) and the sway range in the mediolateral direction ($F_{1,38} = 2.38$, $p = 0.131$, Figure 1C). Vision had a significant main effect in the CoP path length ($F_{1,38} = 30.12$, $p < 0.001$), with a higher CoP displacement found for the eyes-closed condition (Figure 1A). No statistically significant differences were found for either the anteroposterior ($F_{1,38} = 0.14$, $p = 0.706$, Figure 1B) or the mediolateral CoP sway range ($F_{1,38} = 0.54$, $p = 0.468$, Figure 1C) due to vision restriction. The interaction effect of training intervention and vision was statistically significant ($F_{1,38} = 29.67$, $p < 0.001$) in the CoP anteroposterior sway range, where athletes had significantly higher CoP sway values after the 6-week training intervention with eyes closed, as compared with the lower increase in the respective values in the eyes-open condition (Figure 1B). There was a non-statistically significant training intervention for the vision interaction effect for the CoP path length ($F_{1,38} = 0.46$, $p = 0.500$, Figure 1A) and the sway range in the mediolateral direction ($F_{1,38} = 0.20$, $p = 0.661$, Figure 1C).

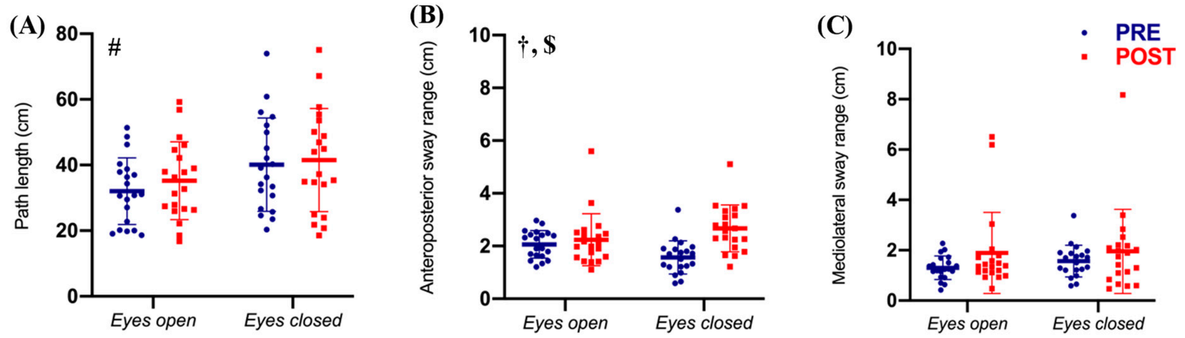


Figure 1. Static two-legged balance performance as determined by: (**A**) CoP path length, (**B**) CoP anteroposterior sway range, and (**C**) CoP mediolateral sway range with eyes open and eyes closed before (PRE, blue circles) and after the 6-week (POST, red squares) training intervention. # Statistically significant main effect of vision, $p < 0.001$. † Statistically significant main effect of training intervention, $p < 0.01$. $ Statistically significant interaction effect of training intervention and vision, $p < 0.001$. Data in scatterplots are individual values and the mean ± SD value is also depicted.

3.3. Static Single-Legged Balance Performance

In Table 3, the results on static single-legged balance performance are shown. The analysis did not yield any statistically significant training intervention by the sex interaction effect in any CoP parameter in the single-legged balance assessment; thus, the results presented here refer to the main effects of training intervention and vision, and to the training intervention by the vision interaction effect. Training intervention did not have a significant main effect on the CoP path length ($F_{1,38} = 0.002$, $p = 0.969$), range of CoP sway in the anteroposterior ($F_{1,38} = 0.11$, $p = 0.745$) and the mediolateral direction ($F_{1,38} = 0.03$, $p = 0.867$)). A significant main effect of vision was found for the CoP path length ($F_{1,38} = 180.2$, $p < 0.001$), the anteroposterior ($F_{1,38} = 133.8$, $p < 0.001$) and mediolateral sway range ($F_{1,38} = 66.8$, $p < 0.001$). There was a non-statistically significant training intervention by vision interaction effect in the CoP path length ($F_{1,38} = 0.07$, $p = 0.800$), the anteroposterior ($F_{1,38} = 1.1$, $p = 0.299$) and the mediolateral CoP sway range ($F_{1,38} = 0.18$, $p = 0.677$) (Table 3).

Table 3. CoP parameters in the single-legged quiet stance trials with eyes open and eyes closed before (PRE) and after (POST) the combined balance and multidirectional plyometric training intervention.

CoP Parameter	Vision	PRE	POST	%Δ	Cohen's d
Path length (cm)	Eyes open	127.9 ± 25.2	129.9 ± 25.0	+1.6%	0.08
Anteroposterior sway range (cm)		5.8 ± 2.1	4.9 ± 1.7	−15.2%	0.46
Mediolateral sway range (cm)		4.2 ± 1.3	3.8 ± 1.4	−9.5%	0.30
Path length (cm)	Eyes closed	258.4 ± 65.5	255.5 ± 66.1	−1.1%	0.04
Anteroposterior sway range (cm)		11.9 ± 3.9	12.2 ± 4.3	+2.8%	0.08
Mediolateral sway range (cm)		9.2 ± 4.5	9.4 ± 4.0	+1.6%	0.03

Note: Results are the average value of the left and right lower limb due to no statistically ($p > 0.05$) significant difference in the examined parameters. %Δ: Post- vs. pre-measurements. Data are mean ± SD.

3.4. Explosive Performance

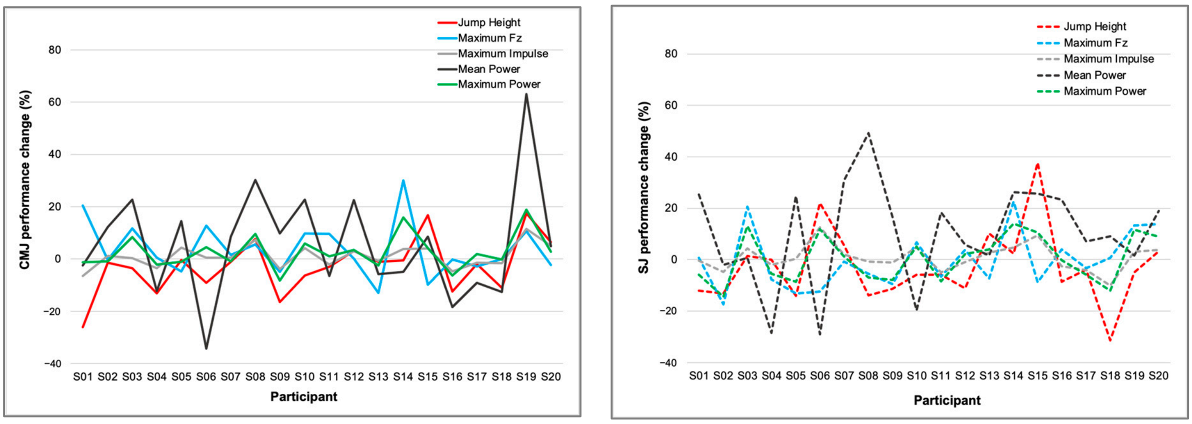
Table 4 presents the results on the lower limb explosive performance and the effect size due to training for all athletes. Due to the non-statistically significant training intervention by the sex interaction effect in any parameter, the results refer to a paired-sample t test analysis. There was no significant change due to training intervention, either in the CMJ's or the SJ's height (CMJ: $p = 0.134$, SJ: $p = 0.303$), maximum vertical force (CMJ: $p = 0.255$, SJ: $p = 0.708$), maximum impulse (CMJ: $p = 0.313$, SJ: $p = 0.449$), and mean (CMJ: $p = 0.988$, SJ: $p = 0.088$) and maximum mechanical power (CMJ: $p = 0.219$, SJ: $p = 0.879$).

Table 4. Parameters of lower limb explosive performance before (PRE) and after (POST) the combined balance and multidirectional plyometric training intervention.

Parameter	Jump	PRE	POST	%Δ	Cohen's d
Jump height (cm)		20.4 ± 4.4	19.6 ± 3.8	−3.9%	0.18
Maximum vertical force (N)	CMJ	1023 ± 275	1051 ± 271	+2.7%	0.10
Maximum impulse (N·sec)		104 ± 30	105 ± 30	+1.0%	0.04
Mean power (Watt)		190 ± 81	190 ± 64	0%	0.02
Maximum power (Watt)		1764 ± 578	1797 ± 561	+1.5%	0.06
Jump height (cm)		17.3 ± 3.5	16.7 ± 3.6	−3.5%	0.16
Maximum vertical force (N)	SJ	1115 ± 271	1104 ± 275	−1.0%	0.04
Maximum impulse (N·sec)		97 ± 27	97 ± 27	0%	0.03
Mean power (Watt)		369 ± 136	401 ± 148	+8.7%	0.23
Maximum power (Watt)		1743 ± 524	1737 ± 499	−0.3%	0.01

%Δ: Post- vs. pre-measurements. Data are mean ± SD.

Figure 2 depicts the individual percent change in the explosive performance parameters of each jump due to the effect of the combined training intervention in the preadolescent soccer athletes.



(**a**) (**b**)

Figure 2. Individual change (%Δ: post- vs. pre-measurement) in lower limb explosive performance during jumping ((**a**): CMJ, (**b**): SJ) in the preadolescent soccer athletes (note: to facilitate visual inspection of the graphs, the first 10 series of data refer to the male and the next 10 series of data to the female soccer athletes).

4. Discussion

The purpose of this study was to examine the effect of combined balance and multidirectional plyometric training in male and female preadolescent soccer athletes. Overall, the results did not confirm the main hypothesis since the postural balance ability and the lower limb explosive performance of athletes did not show a significant improvement post-training. Further, the absence of a significant interaction between training and visual restriction leads to the rejection of the hypothesis, referring to the possible benefit of the training intervention in processing visual information for postural control regulation in preadolescent soccer athletes.

The present findings show that the combined training intervention resulted in a significant deterioration of static two-legged balance due to a 39% increase in the anteroposterior CoP sway range. On the other hand, the CoP path length and mediolateral sway range showed a non-statistically significant increase by 6.4% and 36%, respectively, whereas single-legged static balance performance was found to be slightly, albeit not significantly, improved after training. In the literature, no studies combining balance and plyometric training could be found in soccer, except for two recent studies in other team sports. In particular, and in agreement with our findings, no significant change was found in single-legged quiet stance balance ability in 15–16-year-old female regional-level basketball athletes following 8-week combined balance and plyometric training [27]. On the contrary, an improvement in dynamic anteroposterior and mediolateral CoP sway was seen after landing either from a horizontal or a lateral jump for short distance (40% of height) in elite male badminton players, who participated in a combined intervention (6 weeks at a frequency of 3 days/week for 1 h) [41]. In the present study, the training volume was the result of balance and plyometric training exercises, but in that study [41], plyometric training was twice the volume of the respective balance training's volume. Moreover, the present findings on static balance performance are in disagreement with a previously reported training intervention (8 weeks, 24 sessions) [21], which had several similarities in reference to balance exercises and dynamic stabilization tasks as the ones administered in the current study (see Table 1). In that study [21], an enhancement by approximately 38% and 25% was reported for static and dynamic balance in the dominant leg of adolescent (12–13 years old) male soccer players. Similarly, the static, semi-dynamic as well as dynamic balance in younger (9–11 years old) male soccer athletes was enhanced following their participation in a 12-week balance training, comprising static one- and two-legged balance tasks, semi-dynamic balance tasks, along with walking and running on both stable and unstable surfaces [16]. Further, in elite junior (aged 16.0 $\pm$ 0.5 years) male soccer players who were involved for 10 weeks at a frequency of twice a week in plyometric unloaded versus ankle-loaded (2.5% of body weight) training, there was an improvement in the stork balance time score performed with open eyes in both groups [42]. Previous 6- to 8-week training interventions that have used a plyometric training protocol combining horizontally and vertically directed jumps found a decrease in the anteroposterior and mediolateral sway range by approximately 15% and 19% during two-legged static balance tasks with open and closed eyes in 9–13-year-old male soccer players [11], while an improved dynamic balance ability was seen in male soccer athletes of similar age to our athletes as well as in older ones [9,10].

The disparity between the present findings and those previously reported might be attributed to the heterogeneity of training regimens. Training volume was higher in most of those studies [9,16,21,37,38], as compared to the current study. Solely, Ramirez-Campillo et al. [11] implemented the same training volume (i.e., 12 sessions for a frequency of training twice/week). However, their intervention consisted of a combination of horizontal and vertical jumps, which amounted to a greater total number of plyometric jumps. The difference in the plyometric training stimulus might explain the absence of changes in static balance ability that was observed here, since a meta-analysis concluded that plyometric training interventions have a significant but small effect size on static and dynamic balance performance [43].

Moreover, it has been shown that the transfer of postural balance adaptations is strongly associated with the context in which the physical activity or sport is practiced [13]. It is highly likely that differences in the content of balance exercises used in previous studies [16,21,42] could account for the observed disparity in results. Similar to Bouteraa et al. [27] and Lu et al. [41], who subjected their athletes to a combined balance and plyometric training intervention, our soccer athletes practiced single-legged static stance tasks and dynamic displacements both on various bases of support and types of support surfaces (see Table 1). Even though the present results suggest a trend towards a slight improvement in single-legged balance performance, since CoP parameters were decreased by 0.2 to 1.9%, this study's training stimulus was probably not adequate to bring about significant changes, as previously reported [16,21,41,42].

Static balance assessment based on CoP measures is considered as a reliable method to examine postural steadiness [31,33–37]. With the exception of Cè et al.'s study [16], previous authors, who have found improvements in static balance ability [21,42], assessed performance based on the time score. It can be acknowledged that the time to complete a single- or two-legged balance task is very different to the CoP oscillations, which are being recorded within that same time period and reflect the center-of-mass effort to generate a corrective torque in order to counter the destabilizing gravitational torque [24]. Lu et al. [41] also assessed CoP sway by means of recording the vertical ground reaction force, but their positive results refer to dynamic balance performance, thus limiting any direct comparison with the current findings.

Plyometric training is a widely recognized method to improve and maximize muscle strength and power. As mentioned, the present combined training regimen did not induce any significant effect on the parameters determining the lower limb explosive performance. However, the current findings showed a small ($p > 0.05$) improvement in maximum force, maximum impulse and maximum power in the CMJ task (1.0–2.7%) and a greater non-statistically significant change in mean power (+8.7%) in the SJ task, whereas jump height was decreased (-3.5 to -3.9%). Quite a large number of studies have focused on jump height, the reason being that it is probably the most frequently chosen parameter for assessing lower limb explosive performance. Evidence of a moderate-to-large effect size of plyometric training on jump height has been previously reported, regardless of athletes' sex [43], and a small-to-large effect size has been found for female athletes [44]. In contrast with the present findings, a positive effect of short-term (6–8 weeks) plyometric multidirectional training has been shown in male [9,10] and female soccer athletes [11,12] on CMJ and SJ height. Recently, it was argued that interventions ranging from 400 to 600 min of total duration are required in order to achieve the maximization of the plyometric training effect on vertical jump height during adolescence [45]. Even if these suggested optimal intervention durations were scaled to our preadolescent athletes, the total duration of the plyometric part of the current study's combined intervention would still be considerably shorter, as it amounted to approximately 88 min. It is highly likely that the lesser, by 1/6 plyometric, training stimulus here, as compared to the recommended optimal one, accounts for the observed decrease in CMJ and SJ height.

On the other hand, the results showed a tendency towards an improvement in maximum force, impulse and power in the CMJ and in mean power in the SJ, respectively. It has been theoretically and experimentally argued and examined in vivo for the vastus lateralis muscle that jump height does not constitute the most appropriate parameter for the assessment of the maximal force and power-generating capacity of the lower limbs in systematically trained individuals [46,47]. The production of mechanical power during the propulsion phase of the jump, which will potentially affect the achieved jump height, is determined to a considerable extent by intrinsic neuromuscular mechanisms and, in particular, by the force–length, power–velocity and force–velocity potential. The athletes received specific instructions about how to execute the CMJ and SJ tasks with regard to fully extending their lower limb joints during the propulsion phase [38]. Assuming that their lower limb muscles were maximally activated, as they were also instructed to perform maximum-effort jumps, the lower limb joints' excursion corresponds to the distance over which muscles will generate force and mechanical power, and, thus, it could be hypothesized that all athletes operated on a similar portion of their individual force–length curve.

Based on the proposed maximum dynamic output hypothesis, the optimum loading condition for maximizing power output during jumping is one's own body [48,49]. It has been suggested that due to individual neuromuscular characteristics and training history, some athletes will need a positive load (that is, an additional load to their body mass) and some others a negative one (that is, an assistance to decrease loading) in order to ensure optimal loading conditions [46]. The athletes had similar training history since they trained in the same sport club during the last 2–3 years. While the assessment of biological maturation did not present any statistically significant difference, the inspection of individual values showed that nine (five male and four female athletes) out of the twenty athletes had already achieved their peak height velocity by approximately 0.7 years at the end of the training intervention. These inter-individual differences in the maturation process of the male and female soccer athletes along with a 23% variation in their body mass suggest a differentiation in the jumps' optimal loading conditions and could probably justify the slight trend ($p > 0.05$) towards the improvement in lower limb explosive performance post-training (see Figure 2). It has been found that vertical jumping performance differs between boys and girls from 14 years onwards due to changes mainly in leg length and the respective lean muscle volume [50]. Leg length was significantly increased by 1% for the whole sample after the intervention. A greater change in leg length of the nine biologically more mature athletes could be a possibility that, if true, could imply a considerable variation between the responses of the biologically more mature as compared to their less mature teammates. This inter-individual variation might have resulted in the absence of a significant overall training effect.

Each sport's postural requirements, in combination with the athlete's systematic practice on the sport's motor patterns, are capable of modifying one's degree of dependence on the sensory systems responsible for the regulation of postural control and balance [13]. We hypothesized that the athletes' practice in static balancing conditions under the restriction of vision could be beneficial to them with regard to their reliance on this source of sensory information for balance achievement and/or maintenance. A significant interaction was solely found for the anteroposterior CoP sway range in the two-legged stance task, where the decrease between restriction and no restriction of vision condition pre- training was followed by an increase post-training (see Figure 1B). No significant interaction was found in the other CoP parameters in the two-legged task as well as in any examined CoP parameter in the one-legged balance task. Overall, these findings suggest that the combined balance and MPT intervention did not contribute to shifting the reliance of athletes to a greater extent on other sources of sensory input (e.g., proprioception) for their balance control. This is in disagreement with a previously reported significant improvement in two-legged balance with eyes open by −16.2% and with eyes closed by −18.7% in the anteroposterior CoP sway and by −14.8% and −17.3% with eyes open and closed, respectively, in the mediolateral CoP sway found in male soccer players aged 11.2 ± 2.3 years old [11]. As mentioned already, that training intervention used plyometric exercises with horizontal and vertical jumps [11]. During a vertical and horizontal jump, there exist specific kinematic requirements for the body's center of mass during the propulsion phase in order to perform the jump with as much of an optimal coordination strategy as possible [51]. It is probable that the greater plyometric training load in that study was transferred to those athletes' ability to more efficiently regulate their body's position, hence the improvement in static balance [11].

There exists neurophysiological evidence suggesting that the contribution of vestibular and proprioceptive sensory information increases during the process of postural regulation as a function of the competitive level, while the contribution of visual information decreases [14,15]. Elite and expert soccer players were shown to have a lower reliance on vision and a higher temporal dedication of their eye movements in processing game-related information, as compared to lower-level non-expert players [14,15,52]. Thus, when a player uses the time required to process visual information for performing their own motor actions (e.g., body positioning regulation, ball control), one then reduces the available time to analyze the game and make strategic decisions about offense or defense [15]. In the present study, the limited training experience of the athletes combined with the short-term training stimulus most probably suggest the absence of a training by vision interaction effect, since the preadolescent male and female athletes were not equipped with such a competitive and/or expertise level to efficiently engage their vestibular and/or proprioceptive systems in the regulation of balancing tasks while their vision was being restricted.

A final note is dedicated here with regard to the study's secondary purpose that was related to a possible training intervention by sex interaction effect. As reported in the results, the statistical analyses did not yield a significant interaction effect in any balance or lower limb explosive performance measure. Taking into consideration the fact that most of the previous related work has examined male athletes [9–11,16,21,41,42], it would have been interesting to present a comparison based on sex. However, the absence of any interaction effect suggested that male and female preadolescent athletes responded in a similar manner to the combined training intervention, and, for that reason, respective results were not presented. Neurophysiological adaptations have been reported to be induced by balance training [53], whereas plyometric training can elicit neuromuscular responses [54]. However, a significant interference of gender with the effect of plyometric training on balance performance was not found in a recent meta-analysis [55]. On the other hand, young female athletes were previously reported to show greater adaptive responses when plyometric training interventions were of longer duration (>16 sessions), had greater weekly training frequency (>2 times) as well as longer durations in each session ($\geq$30 min) [56]. This study's short-term training stimulus in combination with the inter-individual differences in the athletes' estimated biological maturation might justify the absence of different responses between the male and female soccer athletes.

There are several limitations to be considered in this study. A main limitation was not evaluating the training load, since, despite the inclusion criteria requiring that the athletes had participated in 80% of the team's training sessions in the last 3 months before the start of the study, the observed variation in biological maturity status among the athletes cannot exclude the possibility that the training load was perceived as low for some athletes and as high for some others. Another important limitation is that the menstrual cycle of the female athletes was not controlled for. Even though the age range of the athletes fell within the time period where menarche typically occurs (10–16 years of age), a record and attempt to measure them at the same stage of their menstrual cycle post-training should have been made. Further, the sample team was competing in a regional soccer league, which probably implies that a part of the soccer training was specific to their playing position. Static balance was not assessed in relation to playing position due to the small sample size. Therefore, the possibility that the absence of a significant training intervention effect might be partly related with a variation in balance performance as a result of playing position-related training-induced postural adaptations [57,58] should be considered.

5. Conclusions

In conclusion, the short-term 6-week combined balance and multidirectional plyometric training was not effective in improving the static balance and lower limb explosive performance in preadolescent male and female soccer athletes. It is recommended that a training intervention of higher training volume and based on individualized training load using similar balance, dynamic stabilization tasks and multidirectional plyometric exercises be further investigated in preadolescent soccer players according to playing position.

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Communication

Antagonist Coactivation of Muscles of Ankle and Thigh in Post-Stroke vs. Healthy Subjects during Sit-to-Stand Task

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Abstract: This study aims to analyse the coactivation of antagonist muscles of the thigh and ankle during the sit-to-stand task in post-stroke subjects, specifically during forward and antigravity sub-phases. A group of 18 healthy subjects and another with 18 subjects with a history of stroke participated voluntarily in this study. Bilateral surface electromyography (EMGs) of the soleus, gastrocnemius medialis, tibialis anterior, rectus femoris and biceps femoris muscles were collected synchronously with ground reaction forces (GRF) during the sit-to-stand task. The magnitude of electromyographic (EMG) activity was analysed during forward translation and antigravity sub-phases which were determined through GRF signals. The coactivation was calculated to quantify the degree of antagonist coactivation according to the role of the muscles during the task. Statistically significant values were found between antagonist coactivation on both sub-phases of the sit-to-stand task when comparing healthy and post-stroke subjects (healthy with ipsilesional (IPSI); healthy with contralesional (CONTRA); and healthy with IPSI and with CONTRA limbs) in all muscle pairs analysed ($p < 0.01$), except on thigh muscles ($p > 0.05$), in the antigravity sub-phase. When comparing IPSI with CONTRA sides in post-stroke subjects, no statistically significant differences were found. Increased values of antagonist coactivation were observed in post-stroke subjects compared to healthy subjects (both IPSI and CONTRA limb) in the two sub-phases analysed. The forward sub-phase CONTRA limb showed higher antagonist coactivation compared to IPSI, while in the antigravity sub-phase, IPSI antagonist coactivation was higher than in the CONTRA. In conclusion, post-stroke subjects presented an antagonist coactivation more dysfunctional at the ankle joint muscles compared to the thigh segment. So, it seems that the distal segment could express more accurately the central nervous system dysfunction in post-stroke subjects, despite the need for further studies to achieve a better spatiotemporal understanding of the variability on coactivation levels.

Keywords: stroke; electromyography; antagonist coactivation; postural control; postural tone

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1. Introduction

Coordinated muscular activity is essential for postural control to achieve a dynamic interplay between postural orientation and stability. The complex synchronization of muscle activity in a coactivation pattern to provide antigravity postural control during centre-of-mass forward and upward translation has been demonstrated in the sit-to-stand (SitTS) task [1–6]. Muscle coactivation, its variation and its expression [7], is the phenomenon by which the central nervous system coordinates the activity of the antagonist muscle during an agonist muscle action, through the simultaneous activation of agonist and antagonist muscles around a joint [8–11]. The level of muscular coactivation, one part of postural tone expression [7], to be constantly adjusted to different tasks and the related phases depends on reticulospinal output due to their role on postural stability demands. Impaired motor control due to nervous system injury, such as that which occurs in strokes, shows increased levels of muscle coactivation particularly in relation to functional tasks and in body segments that assist in stability [12,13], and it has been expressed by high levels of antagonist coactivation [14].

In the SitTS task, the distal muscles in the leg assume a greater role in the control of ankle postural stability than do the thigh muscles [2,15]. Therefore, the assessment of this functional task would help to confirm the hypothesis that antagonist coactivation will result in more dysfunction at the distal level compared to the thigh level in post-stroke subjects. It can also be expected that both sides would present postural control impairment due to the bilateral disposal of the reticulospinal system [16,17] and its influence on neuromodulation of antagonist coactivation. The agonist role of anterior distal leg muscles in the forward translation sub-phase and its reversal role in the antigravity phase makes these two phases, in a more specific way, crucial to understanding the lack of neuromodulation to ensure antigravity postural stability in post-stroke subjects. Consequently, this study aims to analyse thigh and ankle muscles antagonist coactivation during the SitTS task in post-stroke subjects, specifically during forward and antigravity sub-phases.

2. Materials and Methods

2.1. Participants

A total of 18 post-stroke subjects with a first ischemic stroke episode were recruited for this study (age 71 ± 11.51 years, height 169 ± 9.10 cm, weight 53 ± 9.92 kg, 5 females, time post-stroke 26.7 ± 12.10 months, 10 with the contralesional limb (CONTRA) at the right). The inclusion criteria were: (1) the presence of a lesion in the territory of the middle cerebral artery at the sub-cortical level, confirmed by computerized axial tomography of the brain, (2) a score below 34 in the motor subsection of the Fugl-Meyer Assessment of Sensorimotor Recovery After Stroke Scale and (3) the ability to perform the SitTS sequence independently without losing stability. All subjects with previous history of other neurologic diseases, lower limb surgery or any orthopaedic or rheumatoid conditions that would interfere with the SitTS task were excluded. The 18 healthy subjects, age- and sex-matched, were compared with the control group (age 74 ± 11.81 years, height 168 ± 11.45 cm, weight 53 ± 12.41 kg, 5 females). These subjects were considered sedentary according to the Center for Disease Control for the American College Sports Medicine [18]. In both groups, subjects who did not have sufficient cognitive functioning to understand orders (assessment using the Mini-Mental State Examination) were excluded.

This study was approved by the local ethics committee of the Health School of Porto. All subjects gave their informed consent according to the Declaration of Helsinki.

2.2. Instruments

Ground reaction forces (GRF) were collected from one force platform (FP4060-108; Bertec, Columbus, OH, USA) connected to a BERTEC AM6300 signal amplifier. The bilateral EMG signals from the soleus (SOL), gastrocnemius medialis (GM), tibialis anterior (TA), rectus femoris (RF) and biceps femoris (BF) muscles were monitored using a wireless TrignoTM acquisition system (Delsys Inc., Natick, MA, USA). Pre-amplified bipolar differential electrodes (Trigno Avanti Sensor model) with a rectangular configuration of two Ag bars in parallel (inter-electrode distance of 10 millimetres) and a gain of 1000 were used to collect the EMGs signal at an acquisition frequency of 1000 Hz. EMGworks software was used to analyse the EMG signal quality. Skin impedance was measured with an electrode impedance checker (Noraxon, USA; Scotsdale, AZ, USA). The EMGs and force platform signals were analysed with MATLAB R2001a (Mathworks^{TM}, Natick, MA, USA).

2.3. Procedures

The skin of both IPSI and CONTRA limbs was prepared through standard procedures (shaving, removing dead skin cells and non-conductor elements with alcohol and with an abrasive pad) to reduce the electrical resistance to a level equal to or less than 5 KΩ, monitored with the electrode impedance checker, before electrode placement in the muscle's mid-belly according to anatomical references [19,20].

All individuals maintained a sitting position with 2/3 of the femur supported on the seat, whose height was adjusted to 100% of the lower leg length. Then, they were asked to stand up at a self-selected speed without using upper limbs, maintaining them comfortably along the body, or moving the feet [21]. At the end of the task, they remained quietly standing. A one-minute rest period was provided between each trial, and sufficient repetitions were performed to obtain three valid trials. In the post-stroke subjects' group, both lower limbs were analysed, while in the control group, just one lower limb was randomly selected.

The raw EMG signal was filtered with a band-pass filter (20 Hz–500 Hz), and the root mean square was calculated with a sliding window of 100 ms. The EMG signals have been normalized to the maximum voluntary contraction of the subjects' different muscles. The signal from the force plate was also filtered using a low pass filter of 10 Hz, and the values were normalized to the weight of each subject. The onset of the task, defined as time zero (T0), and the definition of forward translation and antigravity sub-phases were performed through GRF signals. These events were identified through the antero-posterior component of the GRF (FAP) and the vertical component of the GRF (FV) as represented by the schematic strategy illustrated in Figure 1 [3,22–24].

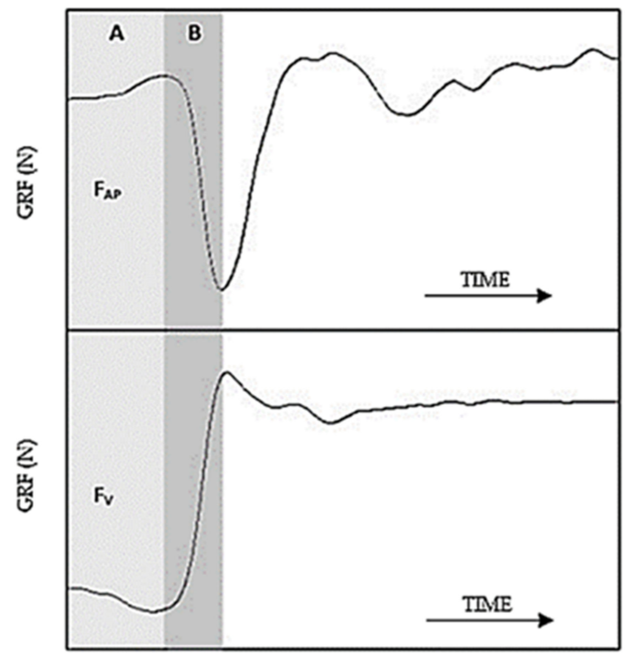

Figure 1. Illustrative identification of forward translation (A) and antigravity sub-phases (B) of the SitTS task through FAP and FV signals.

T0 was defined as the instant when the FAP value was greater or less than the mean of its baseline value plus 2 standard deviations (SD) for at least 50 milliseconds (ms). The forward translation sub-phase was identified in the interval between the end of the previous phase and the maximum value of the FV. As an example, Figure 2 demonstrates a force plate and EMG signals in a 2 s representative window of the SitTS task from two (healthy and stroke) analysed participants.

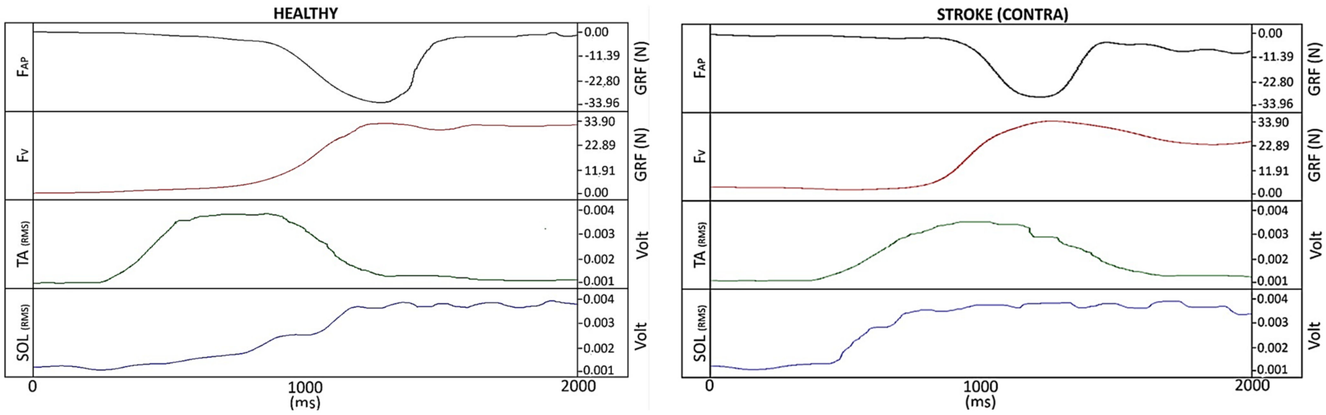

Figure 2. Healthy and Stroke 2 s window force plate and EMG signals. First signal corresponds to FAP, second to FV, third to TA EMG activity and, finally, fourth to SOL EMG activity.

In each sub-phase, the mean of the EMG signal was used to assess antagonist coactivation through the following Formula (1) [25]:

$$C\ (\%) = \text{antagonist activity}/\text{agonist} + \text{antagonist activity} \times 100 \quad (1)$$

Muscles were classified according to their role during the task. Specifically, during the forward translation sub-phase, the TA and BF muscles were considered the agonists, while in the antigravity sub-phase, this role was attributed to the SOL, GM and RF muscles. The TA muscle is likely to be the most representative for postural adjustments considering that it is activated early on to ensure foot stability and it assists in centre-of-mass forward displacement [4,26,27]. When acting in a synergistic pattern with coactivation, the more proximal muscles, despite their secondary role in the task [28], play an important role in thigh stability during trunk-forward translation and forward rotation of the tibia over the foot [29]. In the antigravity sub-phase, the SOL and GM muscles play an agonist role in body support and stability against gravity [30], while the proximal muscles behave as prime movers [27].

2.4. Statistics

The data were analysed using MATLAB R2001a (Mathworks^{TM}, Natick, MA, USA). To ensure that there were no significant differences between groups (stroke vs. healthy) regarding age, height and weight, the independent *t*-test was used. Since a normal distribution was not verified on all the coactivation variables, the Mann–Whitney test was used to compare antagonist coactivation levels between healthy and IPSI, healthy and CONTRA, and IPSI and CONTRA limbs. The Kruskal–Wallis test, with the Dunn-Bonferroni post hoc test, was applied to compare antagonist coactivation levels between healthy, CONTRA and IPSI limbs. A significance of 0.05 was considered for analysis.

3. Results

The characteristics of the participants are summarized on Table 1.

Table 1. Participants characteristics: mean and standard deviation (SD) values of age, height and weight of control and post-stroke groups, as well as side lesion and evolution time of the post-stroke group.

	Mean (SD)		***p*-Value [1]**
	Control Group	**Post-Stroke Group**	
Age (years) Height (cm) Weight (kg)	74 (11.81) 168 (11.45) 53 (12.41)	71 (11.51) 169 (9.10) 53 (9.92)	0.325 0.702 0.896
Gender	Female: $n = 5$ Male: $n = 13$	Female: $n = 5$ Male: $n = 13$	
Contralesional side	--	Left: $n = 8$ Right: $n = 10$	--
Time since stroke (months)	--	26.7 (12.10)	--

[1] Independent *t*-test.

No statistically significant values were found; therefore, both groups were comparable (Table 1).

Statistically significant values were found between antagonist coactivation on both sub-phases (A and B) of the SitTS task, when comparing healthy with IPSI, healthy with CONTRA, and healthy, IPSI and CONTRA in all muscle pairs analysed, except on thigh muscles (BF-RF), in the antigravity sub-phase (B). When comparing IPSI with CONTRA in post-stroke subjects, no statistically significant differences were found (Table 2).

Table 2. Median (MED), 25th (P25) and 75th (P75) percentiles of antagonist coactivation (%) in TA-SOL, TA-GM and BF-RF muscle pairs, in the forward (A) and antigravity (B) sub-phases of the SitTS task, in healthy, IPSI and CONTRA.

		Antagonist Coactivation (%)			***p*-Values #**			***p*-Values $**
	SitTS Sub-Phase	**Healthy MED (P25–P75)**	**IPSI MED (P25–P75)**	**CONTRA MED (P25–P75)**	**Healthy vs. IPSI**	**Healthy vs. CONTRA**	**IPSI vs. CONTRA**	**Healthy vs. IPSI vs. CONTRA**
TA-SOL	A	24.06 (19.33; 27.43)	51.91 (42.54; 61.35)	55.69 (51.11; 60.28)	<0.001 *	<0.001 *	0.335	<0.001 *
	B	34.68 (24.22; 44.50)	60.05 (53.17; 65.34)	53.54 (45.57; 64.33)	<0.001 *	<0.001 *	0.788	0.020 *
TA-GM	A	21.47 (15.09; 24.60)	47.81 (41.61; 51.33)	47.84 (46.82; 55.95)	<0.001 *	<0.001 *	0.289	<0.001 *
	B	42.22 (38.77; 48.42)	62.44 (55.92; 68.47)	58.87 (55.87; 63.20)	<0.001 *	0.010 *	0.211	<0.001 *
BF-RF	A	38.90 (31.95; 41.17)	53.08 (49.67; 54.86)	54.29 (50.10; 59.12)	<0.001 *	<0.001 *	0.275	<0.001 *
	B	40.95 (32.65; 50.25)	50.72 (40.03; 57.66)	43.82 (37.36; 65.98)	0.111	0.402	0.477	0.273

* Significative differences. # Mann–Whitney Test. $ Kruskal–Wallis Test.

It stands the fact that, increased values of antagonist coactivation were observed in post-stroke subjects compared to healthy, both on IPSI and CONTRA, in the two sub-phases analysed (Figure 3). In the forward sub-phase (A), CONTRA showed higher antagonist coactivation compared to IPSI, while in the antigravity sub-phase (B), IPSI antagonist coactivation was higher than CONTRA (Figure 3).

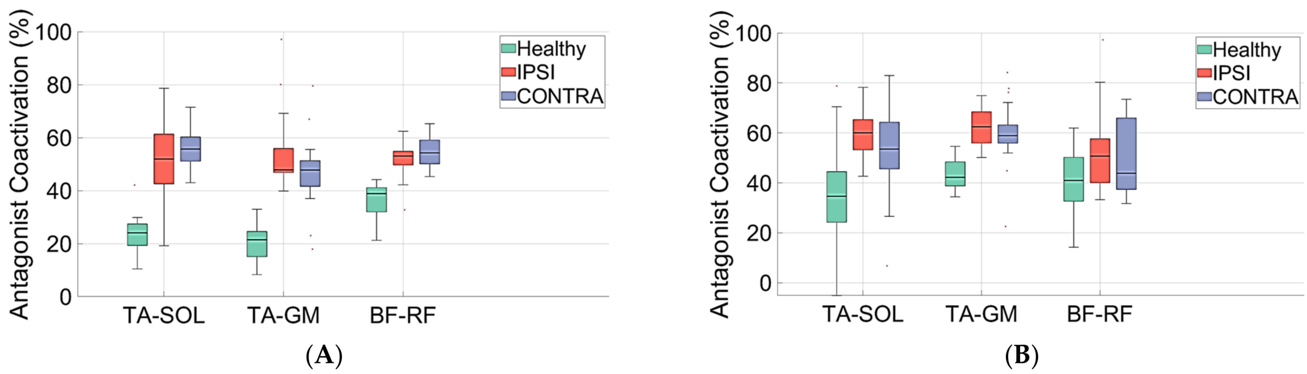

Figure 3. Antagonist coactivation (%) in healthy and post-stroke subjects (CONTRA and IPSI) in both forward (**A**) and antigravity (**B**) sub-phases of the SitTS task in ankle (TA-GM and TA-SOL) and thigh (RF-BF).

4. Discussion

Ankle (TA-SOL and TA-GM muscle pairs) and thigh (RF-BF muscle pairs) antagonist coactivation in forward translation and antigravity sub-phases in the SitTS task in post-stroke subjects exhibit a specific trend toward higher antagonist coactivation. This finding agrees with previous studies involving walking, specifically in the double-support phase of walking [31]. On both, in SitTS and in the double-support phase of walking, antagonist coactivation observed in the two lower limbs seems to reflect a bilateral postural control dysfunction in these subjects. The influence of supra-spinal structures on antagonist coactivation [32] reinforces the value of this synergy in joint stability [33] through precise control of joint position [15].

Previous studies have demonstrated that ankle muscles play a determinant role in influencing proximal muscles activity for stability in both healthy [34,35] and post-stroke subjects [36,37]. Like in walking, SitTS involves sub-phases with specific coactivation patterns across joints and the role of ankle muscles in stability has also been highlighted in elders [38]. The results obtained in the present study showed that it is possible to explore post-stroke subjects' incapacity in performing the SitTS task [39] based on the impairment in the antagonist muscle coactivation modulation against the environment. The results also revealed that during the antigravity sub-phase, IPSI presented the most relevant alteration of antagonist coactivation. These findings are in accordance with other studies in SitTS muscle synergies [37], and also during double support in walking [31], and may be related to the ipsilateral postural control dysfunction often evidenced in lesions affecting the middle cerebral artery territory [26]. In this sub-phase, the plantar flexors act as agonists for stability, and, particularly, the motoneurons of the SOL are more dependent on output from the ventromedial systems than from the reticulospinal system [26,40,41]. Given these assumptions, it is important to think about the mechanisms underlying IPSI vs. CONTRA weakness, while suggesting that IPSI presents the most relevant alteration in antagonist coactivation. Future studies could improve the clarity regarding this area.

Globally, the more variable pattern of ankle muscles antagonist coactivation compared to the thigh in both IPSI and CONTRA legs in post-stroke subjects seems to suggest that the distal ones may present with more accuracy in the postural control impairment in both sides of these subjects when performing functional tasks with higher postural control demand.

However, as no significant differences were observed in the antigravity sub-phase, in the antagonist coactivation of the thigh, it may be thought that during double support, similarly to studies carried out in gait, the role of ankle coactivation for stabilization during movement may have a major influence on the proximal adjustment [34,36]. This might explain the differences obtained where a higher performing postural control demand was expressed. The non-existence of proximal statistically significative differences may demonstrate the lower requirement of postural performance during the task when compared to distal, which was observed in the results. It would be interesting to complement the linear analysis performed with the combination of a non-linear treatment of the variables under study to explore their behaviour. This might show the trend of existing behaviours in a more detailed spatiotemporal way, describing the variability inherent in human movement in order to better characterize it [11]. The variations expressed during coactivations may also suggest improvement in knowledge regarding the biomechanical characteristics of postural tone and its evaluation [7]. Although considering that all post-stroke subjects analysed presented a lesion in the territory of the middle cerebral artery at the sub-cortical level, the correspondence between coactivation mechanisms and the impairment in specific neurophysiologic regions and related pathways cannot be confirmed in the present study. Future studies are required to confirm this association.

5. Conclusions

Considering that the values of the muscle antagonist coactivation demonstrated a bilateral increased ankle (TA-SOL and TA-GM) antagonist coactivation in both forward and antigravity sub-phases of SitTS tasks in post-stroke subjects, it seems possible to confirm the hypothesis that post-stroke subjects present an antagonist coactivation more dysfunctional distally compared to the thigh segment (BF-RF). So, the distal segment could more accurately express the central nervous system dysfunction in post-stroke subjects despite the need for further studies to achieve a better spatiotemporal understanding of the variability of coactivation levels.

Author Contributions: The research authors provided the following contributions: conceptualization, A.S. and A.S.P.S.; methodology, A.S. and A.S.P.S.; software, L.P. and F.P.; validation, L.P.; formal analysis, L.P.; investigation, L.P.; resources, L.P.; data curation, L.P.; writing—original draft preparation, L.P. and A.S.; writing—review and editing, L.P., A.S.P.S., C.S., C.C., R.S., J.M.R.S.T., S.P., A.R.P., J.F., F.P., F.S. and A.S.; visualization, L.P.; supervision, A.S.P.S., A.S. and F.S.; project administration, A.S.; funding acquisition, A.S.P.S. All authors have read and agreed to the published version of the manuscript.

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Institutional Review Board Statement: The study was conducted in accordance with the Declaration of Helsinki and approved by the Ethics Committee of ESCOLA SUPERIOR DE SAÚDE DO POLITÉCNICO DO PORTO (protocol code 1484, 10 April 2018).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The data presented in this study are available on request from the corresponding author. The data are not publicly available due to privacy or ethical restrictions.

Conflicts of Interest: The authors declare no conflict of interest.

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Assessing Isometric Quadriceps and Hamstring Strength in Young Men and Women: Between-Session Reliability and Concurrent Validity

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Abstract: This study assessed the between-session reliability of the FysioMeter H-station when measuring isometric quadriceps and hamstrings strength and H/Q (hamstring/quadriceps) ratio, and the concurrent validity when compared to an isokinetic dynamometer (ID). Sixteen young males and three females (25.6 ± 1.7 years old) performed maximum quadriceps and hamstring contractions using the H-station and the Humac NORM ID over two sessions. Between-session reliability was evaluated by comparing scores from the first and the second session. Concurrent validity was assessed by comparing scores from the H-station with the Humac NORM ID. The H-station showed excellent relative reliability for the quadriceps, hamstring, and the H/Q ratio with ICC2.1 ranging from 0.89 to 0.91. The relative reliability of the Humac NORM was good to moderate for the quadriceps, hamstring, and the H/Q ratio with ICC2.1 ranging from 0.89 to 0.91. Acceptable absolute reliability of the H-station was found across quadriceps, hamstring, and H/Q measures (coefficient of variance (CV) = 7.2 to 9.3%, limit of agreement (LOA) = 20.0 to 24.3%). The validity of the H-station was large for hamstring strength (r = 0.79) and moderate for quadriceps strength (r = 0.69) and H/Q ratio (r = 0.39) compared to the Humac NORM ID. Bland–Altman plots showed LOAs ranging from 37.9 to 59.5%. The results indicate that the H-station can be used to make reliable assessments of relative changes in maximum isometric quadriceps and hamstring strength.

Keywords: reproducibility; H/Q ratio; knee peak torque; hamstring injuries; validity

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1. Introduction

According to a 2016/17 season's UEFA Elite Club Injury Study Report, most injuries in professional football occur in the thigh and knee area [1]. The report further shows that the most common type of injury is muscle rupture, strain, or cramps, representing 45% of all injuries in professional football [1]. Muscle strains in hamstrings accounts for 12–16% of all injuries among football players and have high recurrence rates (14 to 62%) [2–4]. Among professional football clubs, a typical hamstring injury will result in an average of 17 days of absence from training and matches with an average cost of EUR 16.666 per day that the player is unavailable to the team [5].

Several modifiable risk factors for muscle injuries have been identified, including fatigue, high-speed running loads, overall muscle strength, and inter- and intralimb asymmetry [6].

Asymmetry in both the inter- and intralimb has been proposed to increase the risk of sustaining a hamstring strain injury [6]. Interlimb asymmetry has typically been assessed using the bilateral Nordic Hamstring Exercise, but according to Cuthbert et al., this exercise is not necessarily the best way to evaluate hamstring strength and interlimb imbalances [7]. The authors stated that in bilateral exercises, a compensatory strategy is typically adopted and can shift between test sessions. One way of minimizing the risk of compensatory strategies is using a unilateral exercise as an alternative. Furthermore, regarding the intralimb asymmetry, the difference between the quadriceps and hamstrings strength (H/Q ratio) has been identified as particularly important [8,9]. Since the assessment of muscle strength is seen as a key risk factor for preventing injuries in the thigh area, objective measurements with high reliability and validity are essential [10]. Currently, stationary isokinetic dynamometers (ID) have proven to be both reliable and valid and are considered the "golden standard" [11]. However, IDs are expensive, time- and space-consuming, and often difficult to operate, warranting further development and investigation of more practically applicable measurement methods.

Previous investigations have evaluated the reliability and validity to measure peak force in different joints less expensive and with the use of portable measuring equipment, such as field-based tests [12] and handheld dynamometers [10,13–15]. Handheld dynamometers are generally known to be a reliable measure of muscle strength; however, a systematic review highlights that concurrent validity for handheld dynamometers varies in terms of which joint is measured [13]. Correlations between handheld dynamometers and ID have been found to be very high (ICC from 0.73 to 0.98) for hip abductors/adductors and hip flexors/extensors [16,17] and moderate (ICC from 0.37 to 0.91) for ankle plantar/dorsiflexion and knee flexors/extensors [16–20]. Even though handheld dynamometers are less expensive and portable, they still require a strong and experienced practitioner to obtain reliable and valid measures [21].

Recently, a novel device called an H-station (FysioMeter, Aalborg, Denmark) has been developed with the purpose of monitoring athletes' quadriceps and hamstrings strength, using two Nintendo Wii balance boards (WBB). The equipment is cheaper, easier to use, and portable compared to the ID and may therefore be more useful for in-season monitoring of players. However, no studies have investigated the between-session reliability and concurrent validity of the H-station compared to the ID and its potential to quantify the H/Q ratio.

Therefore, the purposes of this study were: (a) to determine between-session reliability of the H-station when measuring isometric quadriceps and hamstrings strength and H/Q ratio, and (b) to determine concurrent validity of these measures when comparing to a golden standard ID (Humac NORM, CSMi, Stoughton, MA, USA).

2. Materials and Methods

2.1. Experimental Approach to the Problem

Measurements were performed on two separate occasions with exactly seven days apart at Aalborg University, Aalborg, Denmark between 1/10-22 and 1/12-22. All tests were conducted by the same rater (F.H.M.). The study followed a counterbalanced testing pattern so half of the participants started in the H-station and the other half started in the ID (Figure 1). The same order of measurements was followed on both test occasions. Measurements were performed at the same time of the day to avoid the circadian rhythm influencing the measurements [22]. All participants were tested on their dominant leg, defined as the leg they prefer kicking a ball with. They were instructed to wear the same shoes at both testing days. At the initial testing session, descriptive characteristics of the participants were collected, including height, body mass (kg), body fat percentage (%), muscle mass (kg), and length of lower leg, (from lateral femoral epicondyle, estimated knee axis of rotation, to the posterior part of the sole of the foot). Body composition was measured with a bioimpedance apparatus (InBody 270, Biospace, San Francisco, CA, USA).

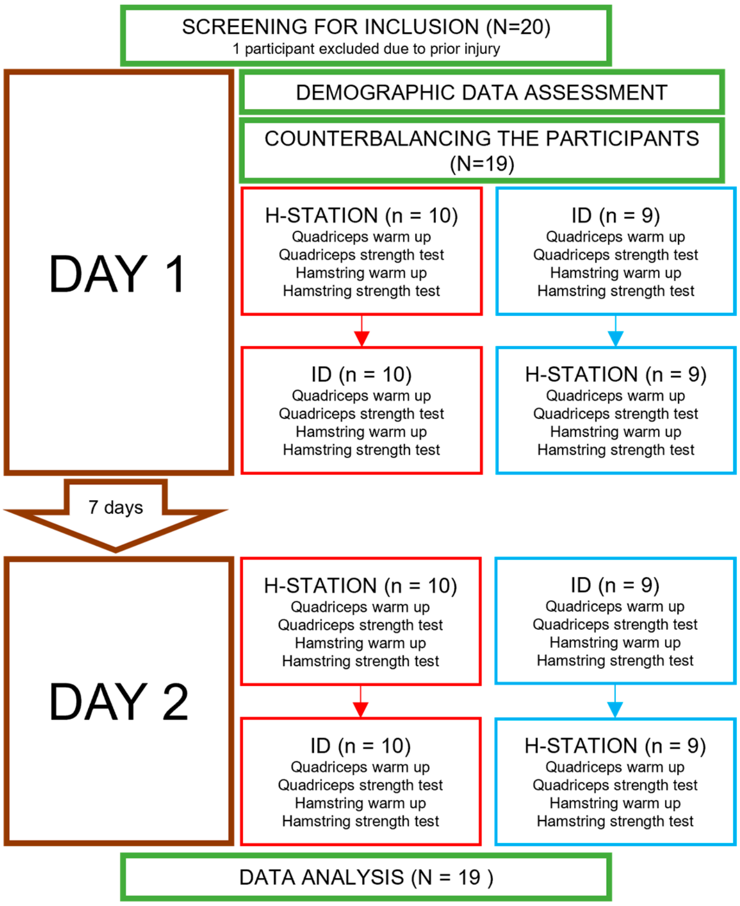

Figure 1. Flowchart of recruitment and testing order of participants. One participant was excluded due to a lower extremity injury not related to their participation in the current study. ID: isokinetic dynamometer.

The study followed the guidelines for reporting reliability and agreement studies (GRRAS) [23]. The study was conducted in accordance with Danish Legislation and the Helsinki Declaration. The North Denmark Region Committee on Health Research Ethics (LBK nr. 1083) was contacted, and the study was granted exemption from requiring ethics approval. All participants provided written informed consent before any study activities were initiated.

2.2. Participants

A convenience sample of 20 asymptomatic and physically active (>3.5 h a week) participants (17 males and 3 females) were recruited from Aalborg University (Aalborg, Denmark). Participants were excluded if they reported one or more of the following criteria: (1) a history of traumatic spine or lower extremity injury within the past three months, (2) pain in spine or lower extremity, (3) strenuous exercise within the last 24 h before testing, and (4) caffeine or nicotine intake within the last 8 h. A short questionnaire was completed to ensure none of the exclusion criteria was violated. An a priori sample size calculation was not conducted, as there were no established data to do so.

2.3. Procedures

2.3.1. Procedure in the H-Station

Participants were tested in an H-station (FysioMeter, Aalborg, Denmark) using a standard Nintendo WBB (Nintendo, Kyoto, Japan) connected to a standard PC. Initially, participants performed a short quadriceps-specific warm-up followed by the isometric quadriceps strength test (Quad-H). Then, they undertook a short hamstring-specific warm-up before the isometric hamstring strength test (Ham-H). The warm-up consisted of four isometric sub-maximal followed by one maximal contraction, in accordance to procedures in previous studies [24–26]. Besides improving performance and preventing injuries, the warm-up also served the purpose of familiarizing the participants to the tests. Following warm-up, the participants had two minutes of rest before the test to minimize the potential effect of fatigue.

For the Quad-H test, participants were seated on a standard treatment table with their hips flexed approximately at 85 degrees. The thigh was secured to the treatment table with two Velcro® straps placed at both proximal and distal part of the thigh (Figure 2a). The H-station was then placed as close to the participant as possible, and participants were instructed to place the tip of their shoe touching the marked center of the WBB. Then, the tester adjusted the height of the treatment table to align the sole of the shoe parallel to the floor. The height of the treatment table and knee angle, measured with a goniometer, were noted and reproduced for the second testing occasion. To secure that the H-station did not move during testing, a weight plate of 25 kg was placed in front of the H-station with the tester standing on the weight plate and securing both the top and bottom of the H-station (Figure 2a). The Quad-H test consisted of three sets of one isometric maximal knee extension, with one minute of rest between each set. Participants were instructed to keep their arms crossed over their shoulders and to push "as fast and hard as possible" with their toes and to hold the contraction for 3 s. Before the Ham-H test, a two-minute rest period was given to the participants.

(**a**) (**b**)

Figure 2. Experimental setup for the quadriceps strength test (Quad-H) and hamstring strength test (Ham-H) in the H-station. (**a**) General setup for the Quad-H test with a 25 kg weight plate and tester securing the H-station during testing. (**b**) General setup for the Ham-H test.

For the Ham-H test, the H-station was rotated 180 degrees horizontal to the treatment table (Figure 2b). Thereby, the WBB was positioned posterior to the participant's feet and the dominant knee of the participant was touching the knee padding of the H-station. The participant's thigh was secured using two straps, similar to the Quad-H test. The Ham-H test followed the same procedure as the Quad-H test, except participants were instructed to perform a knee flexion.

For both the Quad-H- and the Ham-H test, verbal encouragement, and real-time visual biofeedback of the force trace were provided to ensure maximum effort during each repetition [27]. The trial was repeated if a counter movement was detected to ensure a contraction starting from rest, or if one of the three repetitions varied with more than 20% from maximum peak value.

Force data were collected via one Nintendo WBB with four strain gauge transducers positioned at each of the four corners. Data from the H-station were sampled at 100 Hz from each transducer and were transferred via Bluetooth to FysioMeter's software version 5.0.1 on a tablet. Subsequently, FysioMeter software filtered data by using a 4th-order Butterworth lowpass filter with a cutoff frequency of 20 Hz. The FysioMeter software calculates one variable: absolute maximum force measured in kg.

2.3.2. Procedure in the ID

Participants were tested in a Humac NORM ID (CSMi, Stoughton, MA, USA) connected to a Windows laptop with Humac NORM software installed (HUMAC 2009, v.9.7.1). Participants performed a short warm-up followed by the isometric strength test, initially for the quadriceps (Quad-I) and then, the hamstring (Ham-I). The warm-up consisted of four isometric sub-maximal contractions followed by one maximal knee contraction. Following warm-up, the participants had two minutes of rest before the test.

For both the Quad-I and the Ham-I test, participants were seated with their hips flexed at 85 degrees and their thighs and torso were secured using a strap and a safety belt. An ankle strap was placed five centimeters proximal to the distal aspect of the lateral malleolus. The mechanical lever arm of the ID was aligned with the lateral epicondyle of the knee. The knee angle was adjusted to the same angle as in the Quad-H and Ham-H tests, respectively.

After the warm-up was performed, participants completed the Quad-I test, which consisted of three sets of one maximal isometric knee extension lasting three seconds, with one minute of rest in between each set. After the Quad-I test, a two-minute rest period was given to the participants before the Ham-I test. After the rest period, the participants completed the Ham-I test. The procedure for the Ham-I test was the same as Quad-I except participants were instructed to perform a knee flexion. For both the Quad-I and Ham-I tests, verbal encouragement and real-time visual biofeedback of the torque signal were provided.

2.4. Data Processing

To compare the two measuring devices, the outcome values from the H-station, peak force in kg, were converted to maximum quadriceps and hamstring torque (Nm), respectively. For the Quad-H test, torque was calculated by first converting kg into Newton (N) by multiplying kg with gravitational acceleration (9.81 m/s^2). Secondly, the lever arm perpendicular to the force vector applied on the WBB was calculated and finally the quadriceps torque was calculated by multiplying the applied force with the lever arm.

Maximum hamstring torque for the Ham-H test was calculated by first converting kg values to N. Since knee angle for the Ham-H test was 90 degrees, torque was calculated by multiplying the force applied on the WBB with the lever arm being the length of the lower leg.

Data processing for the ID was performed by using a custom-made MATLAB version R2021a (MathWorks, Natick, MA, USA) script. Maximum torque values were extracted for each maximal isometric knee extension and flexion.

The mean for the maximum quadriceps and hamstring torque was calculated, respectively, for all four tests on both session 1 and 2 and was used for further statistical analysis. H/Q ratio was calculated by dividing maximum mean hamstring torque by the maximum mean quadriceps torque.

2.5. Statistical Analysis

Statistical analysis was completed using SPSS version 27 (IBM, Chicago, IL, USA). p-values ≤ 0.05 were interpreted as significant. All torque values are presented in Nm. Histograms and Shapiro–Wilks test of normality ($p > 0.05$) were used to test for normal distribution for the maximum mean quadriceps and hamstring torque and for the H/Q ratio measures. All outcomes met the assumption of normal distribution.

As recommended, both relative and absolute reliability were reported [28]. The relative reliability was quantified using an $ICC_{2.1}$ two-way mixed model using absolute agreement and the corresponding 95% confidence intervals (CI 95%) were also calculated for the ICC values. The ICCs were interpreted using the following criteria: poor (<0.5), moderate (0.5–0.75), good (0.75–0.90), and excellent (>0.90) [3].

Absolute reliability was investigated with Bland–Altman plots with 95% limits of agreement (LOA) represented in absolute values and in percentage, standard error of measurement (SEM), coefficient of variance (CV), and minimal detectable change (MDC). Lastly, a paired sample *t*-test was used to test if a systematic bias was present between session 1 and session 2 for the Quad-H, Ham-H, Quad-I, and Ham-I tests.

The concurrent validity of the H-station compared to the ID was investigated using the Pearson product moment correlations (Pearson r), and Bland–Altman plots with 95% LOA. The Pearson r correlation values were interpreted using the following criteria: ±0.0–0.09 indicates no correlation, ±0.1–0.3 indicates a small correlation, ±0.3–0.5 indicates a moderate correlation, and ±0.5–1.0 indicates a large correlation [14].

3. Results

Of the 20 individuals screened, 19 participants were included in the analysis (16 males, 3 females). Table 1 displays all demographic data.

Table 1. Participants' demographic data (N = 19). Values are mean ± standard deviation.

Characteristics	Participants
Age (years)	25.6 ± 1.7
Height (cm)	182.7 ± 7.6
Body mass (kg)	81.4 ± 13.2
BMI (kg/m^2)	24.3 ± 3.2
Muscle mass (kg)	38.2 ± 6.3
BFP (%)	17.1 ± 7.5
Lower leg length (cm)	50.6 ± 3.1
Knee angle (degrees) [1]	70.8 ± 3.4

Abbreviation: BMI: body mass index. BFP: body fat percentage. [1] Knee angle measured for the quadriceps H-station test.

3.1. Between-Session Reliability

Table 2 reports the descriptive statistics for the H-station and ID between-session reliability data for quadriceps and hamstrings strength tests and H/Q ratio. The Quad-H and Ham-H tests showed excellent reliability (ICC: 0.91 for both tests) while the H/Q-H had a good reliability (ICC: 0.89). The relative reliability for the Quad-I- and Ham-I tests was good (ICC of 0.80 to 0.89) while it was moderate for the H/Q-I (ICC: 0.65).

For the H-station recordings, SEM and CV between sessions were 22.5 Nm and 8.8% for Quad-H, 10.4 Nm and 7.2% for Ham-H, and 0.05 and 9.3% for H/Q-H. For the ID recordings, SEM and CV between sessions were 18.6 Nm and 6.5 for Quad-I, 17.0 Nm and 11.7% for Ham-I, and 0.06 and 12.4% for H/Q-I.

Figure 3 displays the Bland–Altman plots with LOA, investigating absolute reliability of the H-station and ID. For the Quad-H test (plot A), Bland–Altman displayed a mean difference of 14.5 Nm (p = 0.063) with LOA% of 24.4%. For the Ham-H test (plot B), a significant mean difference of 11.4 Nm (p = 0.049) with LOA% of 20.0% was found. For H/Q-H (plot C), a mean difference of 0.01 (p = 0.768) with LOA% of 22.6% was found. A mean difference of 10.2 Nm (p = 0.108) and 5.4 Nm (p = 0.344) and a LOA% of 17.9 and 32.4% was found for the Quad-I and Ham-I tests, respectively. For the H/Q-I, a mean difference of 0.04 (p = 0.082) and LOA% of 29.4% was found.

MDC was 62.5 Nm for Quad-H, 28.8 Nm for Ham-H, 0.15 for H/Q-H, 51.5 Nm for Quad-I, 47.1 Nm for Ham-I, and 0.17 for H/Q-I.

Table 2. Descriptive statistics and relative and absolute reliability data for the H-station and the ID (N = 19).

	Session 1		Session 2		Diff ± SD	ICC (95% CI)	95% LOA	LOA (%)	SEM	CV (%)	MDC
	Mean	**SD**	**Mean**	**SD**			**(lb–ub)**				
H-station recordings											
Quad-H (Nm)	249.5	±81.3	263.9	±76.1	14.5 ± 31.9	0.91 (0.76 to 0.96)	(−76.9–48.0)	24.3	22.5	8.8	62.5
Ham-H (Nm)	138.2	±43.2	149.6	±43.4	11.4 * ± 14.7	0.91 (0.64 to 0.99)	(−40.2–17.4)	20.0	10.4	7.2	28.8
H/Q-H	0.57	±0.17	0.58	±0.15	0.01 ± 0.08	0.89 (0.73 to 0.96)	(−0.15–0.14)	22.6	0.05	9.3	0.15
ID recordings											
Quad-I (Nm)	281.9	±54.2	292.1	±49.3	10.2 ± 26.3	0.86 (0.67 to 0.94)	(−61.7–41.3)	17.9	18.6	6.5	51.5
Ham-I (Nm)	148.1	±41.8	142.7	±34.0	5.4 ± 24.0	0.80 (0.56 to 0.92)	(−41.7–52.5)	32.4	17.0	11.7	47.1
H/Q-I	0.53	±0.12	0.49	±0.09	0.04 ± 0.09	0.65 (0.29 to 0.85)	(−0.14–0.21)	29.4	0.06	12.4	0.17

Abbreviation: Quad-H: quadriceps strength test in H-station. Ham-H: hamstring strength test in H-station. Quad-I: quadriceps strength test in isokinetic dynamometer. Ham-I: hamstring strength test in isokinetic dynamometer. Diff: mean difference between sessions (Nm). SD: standard deviation of mean difference (Nm). LOA: limits of agreement (lb: lower boundary—ub: upper boundary) (Nm). ICC: intra-class correlation coefficient. CI: 95% confidence interval. SEM: standard error of measurement (Nm). CV: coefficient of variance (%). MDC: minimal detectable change (Nm). * Significant difference $p < 0.05$.

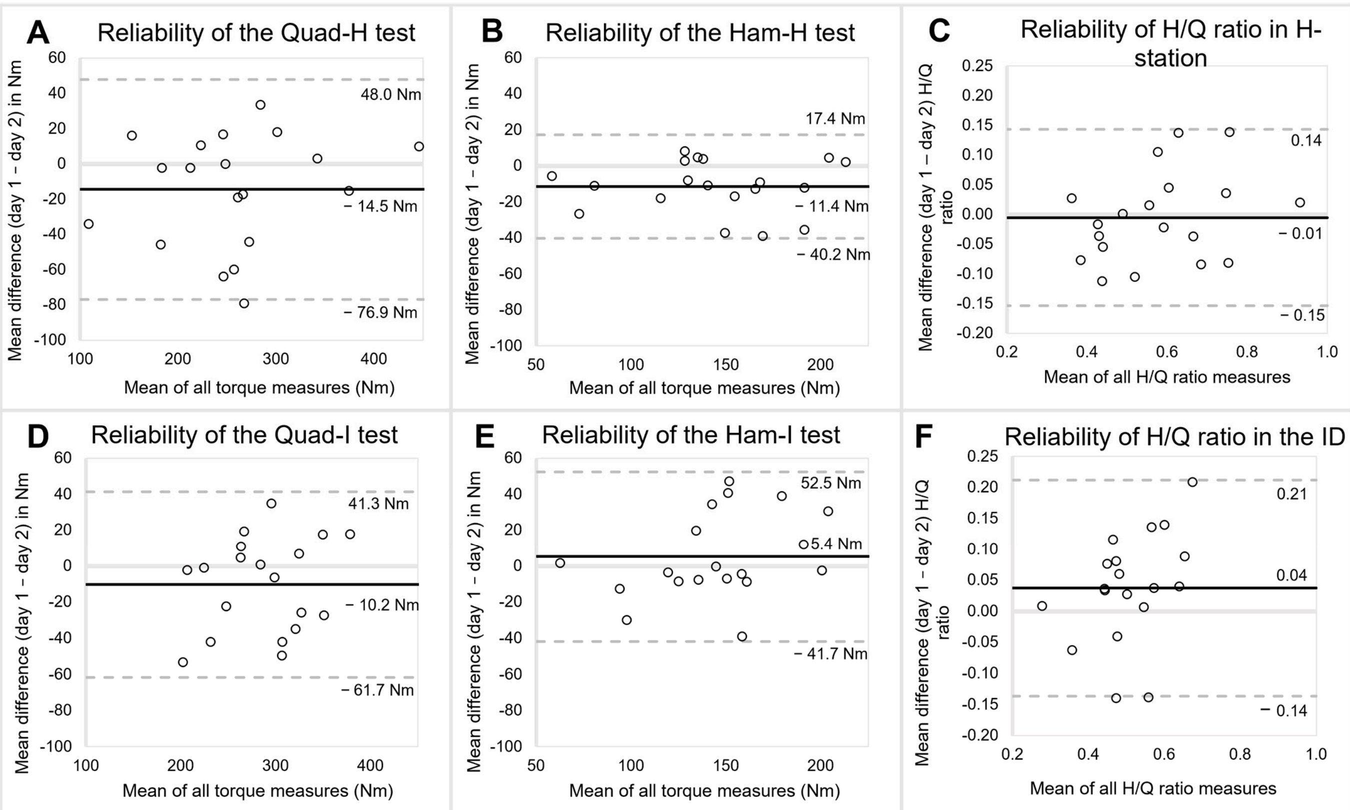

Figure 3. Bland–Altman plots displaying absolute reliability for (**A**) the quadriceps strength test (Quad-H) in the H-station. (**B**) The hamstring strength test (Ham-H) in the H-station. (**C**) The hamstring/quadriceps ratio (H/Q-H) for the H-station. (**D**) The quadriceps strength test (Quad-I) in the isokinetic dynamometer. (**E**) The hamstring strength test (Ham-I) in the isokinetic dynamometer. (**F**) The hamstring/quadriceps ratio (H/Q-I) for the isokinetic dynamometer. Dotted gray lines represent the limits of agreement and solid black lines represent the mean difference between sessions.

3.2. Concurrent Validity

Table 3 reports the correlations between the H-station and the ID for each condition and Figure 4 displays the Bland–Altman plots with LOA. Maximum hamstring torque in the H-station showed a large correlation with the ID with a Pearson r value of 0.79. A moderate correlation was found for the maximum quadriceps torque with Pearson r value of 0.69. The correlation of H/Q ratio between the H-station and the ID showed a moderate correlation with Pearson r value of 0.37.

Table 3. Descriptive statistics and concurrent validity of the H-station compared with the Humac NORM ID (N = 19) from the first session.

	H-Station		ID		Diff	Pearson r	95% LOA	LOA (%)
	Mean	SD	Mean	SD			(lb–ub)	
Quad (Nm)	249.5	±81.3	281.9	±54.2	32.4 *	0.69	(−147.6–82.7)	43.3
Ham (Nm)	138.3	±43.2	148.1	±41.8	9.9	0.79	(−64.2–44.5)	37.9
H/Q ratio	0.58	±0.17	0.53	±0.12	0.05	0.37	(−0.28–0.38)	59.5

Abbreviation: Quad: maximum quadriceps torque (Nm). Ham: maximum hamstring torque (Nm). ID: isokinetic dynamometer. Diff: mean difference (Nm). LOA: limits of agreement (lb: lower boundary—ub: upper boundary) (Nm). * Significant difference $p < 0.05$.

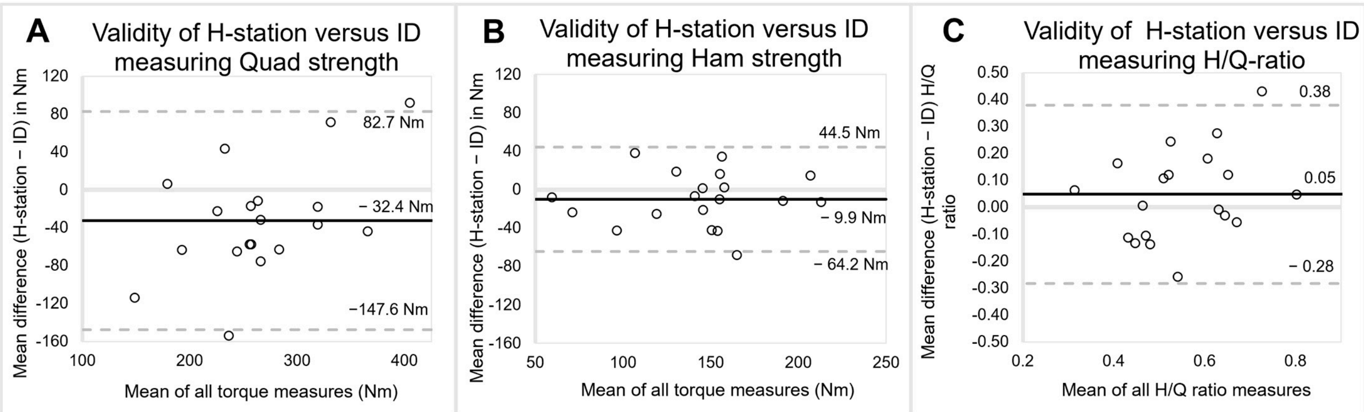

Figure 4. Bland–Altman plot (**A**) concurrent validity of H-station compared to the isokinetic dynamometer measuring maximum quadriceps strength. (**B**) Concurrent validity of H-station compared to the isokinetic dynamometer measuring maximum hamstring strength. (**C**) Concurrent validity of H-station compared to the isokinetic dynamometer measuring hamstring/quadriceps ratio. Dotted gray lines represent the limits of agreement and solid black lines represent the mean difference between devices.

Bland–Altman plots displayed a significant systematic bias of 32.4 Nm ($p = 0.027$) with LOA and LOA% of 115.1 Nm and 43.3% for the maximum quadriceps torque (plot A). A mean difference of 9.9 Nm ($p = 0.138$) with LOA and LOA% of 54.3 Nm and 37.9% were found for the maximum hamstring torque (plot B). A mean difference of 0.05 ($p = 0.228$) with LOA and LOA% of 0.33 and 59.5% were found for the H/Q ratio measures (plot C).

4. Discussion

This study evaluated (i) the between-session reliability and (ii) the concurrent validity of quadriceps and hamstring maximum strength and the associated ratio (H/Q-ratio) on a novel, portable, and easy to use alternative to the golden standard IDs. The FysioMeter H-station exhibited good to excellent relative between-session reliability for the quadriceps, hamstring, and H/Q-ratio measures. Furthermore, we found an acceptable absolute reliability for all the above measures. Lastly, compared to the ID, the quadriceps test showed a high correlation, while the hamstring and the H/Q showed a moderate correlation.

4.1. Between-Session Reliability

The absolute maximum torque values obtained by the H-station in the present study were very similar to findings of studies investigating a similar population [24,26,29]. Several studies have investigated the reliability of a handheld dynamometer (HHD) in relation to quadriceps and hamstring strength, but only a few have expressed the absolute values in Nm. Instead, most of the studies have used kg or N to express their absolute values obtained by HHD, which makes it difficult to compare the results obtained by the current study.

In the current study, the H-station was found to be highly reliable for measuring both maximum quadriceps and hamstring torque (ICC = 0.91 for both). Furthermore, the reliability of the H-station measuring H/Q ratio was good (ICC = 0.89). One systematic bias was found between the mean of session 1 (138.2 Nm) and session 2 (149.6 Nm) for the Ham-H test, which could indicate a learning effect of the test [30]. Only one study has previously evaluated the reliability of the Nintendo Wii balance boards (WBB) for measuring lower limb strength in older adults [25]. This previous study found very similar reliability values compared to the current study, with ICC values for WBB ranging from 0.91 to 0.97, SEM from 9.7 to 13.9% and LOA ranging from 20.3 to 28.7%. Other portable strength measuring devices, such as HHD's, have shown ICC values between 0.86 and 0.96, SEM/CV between 4.2 and 14.7% and LOA (%)/MDC between 11.6 and 24.9% [17,18,29,31]. For the isometric hamstring strength, previous studies found ICC's between 0.89 and 0.96, SEM/CV between 4.8 and 8.6% and LOA/MDC between 13.3 and 23.8 [17,29]. Our results for both relative and absolute reliability were within the same ranges One study has evaluated the HHD's ability to measure H/Q ratio [29]. This latter study found ICC values ranging from 0.87 to 0.90, which corresponds to our study's ICC (ICC = 0.89) for H/Q ratio of the H-station.

The SEM values reported here represent variation in torque due to three overall factors: instrumental-, biological- and/or experimental protocol variations [30]. Instrumental variation refers to the error or noise in the measuring equipment used. In relation to measuring error, one study has found excellent correlations (ICC = 0.83–0.99) of WBB compared to a laboratory-based force plate measuring force during a dynamic task [32]. This finding implies that the WBB is a reliable tool for measuring force in dynamic tasks, but it does not eliminate the possibility of variation due to signal noise or calibration errors. When testing human performance, random variations are usually caused by biological- and experimental protocol differences [25,30]. Among biological factors, changes in mental and physical states between session 1 and 2 can account for some of the random variation [13]. The mental state was attempted controlled by providing verbal encouragement and visual bio feedback, which are suggested to have a positive influence on participants' physical performance [27]. It was not possible to control completely the participants' physical state, thus it is possible that the biological variations could account for the relatively high between subject variance in the data.

4.2. Concurrent Validity

In the present study, the maximum quadriceps torque measured with the ID was significantly higher than the maximum quadriceps torque measured with the H-station. These results point towards a similar tendency as previous studies investigating concurrent validity of HHD compared to an ID reported ICC values ranging from 0.42 to 0.85 for isometric quadriceps strength [17–20] and 0.66 to 0.91 for isometric hamstring strength [17,19].

The paired sample *t*-test showed that the H-station significantly underestimated the maximum quadriceps torque values with a mean difference of 32.4 Nm (SD = 58.7 Nm). This underestimation can be explained due to the manual calculation of torque values for the H-station. This calculation is only valid on the premise of the foot contact with WBB being perpendicular. In case of dorsiflexion of the ankle joint during the isometric knee extension, the force applied on the WBB would not have acted perpendicularly resulting in an erroneous lever arm. This could introduce a measurement error that may have contributed to the underestimation of the maximum quadriceps torque measured in the H-station.

Aside from the previous discussed instrumental and/or biological variations causing random error, the current experimental protocol had limitations, which could account for some of the variation within and between participants and the low correlation between devices. The Quad-H test depended on the practitioner's ability to maintain the H-station in place during the tests. Chamorro et al. highlight that a higher reliability of HHD can be achieved when the practitioner is stronger than the participant [13]. In addition, this aspect is especially important when testing large muscle groups like quadriceps. Movement of the H-station during the tests would result in lower maximum quadriceps or hamstrings torque. The experimenter noticed that the H-station did move during a few tests. This was compensated by simply repeating the specific test. For the Quad-H test, some participants reported discomfort (tip of the toes) during the isometric knee extension. This may have affected the motivation of the participant to produce a maximal and consistent knee extension. Furthermore, all participants wore different types of shoes during testing. It is uncertain if the shoe type influences the torque generation, but it is recommended that further studies should try to control this parameter, by providing shoes for the participants or by modifying the H-station (e.g., adding a foot pad).

4.3. Practical Applications

Our findings indicate that the H-station is a reliable tool for measuring relative strength changes in quadriceps, hamstring, and H/Q-ratio. Moreover, the H-station is relatively cheap, enabling the device to be placed in public and private clinics to, e.g., monitor patients after an anterior cruciate ligament injury, as well as in professional or semi-professional sport clubs. This has important implications for practitioners, coaches, and clinicians, as the device will enable monitoring of the strength of the lower limb of athletes over a season or after an injury for handball or football players. In that respect, SEM and MDC values are very useful to ensure that gains in quadriceps and hamstring strength are above measurement errors and clinically relevant.

5. Conclusions

The novel approach of using the FysioMeter H-station for measuring isometric quadriceps and hamstring strength in healthy young adults showed an excellent relative reliability, an acceptable absolute reliability, and a moderate to good concurrent validity. For the H/Q ratio measures, the H-station was highly reliable for the relative reliability, acceptable for the absolute reliability, but poor for the concurrent validity.

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Data Availability Statement: The data presented in this study are available on request from the corresponding author (P.M.).

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The Landing Biomechanics in Youth Female Handball Players Does Not Change When Applying a Specific Model of Game and Weekly Training Workload

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Abstract: This study aimed to explore the effects of competitive match play and subsequent training during typical competitive microcycle on landing biomechanics in female youth handball players. A group of 11 elite female youth players (age: 14.3 ± 0.6 years; stature: 165.9 ± 8.1 cm; body mass: 58.4 ± 10.6 kg; maturity offset: 0.4 ± 0.8 years) were tested prior to a competitive match, immediately after the match, 48 h after the match, 96 h after the match, and before the next match. The players performed two analyzed trials of a single leg (preferred) counter movement jump. The "Landing Error Scoring System" (LESS) was used to analyze the participants' landing biomechanics. Results: There was no significant effect of a competitive handball match on LESS ($Z = 0.28$; $p = 0.78$). No statistically significant difference in LESS was found between the first and the last measurement ($Z = 1.01$; $p = 0.31$). No significant main effect of time was found for landing biomechanics in the observed eight-day period ($\chi^2 = 4.02$; $p = 0.40$). The results of the study indicate that a model of weekly loading during in-season, including competitive match play, does not decrease lower limb biomechanics during landing and does not contribute to an increased risk of anterior cruciate ligament injury in female youth handball players during a competitive microcycle.

Keywords: fatigue; landing error scoring system; injury; ACL; risk factor; maturation

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1. Introduction

Handball is a team sport of an intermittent character with frequent physical contacts between players in which players perform a significant number of sudden accelerations and decelerations, changes in speed, unanticipated rapid changes in directions, and jumping and landing tasks [1,2]. Although differences in gender, age category, and performance levels in game load exist [3], all handball players are exposed to a high risk of injury when the hours of athlete exposure are taken into account [3,4]. In youth female players, the reported incidence was 6.8 injuries/h of exposure in a recent systematic review by Gonzales et al. [5]. In the case of anterior cruciate ligament (ACL) injury, which represents one of the most serious and frequent injuries in youth female players [6,7], the risk of injury during handball matches was found to be thirty times higher compared to training, while 90% of ACL injuries were non-contact injuries [8,9]. One of the reasons is that the biomechanics of the abovementioned specific movements may be greatly influenced by both external and internal factors, including the contact nature of the game and its specific requirements, female gender, and muscle fatigue [10–12]. As frequently suggested by the available literature, ACL injuries seem to be more prevalent in the later stages of handball matches, when muscle fatigue is present [8,13,14] and movement biomechanics and neuromuscular functions are altered [15–17]. Impaired activity of the muscles that maintain a centered position in the knee joint during dynamic activities is manifested by changes in landing biomechanics associated with an increase in ACL loading and thus increased susceptibility to ACL injury [18,19].

Especially in young players, it is important to systematically develop the training status, including players' physical qualities during the whole season [20,21]. However, not only in adult sport but also in youth sport, due to the unequal training and/or game workload, the accumulation of muscle fatigue may appear in training microcycles and consequently reduce the efficiency of the training process or match performance and increase the acute risk of injury. Therefore, it is surprising that only a few studies appear to have explored the potential of the accumulated effects of game load and the subsequent training week load on injury risk mechanisms in female youth sport. In team sports, most attention has been paid to youth soccer. One study has suggested that physical stress (determined as a weekly total of minutes in match play and training) and the session rate of perceived exertion are related to injury incidence in youth soccer players and female youth basketball players [22]. It has also been identified that higher accumulated workloads are associated with a greater injury risk [23]. The findings of recent studies on female youth soccer players [24,25] suggest that neuromuscular functioning may still be compromised at least 4 days after match play and may therefore increase the risk of injury. A recent study on male youth soccer players by Lehnert et al. [26] shows that a weekly load including competitive match play may compromise performance and increase injury risk in players.

It has been proved that jump landing is one of the most common movement patterns for ACL injury [27]. It has also been demonstrated that a proper biomechanical strategy may reduce ACL loading during jump landing and decrease the risk of knee injury [28–30], and that lower limb movement kinematics during jump landing is a modifiable risk factor of ACL injury [12,23,31,32]. The knee-extension moment, proximal anterior–tibial shear force, knee valgus–varus moment, and knee internal–external rotation moment are frequently involved in ACL injury [20]. Numerous studies have demonstrated that females land more frequently with increased knee valgus, reduced knee and hip flexion, increased tibial shear and tibial rotation [33,34]. Moreover, it was suggested that increasing neuromuscular fatigue was reflected in the change in the position of the lower limbs during jump landing [26,31,35]. For this reason, different movement patterns during the early landing phase are monitored and evaluated to identify biomechanical deficits and reduce the risk of injury in players [26,36,37]. In this context, a video-based analysis is used to identify potential ACL risk factors related to movement parameters [29,38,39] and the "Landing Error Scoring System" (LESS) is a valid and reliable screening tool which identifies landing-related movement patterns associated with non-contact ACL injuries [12,29,31,40]. In the referred study Padua et al. [26], the authors reported good sensitivity of 86% and acceptable specificity of 64% to identify the risk of non-contact ACL injury in youth female and male soccer players. The authors also found out that the relative risk of sustaining an ACL injury was nearly 11 times higher in players with a LESS score five and more in comparison with players who had scored less than five.

The available findings, albeit limited, suggest that the accumulated effects of competitive match play and subsequent training prior to next competitive match play need to be explored in team sports to determine the risk of injury during training microcycles as well as the readiness to reperform. As far as we are aware, limited number of studies assessed neuromuscular and biomechanical deficits in youth female handball [26,36,41]. However, no study has observed the changes in lower limb movement biomechanics during landing as a modifiable risk factor of ACL injury in youth handball during the training microcycle. This knowledge may improve our understanding of age-specific responses to training load, help determine the periods with an increased risk of injury and consequently help optimize the training and game load during training microcycles. For this reason, the objective of the present paper was to investigate the effects of competitive match play and subsequent training during a typical competitive microcycle on landing biomechanics in female youth handball players.

2. Materials and Methods

2.1. Participants

This study included a group of 11 elite female youth handball players (age: 14.3 ± 0.6 years; stature: 165.9 ± 8.1 cm; body mass: 58.4 ± 10.6 kg; maturity offset: 0.4 ± 0.8 years) from a female elite handball club. All players had 6–7 years of experience with a regular training process. In the current annual training cycle, players participated on average in a total of three training sessions per week (one 90 min fitness session included; total training load, 270 min) and played 44 competitive matches and 20 friendly matches in the competitive season. All players participated in the competitive match and completed all training sessions during the following microcycle. Players completed a health questionnaire, and only players free of musculoskeletal lower-extremity injury in the previous 4 weeks or serious injury in the previous 6 months were included in the research. The study was approved by the Institution's ethics committee and conformed to the Declaration of Helsinki regarding the use of human subjects. All tested players were fully informed about the aim of the study and the testing procedures that would be employed. Before the study, a written informed parental consent and players' verbal assent to the testing procedures were obtained.

2.2. Procedures

The study was of a cross-sectional design. The players were tested in the gym five times over an eight-day period during the second part of the competitive season. The testing was performed prior to a competitive match (the match time was 50 min; player match time was similar as a result of regular rotation of the players controlled by the coach), immediately after match play, and during the following training microcycle, specifically 48 h, 96 h, and 168 h after match play (prior to the next competitive match). Before the 1st testing, players undertook a familiarization session and carried out the test in training sessions a week before the testing to avoid the learning effect of the test. The training content of the observed female players during the competitive microcycle is shown in Table 1.

Table 1. Training content of the observed group of female handball players.

Special Training Indicators	Minutes per Week
Warm-up	45
Anaerobic endurance	30
Strength	25
Speed	20
Coordination	20
Individual attacking actions	20
Individual defensive actions	20
Offensive combinations	20
Defensive combinations	20
Offensive systems	10
Defensive systems	10
Training game	30

2.2.1. Anthropometry

Body mass measures were taken at the beginning of the 1st testing session, using a weight scale, Tanita UM-075 (Tanita, Tokyo, Japan). For the purpose of biological maturity assessment, leg length, tibia length, and standing and sitting height measures were taken at the beginning of each measurement, using a stadiometer, A-226 Anthropometer (Trystom, Olomouc, Czech Republic). Biological maturity was predicted by the calculation of maturity offset, using the sex-appropriate equation [42].

2.2.2. Landing Mechanics

The participants performed three trials of a single leg counter movement jump (one practice and two analyzed trials), with a 1-min rest between the trials. The average value of the two trials was used in a subsequent analysis [43]. The participants were instructed to take two steps forward; immediately jump as high as possible off one leg (preferred), imagining that they were reaching for a ball above their head; and land on two feet. This stop-jump task is considered more game-specific compared with the previously used drop-jump task off a box and is a convenient testing tool particularly for female athletes and increases the ecological validity of testing [44–46]. Furthermore, the practice of jumping from a box in the context of the LESS has encountered criticism, with concerns raised about its sport-specificity, its efficacy in detecting landing biomechanics linked to lower limb injuries, and its utilization in predicting ACL injuries [47–49]. Recordings from two high-definition video cameras, SONY HXR-MC2000 and SONY HXR-NX5E (SONY Corporation, Tokyo, Japan; frequency 25 Hz), were used for the purpose of a two-dimensional biomechanical landing analysis. The cameras were positioned on tripods 3.5 m from the marked landing area in the frontal and sagittal plane. For the evaluation of images, the ImageJ software 1.50i (National Institute of Health, Bethesda, MD, USA) was used. The videos of the participants were scored retrospectively by the same experienced rater [50]. The LESS, a 17-item scale devised by Padua et al. [40], was applied to evaluate the landing mechanics of the participants. The initial set of observable criteria pertains to the alignment of the lower limbs and the trunk during the moment of the first contact with the ground (Items 1–6). The second set of criteria aims to evaluate inaccuracies in foot positioning (Items 7–11). The third set of criteria is dedicated to examining the motion of the lower extremities and trunk from the point of initial ground contact to the instance of either maximum knee flexion angle (Items 12–14) or maximum knee valgus angle (Item 15). The final two "comprehensive" criteria encompass the overall movement in the sagittal plane and the rater's general assessment of landing quality (Items 16 and 17). The LESS is a comprehensive, valid, and reliable procedure which includes a comprehensive assessment of multiplanar biomechanics which could be used to identify individuals at a risk of sustaining a non-contact ACL injury [31,43], where the landing is evaluated by analyzing the records of landing in the sagittal and frontal planes. The scoring is based on the presence or absence of kinematic characteristics, using a standardized checklist where a higher LESS score suggests poor landing technique, while a lower score is indicative of a proficient jump-landing technique [40]. The overview of the LESS specific errors is presented in Table 2 (adopted from Hanzlíková and Hébert-Losier [51], with permission of authors).

Table 2. Landing Error Scoring System's specific errors.

No.	Item	Definition of Error
1.	Knee flexion at IC	Knee flexion < 30°
2.	Hip flexion at IC	Thigh is in line with the trunk (hips not flexed)
3.	Trunk flexion at IC	Trunk is vertical or extended at the hips (i.e., not flexed)
4.	Ankle plantar flexion at IC	Heel-to-toe or flat foot landing at initial contact
5.	Knee valgus at IC	Center of the patella is medial to the midfoot at initial contact
6.	Lateral trunk flexion at IC	Midline of the trunk is flexed to the left or the right side of the body at initial contact
7.	Stance width (wide)	Feet are positioned greater than shoulder-width apart (acromion processes) at initial contact
8.	Stance width (narrow)	Feet are positioned less than shoulder-width apart (acromion processes) at initial contact
9.	Foot position (toe-in)	Foot is externally rotated more than 30° between initial contact and maximum knee flexion

Table 2. *Cont.*

No.	Item	Definition of Error
10.	Foot position (toe-out)	Foot is internally rotated more than 30° between initial contact and maximum knee flexion
11.	Symmetric foot contact at IC	One foot lands before the other foot, or one foot lands heel to toe and the other foot lands toe to heel
12.	Knee flexion displacement	Knee flexes less than 45° between initial contact and maximum knee flexion
13.	Hip flexion at MKF	Thigh does not flex more on the trunk between initial contact and maximum knee flexion
14.	Trunk flexion at MKF	Trunk does not flex more between initial contact and maximum knee flexion
15.	Knee valgus displacement	At the point of maximum medial knee position, the center of the patella is medial to the midfoot
16.	Joint displacement	Joint displacement: soft (0), average (1), stiff (2)
17.	Overall impression	Overall impression: excellent (0), average (1), poor (2)

Abbreviations: IC, initial contact; MKF, maximum knee flexion.

2.2.3. Statistical Analysis

The data analysis was performed in the Statistica program (version 12; StatSoft, Inc., Tulsa, OK, USA). The distribution of raw data sets was analyzed for homogeneity and skewness by means of the Kolmogorov–Smirnov test. Basic descriptive statistics (means and standard deviations and medians) were used to describe the LESS measure. The one-way Friedman ANOVA was used to investigate the effect of time on the LESS score because of ordinal nature of data and normality violation of the variables measured. The Wilcoxon paired test was applied to perform a comparison of the results before and after the competition and a comparison of the results at the beginning and end of the microcycle. Statistical significance was accepted at $p \leq 0.05$ for all statistical tests.

3. Results

The mean (±SD) and median values of the LESS score for the observed female handball players are shown in Table 3. No significant effect of match play on LESS was evident ($Z = 0.28$; $p = 0.78$). No statistically significant difference in LESS was found between the first and the last measurement (i.e., before the next match) ($Z = 1.01$; $p = 0.31$). No significant main effect of time was evidenced for landing biomechanics indicated by means of the LESS score in a competitive microcycle ($\chi^2 = 4.02$; $p = 0.40$).

Table 3. Landing Error Scoring System—mean (±SD) and median values during the observed periods in U14 female handball players ($n = 11$).

Measurement	Mean (±SD)	Median
Prior to 1st match play	6.36 ± 0.42	6.00
Post 1st match play	6.50 ± 1.16	6.00
48 h post 1st match play	5.82 ± 1.15	5.50
96 h post 1st match play	6.72 ± 1.09	6.50
Prior to 2nd match play	5.95 ± 1.04	5.50

4. Discussion

This study was deliberately designed in an applied ecological setting so as to investigate changes in players over a “typical” weekly cycle. The findings from the study indicate that landing biomechanics was not compromised after the competitive match, during the following weekly competitive microcycle, and before the next competitive match. After the match, there was a non-significant increase in the LESS score by 2.2%, and a comparison of the first measurement (before the first competitive match) and the measurement at the end of the competition microcycle (before the second competitive match) showed a non-significant decrease in the LESS score by 6.5%.

4.1. Changes in Post-Game Landing Biomechanics

After the first match, the landing biomechanics based on the LESS test did not indicate potential acute related match effects that would compromise the landing technique [31]. Thus, the coordinated activity of the large muscle groups that help to control knee joint stability during landing after the jump was probably not compromised, and neuromuscular performance was maintained, thus reducing the risk of injury [18].

The finding of the current study after the match play is not consistent with the results of the previous studies involving youth athletes. These studies have demonstrated a deterioration of most of the observed parameters after match play and/or after a specific fatigue protocol both in youth females [24,52] and males [16,53–56] soccer players. The finding of the current study suggesting no significant decrease in landing biomechanics after match play could be explained, in particular, by the equally distributed match workload among eleven players, which lowered match-related fatigue. Other reasons for the difference between the findings of the current study and the abovementioned studies could be the differences in age and gender of the players, the differences between game performance in handball and soccer, and the differences between real match-play workload and a specific fatigue protocol.

4.2. Changes in Landing Biomechanics during the Competitive Microcycle

The data suggest that, also during the following weekly competitive microcycle, players' landing movement pattern did not change. This result indicates that there were no accumulated effects on the LESS score. As landing mechanics was not compromised before the second competitive match and was similar to baseline levels (before the first match), players were probably ready to reperform at the end of the observed competitive microcycle with a typical training load. The results show that the applied model of game and weekly training workload is proportional to the training status of the players and provides enough space for recovery of the players and elimination of residual muscle fatigue. On the contrary, the question is whether three training units per week represent a sufficient workload for 14-year-old handball players on their development pathway.

It is not possible to compare the results of the current study with other findings in female youth handball players, as no available studies have examined changes in the landing biomechanics during a microcycle. Nevertheless, these data do not correspond to the findings of a few studies on youth soccer players which showed that some observed indicators of the risk of injury were compromised during a competitive microcycle. Specifically, a study on female youth soccer players indicated that, 96 h later, the eccentric fatigue task torque returned to pre-fatigue levels, but the electromechanical delay was still significantly compromised post fatigue [24]. Also in soccer, Hughes et al. [25] reported significantly elevated creatine kinase levels in 13–16-year-old female players 80 h post-match play and during the training week. In another study by Lehnert et al. [26] on U14 and U16 male youth soccer players, the LESS and other injury risk indicators were not compromised at the end of the observed microcycle (before the next competitive match) in the U14 category; nevertheless, there were significant changes during the microcycle in most of the indicators. In the U16 age group, players demonstrated reductions in the reactive strength index and increases in the creatine kinase level at the end of the observed microcycle.

Although no changes in the landing biomechanics in youth female players were found in the current study, the LESS score in all of the measurements indicated that the players had poor landing mechanics, which might place them in the high-risk category (value range 5.95 to 6.72; average score 6.27 $\pm$ 1.17) and consequently could point to an increased risk of injury during the match play and training process, not only in the observed period. According to the authors of the LESS scale, the key cutoff value suggested for high injury risk is five points [40]. The performance of the tested person is evaluated as good if below five points, while a score above five points is associated with a higher risk of ACL [40,43]. Padua et al. [31], in their study involving 348 boys and 481 girls (soccer players) whose average age was similar to the players in our study (13.9 years vs. 14.3 years), proved that a LESS score equal to or greater than six was associated with a greater risk of injury as opposed to individuals with a value equal to or lower than four. Although the conditioning program designed for the group of female adolescent handball players included plyometric training, they did not take part in systematic training of landing and/or neuromuscular training program. As preventive training programs have been shown to reduce the risk of lower limb injuries in youth athletes [57,58], we believe that a preventive training program, which would also encompass movement skills, including jumping and landing, would be useful for the observed female handball players to improve their landing biomechanics [59]. This applies not only to female players of the observed age category but also to other categories, including prepubescent players. This could potentially contribute to a decrease in the relatively high rate of lower limb injury in female youth handball players, particularly those around and/or after Peak Height Velocity (PHV).

It should also be noted that the participants in this study were female, which is considered one of the main risk factors for developing ACL injuries [60]. Higher LESS values in females compared to males, indicating an increased risk of ACL injury, were found in several studies in which the average LESS values in females were in the risk zone for ACL injury [31,40,61]. Previous studies have shown differences in lower limb biomechanics between men and women [37], particularly in the position of the hip and knee joint during flight and land. The reduced abductor muscle force in females (especially musculus gluteus medius) may be one of the causes of the occurrence of non-contact ACL injury, as it contributes to insufficient stabilization and excessive adduction and internal rotation in the hip joint. This has an impact not only on the hip position itself but also on the knee joint, where the abduction angle increases. These variables increase the forces on ACL, thereby increasing the risk of injury [62].

Moreover, the age of the subjects in this study was 14.3 ± 0.6 years, and the maturity offset was 0.4 ± 0.8 years, indicating that the subjects were around and/or after PHV. Previous studies [63,64] showed that girls after PHV had the largest increase in tibial abduction moment after landing, which has significant implications for the biomechanics of knee joint loading. Also, Hewett et al. [63] consider the period of PHV or post-PHV a risk factor of ACL rupture due to impairment of knee landing biomechanics. In this regard, we can consider the observed female players as an "at-risk" group.

4.3. Limitations

One of the limitations of the current study was that only 11 female youth players from the U14 competitive age group from a professional handball club were observed. Therefore, the results of this study should only be generalized to similar groups of players. Moreover, it was not possible to determine the weekly load using GPS, as this was not available to the research team. Another limitation is that the study design did not allow testing prior to a subsequent weekly training session. In future studies, it would also be appropriate to monitor the effects of competitive match play and subsequent training during training microcycles not only on landing kinematics but also on other parameters obtained by both objective (e.g., force platform) and subjective (e.g., the Visual Analogue Scale) diagnostic methods.

5. Conclusions

It appears that this study is the first to show that landing biomechanics, as one of the ACL injury risk factors, is not impaired after competitive match play in youth female handball players around and/or after PHV provided that the match time is similar as a result of regular rotation of the players controlled by the coach during the game. Similarly, landing biomechanics does not change during the subsequent competitive microcycle with three team training units (which is typically applied in handball youth female academies in the Czech Republic). This suggests players' readiness to reperform at the end of the competitive microcycle. These results indicate that this model of weekly loading during the in-season, which includes competitive match play, does not increase susceptibility to ACL injury in 14-year-old female players during a competitive microcycle. However, in the observed group of handball players, the LESS score suggests that the players may be considered an "at-risk" group, and, therefore, more attention should be paid to the jump-landing technique of these players during their long-term development.

Although the results of the current study cannot be generalized, considering the high incidence of injuries in female youth sport, handball included, the findings of the current study have implications for weekly training strategies, especially in competitive microcycles. Coaches are recommended to consider the presented training load and match load strategy, which seems to provide enough time for recovery, and also pay attention to the quality of landing mechanics and other handball-specific movements of individual players. Coaches should also pay attention to the technique of fundamental movement skills, including takeoff and landing skills, particularly in preadolescent players, introducing players to, e.g., low-impact hopping and jumping. Moreover, specific intervention programs should be applied to reduce neuromuscular and biomechanical deficits in players and contribute to the optimization of players' developmental pathway.

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Differences in Nurses' Upper-Body Posture in Manual Patient Handling—A Qualitative Case Study

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Abstract: (1) Background: In the context of nursing challenges and workforce shortages, nurses experience significant physical and psychological strain due to manual patient handling. (2) Methods: This study investigates differences in nurses' upper body postures, patient turning acceleration, and perceived exertion during a typical repositioning process within two repositioning maneuvers. (3) Results: The results reveal variations in positioning duration, upper-body posture angles, and turning acceleration between nurses and sequences. Nurse 2 exhibits more extreme postures (e.g., lateral flexion $p < 0.001$) and accelerations (e.g., shoulder $p < 0.001$) but reports lower perceived exertion ($p = 0.03$). (4) Discussion: These findings emphasize the need for ergonomic adherence and targeted training to enhance patient repositioning. Comprehensive solutions are necessary for patient and nurse comfort, particularly in cases of higher patient weights. Against the background of ergonomic body posture, this study highlights the potential of innovative tools and ongoing research to alleviate physical strain and enhance patient care.

Keywords: nursing; bed-bound; care; positioning techniques; movement analysis

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1. Introduction

Nurses globally are faced with substantial challenges and workforce shortages in their daily work activities, with heightened significance due to the COVID-19 pandemic, leading to considerable physical and psychological strain [1–4]. These challenges predominantly stem from the physically demanding nature of manual patient handling tasks, encompassing repositioning, prolonged standing, uncomfortable postures, frequent lifting, carrying, and forceful exertions within a standard 12 h shift [5,6]. This routine contributes to heightened risks of spinal loading and prolonged physical strain, particularly chronic back pain [7–9], rendering nurses particularly susceptible to work-related musculoskeletal disorders (WMSDs) and work incapacities [6,10,11]. Particularly in the intensive care of individuals with special needs and disabilities and bed-bound elderly patients, standardized repositioning protocols executed by nurses are pivotal to preventing pressure sore formation [12,13]. However, the repetitive repositioning process, involving leaning over the bed and gently turning patients, may lead to various musculoskeletal discomforts in nurses [5].

To mitigate WMSDs and reduce the impact of forceful exertions, adhering to ergonomic guidelines is crucial. They recommend that manual lifting, carrying, or pulling should not exceed a maximum weight of 23–25 kg, especially if a high frequency of movement is performed per day [14–17]. Furthermore, forced postures are characterized by the surpassing of 15% of the maximum holding force and a duration exceeding 4 s. A forward-bent posture should not persist for more than 10 s if repeated or for more than 60 s at once [18]. Additionally, ergonomic risk assessment tools such as RULA [19], OWAS [20], and EAWS [21] consider both the intensity and duration of body postures to delineate risk factors and non-ergonomic stances. Guidelines for upper-body posture outlined in ISO and DIN EN standards, employing green, yellow, and red tolerance ranges [16,17], underscore the importance of maintaining optimal upper-body flexion within a range of 0° (neutral posture) to 20° (green tolerance range). Flexion beyond 60° enters the red tolerance range, indicative of non-ergonomic posture. Lateral flexion is advised to be constrained within −10° (left lateral flexion) to 10° (right lateral flexion), with a permissible range of −20° to 20° to mitigate overloading risks. Similarly, back torsion (rotation) should remain within −10° (left rotation) to 10° (right rotation), within a maximum range of −20° to 20°, to minimize strain risks. Adhering to these recommendations can effectively mitigate spinal overloads, provided sustained non-ergonomic postures and the concurrent handling of heavy loads are avoided.

While the literature includes some field studies investigating work-related spinal loads and assessing body postures in manual handling [22–26], the majority are situated within industrial contexts. In the nursing domain, evaluations of nurses' working postures mainly focus on manual patient bed-to-wheelchair transfers [27–31]. Studies addressing manual repositioning maneuvers in bed, particularly those assessing objective kinematic motion analysis for evaluating body postures, are scarce. Consequently, an analysis of the physical and psychological burdens borne by nurses, especially in intensive care settings for individuals with special needs and elderly bed-bound patients and residents, becomes imperative. This analysis is essential for enhancing overall well-being and upholding high standards of care.

Considering these factors, the need to prevent nurses' overload in manual patient handling becomes evident. Longitudinal studies examining the effects of technical training on stress-tolerant work behavior yield conflicting evidence on back pain but demonstrate positive impacts in improving patient and resident repositioning quality [32–35]. A subset of studies exploring technical training's effects on aid usage consistently report reductions in work-related injuries [36,37]. Additionally, training ergonomic supervisors in repositioning processes has been linked to reductions in work-related injuries, neck and shoulder discomfort, and upper and lower back issues [38,39]. Although these studies identify effective technical training types, specific intervention content and their alignment with individual nurses' movement patterns remain unclear, necessitating further research to formulate intervention recommendations.

Motivated by these considerations and prior observations in nursing homes, we identified the prevalent use of bilateral 30° positioning maneuvers for bed-bound elderly patients and residents, which is one of the most common positioning maneuvers, especially in nursing home settings with older adults [40] (cf. Figure 1). This repositioning procedure involves the adjustment of the patient's body position by tilting them at a 30° angle and slowly rolling them onto one side while maintaining the angle. It has been shown that this positioning is most effective in reducing the pressure on different body segments and is therefore mostly recommended [41]. Adequate manual patient repositioning techniques, such as supporting the patient's limbs and body weight, are conducted to ensure secure and controlled repositioning for both sides (left and right positioning maneuvers). It is important to consider that the execution and the duration of the 30° repositioning process may vary based on individual patient needs and healthcare facility protocols. Moreover, due to the conditions of the positioning of the bed of the immobile persons, nurses have to work from both sides (left and right) in their daily routines. This might lead to unfavorable positions for the nurses if they have a preferred side due to their handedness. According to the actual situation in healthcare (e.g., time pressure, skill shortage, etc.; [42]) we wanted to explore this practice further with an explorative laboratory study. The specific research aims were the analysis of nurses' upper-body postures, their perceived exertion, and the bed-bound individual's turning acceleration during the implementation of this maneuver.

Left-sided 30° Right-sided 30°

Figure 1. Thirty-degree positioning.

Therefore, this exploratory case study aims to investigate differences between two nurses regarding upper body flexion, lateral flexion, and rotation angles as well as participants' turning acceleration and nurses' perceived exertion.

Against this background, we derived the following questions:

1. Do the nurses' durations of the repositioning process differ?
2. Do the nurses' body positions differ during the repositioning maneuver?
3. Does the acceleration of the participants lying in bed significantly differ between both nurses and does it change over time?
4. Does the perceived physical exertion significantly differ between nurses?

We assumed differences between the nurses considering their positioning technique and accompanying positioning durations, causing differing upper body positions in flexion, lateral flexion, and rotation. Furthermore, these variances in upper body positions may change over time during the turning maneuver. Additionally, we hypothesized that the nurses' positioning technique is related to the participants' turning acceleration and that it may change over time during the different turning maneuver sequences. Considering the expected differences in the individual nurses' positioning techniques, we assumed variances in physical perceived exertion between both nurses. These results could be used afterward for conducting additional field studies with bigger cohorts of nurses.

2. Materials and Methods

2.1. Trial Design

The experimental design followed a randomized crossover procedure. Accordingly, the positioning techniques (30° inclined position to the right, 30° inclined position to the left) were performed in randomly selected order by two nurses with all patient subjects. The laboratory experiment was performed on the premises of the University of Hamburg—Department of Movement and Exercise Science. The investigations took place between 6 June and 2 July 2020.

2.2. Participants

The subject sample included two trained professional female nurses, both aged 29 years, as well as 15 participants serving as patients (further named patient participants), covering the range of 5th to 95th height percentiles for representative results. All patient participants recruited through the University of Hamburg were healthy adults. For analysis, we matched 12 patient participants for left-sided positioning, and seven patient participants for right-sided positioning for each nurse. Tables 1 and 2 show the nurses' anthropometric data, as well as the patient subjects' anthropometric data for each positioning side.

Table 1. Nurses' age and anthropometric data (N = 2) *†.

Nurse (n)	Age (y)	B_M (kg)	B_H (cm)
Total	29.0 ± 0.0	69.0 ± 2.8	174.5 ± 5.0
1 (female)	29	67	170
2 (female)	29	71	177

* BM = body mass; BH = body height. † Values are mean ± SD.

Table 2. Patient subjects' anthropometric data (N = 15) *†.

Positioning Side	N	Age (y)	B_M (kg)	B_H (cm)
Right	7	35.6 ± 15.0	72.7 ± 11.4	172.6 ± 7.7
	2 males	29.0 ± 2.0	80.5 ± 0.5	183.0 ± 4.0
	5 females	38.2 ± 17.0	69.6 ± 12.2	168.4 ± 3.9
Left	12	31.8 ± 16.5	73.1 ± 15.1	173.2 ± 9.6
	6 males	32.8 ± 16.0	83.3 ± 11.9	181.5 ± 4.5
	6 females	30.8 ± 16.9	62.8 ± 10.2	164.8 ± 5.1

* BM = body mass; BH = body height. † Values are mean ± SD.

2.3. Measurements and Test Instruments

The following standardized instruments were used.

2.4. Three-Dimensional Motion Analysis

The 3D motion analysis was performed via Xsens MVN 2018, a three-dimensional kinematic motion measurement system using a constant frame rate of 60 Hz. The nurses' body movements were analyzed using 17 inertial sensors on various body segments such as the head, sternum, both sides of the shoulder, upper arm, forearm, hand, thigh, lower leg, feet, and pelvis. This setup allowed a three-dimensional analysis of the body segments in a Cartesian coordinate system [43]. The system showed strong correlations compared to the gold standard "vicon motion analysis" (Vicon Motion Systems, INC, Vicon, Oxford, UK) for measuring kinematics [44]. Based on its biomechanical model, the trunk flexion (forward lean angle of the trunk including pelvis tilt with respect to the vertical axis), the lateral flexion of the trunk (side lean angle of the trunk with respect to the pelvis), and the trunk's rotation (offset of the shoulder girdle with respect to the position of the pelvis) were used for upper body kinematics. The data from inertial sensors on their hands were used to analyze the acceleration of the repositioning process. This attachment allowed the acceleration of the repositioning to be recorded at the points of application (shoulder and hip) of the bed-bound patient participant (Figure 2) in X (transverse), Y (sagittal), and Z (longitudinal) directions. In the left-sided 30° inclined positioning, the right hand mainly guided the shoulder, while the left hand guided the hip area. In the right-sided 30° inclined position, the right hand guided the hip area and the left hand guided the shoulder area.

Figure 2. Xsens sensor coordinates placed on nurses' hands.

2.5. Subjective Perceived Exertion—Borg Scale

A 15-point Borg scale (6–20) was used to subjectively assess the general and specific (neck, shoulders, arms, upper back, lower back, legs) exertion. Here, 6 points correspond to very, very light exertion and 20 points to very, very heavy exertion [45,46].

2.6. Study Procedure

Each nurse performed the manual 30° positioning maneuver three times to the left and the right side with each patient's participant. The positioning technique followed the nursing standards for geriatric care and thus represented common repositioning from everyday professional life in the nursing sector [47]. Prior to performing the positioning process, the individual bed height was set to ensure an ergonomic posture, following the ISO and DIN EN guidelines [16,17]. They started from the neutral supine lying position. A fixed randomization schedule determined the order of the sides. Before each measurement session, nurses received instructions on the sequence and extent of positioning techniques. Each positioning technique was performed 3 times, and the mean value was calculated. After each positioning, the nurses assessed their general and specific perceived exertion via the Borg scale. Participants were instructed to mimic immobile patients, meaning that they were not allowed to actively move during the positioning maneuver.

2.7. Statistical Analysis

The recordings were analyzed and divided into four sequences by defining six time points during the respective 30° inclined positioning (cf. Figure 3). These were (1) the start of the overall positioning process (initial position—supine position), (2) the start of sequence 1 in the positioning process, (3) the end of sequence 1 in the positioning process, (4) the start of sequence 2 in the positioning process, (5) the end of sequence 2 in the positioning process, and (6) the end of the overall positioning process (final position—oblique position). For data analysis, we divided the whole turning maneuver into two sequences. The first sequence included the start of the positioning process until the participant was moved into a lateral posture and the nurse placed a blanket as a back support structure behind the participant. The second sequence included the positioning of the participant at the aiding blanket at their back until the end positioning. Data analysis included the calculation of maximum acceleration values in a Cartesian coordinate system (X, Y, Z; cf. Figure 2) and maximum upper-body postures in flexion/extension, lateral flexion, and rotation using MATLAB (MATLAB R2019b, Natick, MA, USA). Furthermore, data were cropped at the start and end points and smoothed with a moving average. An interpolation for 101 data points resulted in a time normalization (100%) for each sequence.

Statistical analysis included a two-way analysis of variance to compare between nurses and sequences. The alpha level was set at 0.05 for all analyses with Bonferroni correction for multiple testing. Effect sizes were calculated by partial eta square (ηp^2). SPSS version 25.0 was used for all statistical analyses.

To detect significant differences between both nurses' continuous trunk motions, as well as for accelerations, a two-step statistical parametric mapping (SPM) was used. Firstly, Hotelling's T2 (SPM{T2}) test was used to identify if statistically significant differences between both nurses' motions occurred, combining flexion, lateral flexion, and rotation of the trunk as a three-component vector, with changing values over time. If significant differences were found, secondary post hoc analysis using a paired-t-test (SPM{t}) was conducted for all three trunk movements separately to reveal differences for each motion of the trunk (flexion, lateral flexion, and rotation). The null hypothesis was rejected if the computed T2 (or t-value, respectively) for the trajectories exceeded the critical threshold values. All SPM analyses were conducted using the open-source code provided by Pataky et al. (version 0.4.7) (www.spm1d.org, accessed on 1 January 2021) in Matlab 2019a.

Missing data values occurred due to technical errors during data collection.

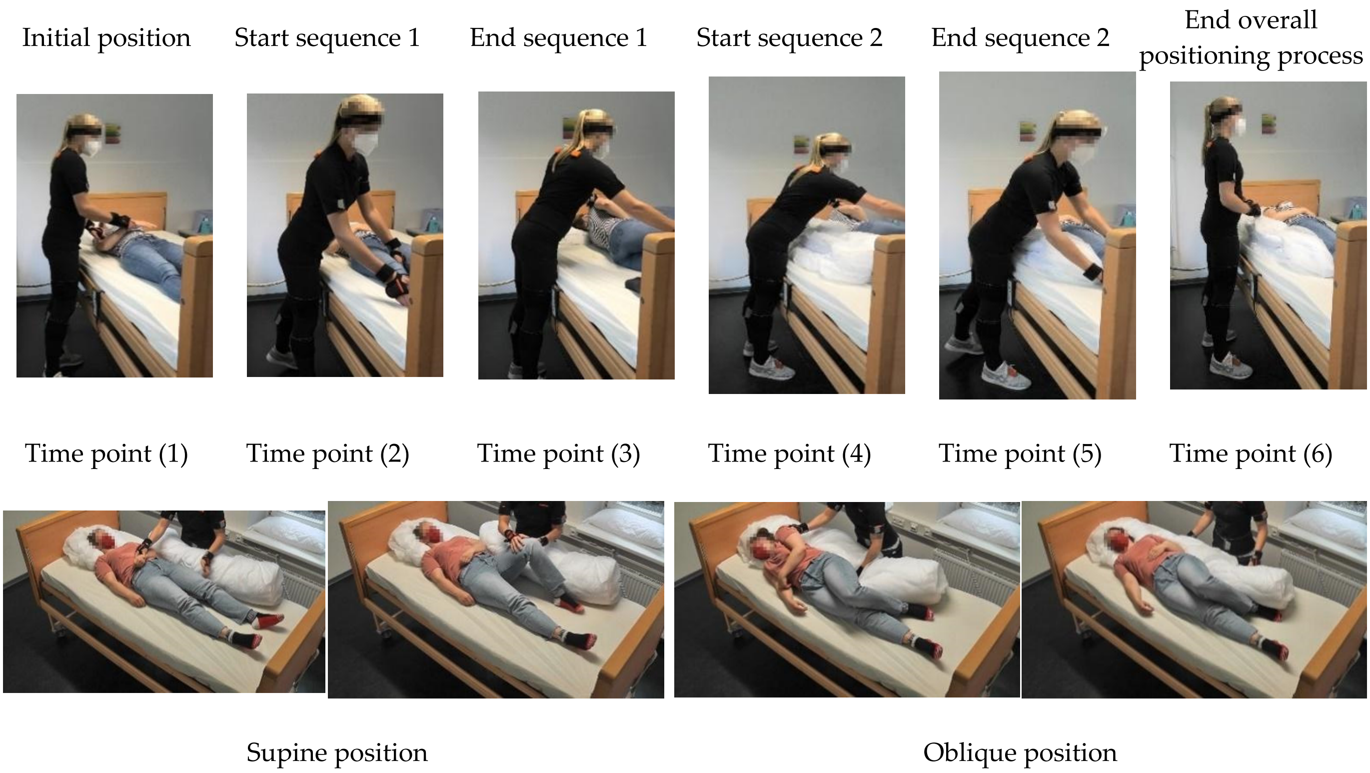


Figure 3. Nurses' upper body posture during the turning maneuver.

3. Results

3.1. Duration of the Turning Process

Table 3 shows the duration time for the left- and right-sided positioning for both nurses, as well as sequence 1 and sequence 2 (cf. Table 3). We found a significant main effect for left-sided and right-sided positioning between nurses and sequences ($p < 0.001$) after correction for multiple testing, with overall shorter durations in sequence 1, as well as shorter overall durations in nurse 1. Statistical analyses also showed significant results in between-subject effects for nurses (left: $p < 0.001$, $\eta p^2 = 0.258$; right: $p = 0.0045$, $\eta p^2 = 0.157$), as well as for sequences (left: $p < 0.001$, $\eta p^2 = 0.836$; right: $p < 0.001$, $\eta p^2 = 0.865$). There was no significant interaction effect between nurse and sequence.

Table 3. Time comparison between nurses and sequences for left- and right-sided positioning; two-way analysis of variance, main effects (ME), interaction of nurse x sequence (IA), and between-subject effect of nurses (BE). (N = 19) *†.

Positioning Side	Seq	Nurse 1 (s)	Nurse 2 (s)	ME		IA	BE(n)	BE (seq)
				p	η_p^2	p	η_p^2	η_p^2
Left (N = 12)	1	1.88 ± 0.18 (1.64–2.34)	2.24 ± 0.39 (1.52–3.07)	**<0.001**	**0.845**	0.263	**0.258**	**0.836**
	2	3.67 ± 0.52 (2.32–4.30)	4.33 ± 0.53 (3.72–5.62)					
Right (N = 7)	1	1.83 ± 0.29 (1.29–2.33)	2.10 ± 0.23 (1.88–2.59)	**<0.001**	**0.868**	0.672	**0.157**	**0.865**
	2							

* BM = body mass; BH = body height. † Values are mean ± SD.

3.2. Upper-Body Posture

Figure 3 shows the exemplary sequences and nurses' upper body posture, as well as the positioned participants. Furthermore, Table 4 shows the maximum upper-body angles during the turning maneuver in the analysis of the time course.

Table 4. Comparison in upper-body positions in left- and right-sided positioning between nurses and sequences; two-way ANOVA, main effects (ME), interaction of nurse x sequence (IA), and between-subject effect (BE) *†.

Upper-Body Position	Seq	Nurse 1 (°)	Nurse 2 (°)	ME		IA	BE (n)	BE (Seq)
				p	η_p^2	p	η_p^2	η_p^2
Left-sided (N = 12)								
max flexion	1	30.1 ± 3.1 (24.7–35.7)	32.9 ± 3.9 (26.2–38.4)	**0.001**	**0.301**	0.167	**0.240**	**0.065**
	2	30.6 ± 4.5 (21.6–36.7)	36.7 ± 4.3 (29.1–45.8)					
max lateral flexion	1	−1.1 ± 1.3 (−3.2–1.4)	3.3 ± 10.5 (−7.5–32.1)	**<0.001**	**0.552**	0.354	**0.178**	**0.499**
	2	10.4 ± 5.2 (4.5–22.9)	18.5 ± 6.3 (6.0–28.4)					
max rotation	1	8.3 ± 2.9 (4.9–14.3)	15.5 ± 7.6 (−3.1–25.7)	**0.001**	**0.323**	0.329	**0.223**	**0.145**
	2	14.2 ± 4.3 (7.9–20.7)	18.3 ± 5.3 (8.1–25.3)					
Right-Sided (N = 7)								
max flexion	1	33.4 ± 3.0 (30.6–40.0)	34.6 ± 3.1 (30.9–40.8)	0.845	0.033	0.863	0.008	0.024
	2	32.3 ± 7.8 (17.7–41.2)	32.8 ± 3.4 (27.9–39.0)					
max lateral flexion	1	3.5 ± 6.2 (−9.8–12.4)	19.5 ± 7.0 (7.3–28.6)	**<0.001**	**0.616**	**<0.001**	**0.210**	**0.397**
	2	3.7 ± 3.9 (−3.8–7.6)	0.0 ± 6.4 (−8.2–11.0)					
max rotation	1	4.1 ± 6.6 (−0.7–19.9)	3.6 ± 4.0 (−4.3–8.1)	**<0.001**	**0.828**	0.177	**0.056**	**0.824**
	2	25.5 ± 6.7 (15.3–34.4)	32.5 ± 5.1 (25.8–39.6)					

* seq = sequence; n = nurse; ° = degree; † = values are mean ± SD (range). Results in bold type indicate significance at the $\alpha \leq 0.05$ level.

3.2.1. Maximum Upper-Body Angles

The maximum upper body angles measured in left-sided positioning achieved by the two nurses with N = 12 participants and in right-sided positioning performed with N = 7 participants are shown in Table 4 as well as Figure 3.

Upper-Body Flexion

For left-sided positioning, we found more pronounced flexion angles within both nurses in sequence 2, as well as overall less pronounced flexion angles in nurse 1 compared to nurse 2. The between-subject effect of the nurses was significant (p = 0.001, ηp^2 = 0.240). Both nurses showed an ergonomically acceptable upper body flexion, as long as this was not maintained over a longer period of time according to guidelines ISO 11226 [17] and the DIN EN 1005-4 [18]. For right-sided positioning, we found comparable angles for both nurses but more pronounced flexion angles within sequence 1 (cf. Table 4).

Upper-Body Lateral Flexion

For left-sided lateral flexion, we found more pronounced positions to the right side within both nurses in sequence 2. Nurse 1 showed an ergonomically optimal upper-body position in lateral flexion in sequence 1 but a partly not-quite-optimal posture in sequence 2. Nurse 2 showed predominantly an acceptable ergonomic upper body posture but partially extreme postures in upper-body lateral flexion according to the guidelines. The statistical analysis showed a significant between-subject effect for nurses (p = 0.004, ηp^2 = 0.178), as well as for sequences (p < 0.001, ηp^2 = 0.499). For right-sided lateral flexion, we found smaller angles in nurse 1 and more pronounced angles in nurse 2 within sequence 2. Nurse 1 mainly stayed in an optimal ergonomic upper-body position in lateral flexion within both sequences, whereas nurse 2 only showed an ergonomically optimal upper-body posture in sequence 1. In sequence 2, nurse 2 showed lateral flexion angles within a tolerable range, but partially poor posture, according to the guidelines. The statistical analysis showed a significant interaction effect between nurse and sequence ($p < 0.001$, $\eta p^2 = 0.405$) and a significant between-subject effect for nurses ($p = 0.019$, $\eta p^2 = 0.210$), as well as for sequences ($p = 0.001$, $\eta p^2 = 0.397$).

Upper-Body Rotation

For left-sided positioning, we found more pronounced upper-body rotations within sequence 2 but overall smaller rotation angles in nurse 1. Nurse 1 mainly stayed in an ergonomically optimal upper-body posture in rotation within sequence 1, but in a posture that was no longer ergonomically tolerable in sequence 2. Nurse 2 mainly stayed in a just barely acceptable tolerance range within sequence 1 but partially showed a poor upper-body posture in rotation within sequence 2, according to the guidelines. The statistical analysis revealed significant between-subject effects for nurses ($p = 0.001$, $\eta p^2 = 0.223$), as well as for sequences ($p = 0.009$, $\eta p^2 = 0.145$). For right-sided positioning, we found more pronounced upper-body rotations within sequence 2 with poor upper-body postures for both. Within sequence 1, both nurses mainly showed an ergonomically optimal upper-body posture in rotation according to the guidelines. The statistical analysis revealed a significant interaction effect for nurse and sequence ($p = 0.177$, $\eta p^2 = 0.075$) and a significant between-subject effect of nurses ($p = 0.246$, $\eta p^2 = 0.056$), as well as for sequences ($p < 0.001$, $\eta p^2 = 0.824$).

3.2.2. SPM Analysis for Upper-Body Kinematics

SPM analysis (Figure 4) showed upper-body position changes over time during the turning maneuvers. Hotelling's T2 test revealed a significant difference for sequence 1 between 0% and 28% of the cycle as well as for 51% and 100% of the cycle. For the second sequence, significant differences occurred between 0% and 28%.

A Shapiro–Wilk test revealed a non-normal distribution for all angles. Therefore, for the post hoc tests, a non-parametric equivalent SnPM{t} to the SPM{t} analysis was conducted. To account for multiple tests, a Bonferroni correction set the alpha value at 0.167. We detected significant differences between the movement of both nurses in specific parts of the movement cycles for the flexion in sequence 1 (51–100%), the flexion in sequence 2 (0–31%), the lateral flexion in sequence 1 (0–30%), and the rotation in sequence 1 (55–87%).

Flexion Sequence 1 | Flexion sequence 2

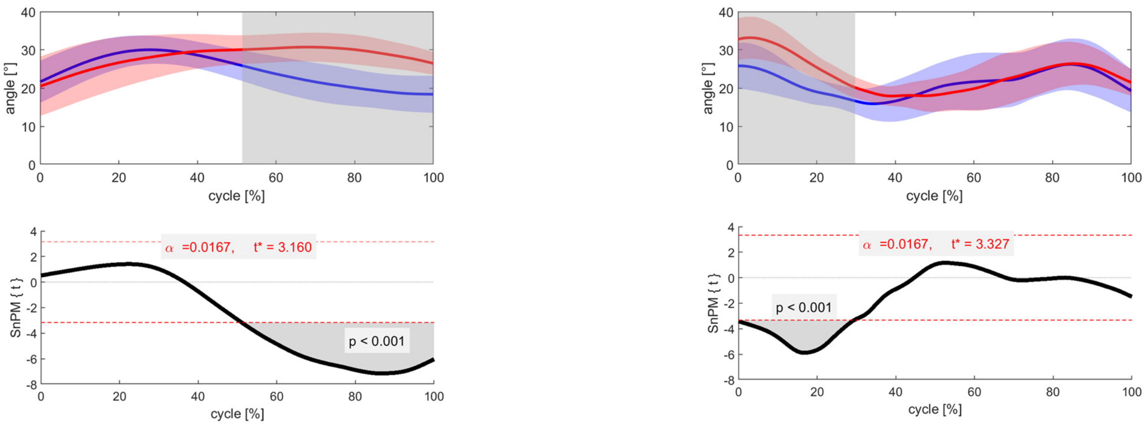


Figure 4. *Cont.*

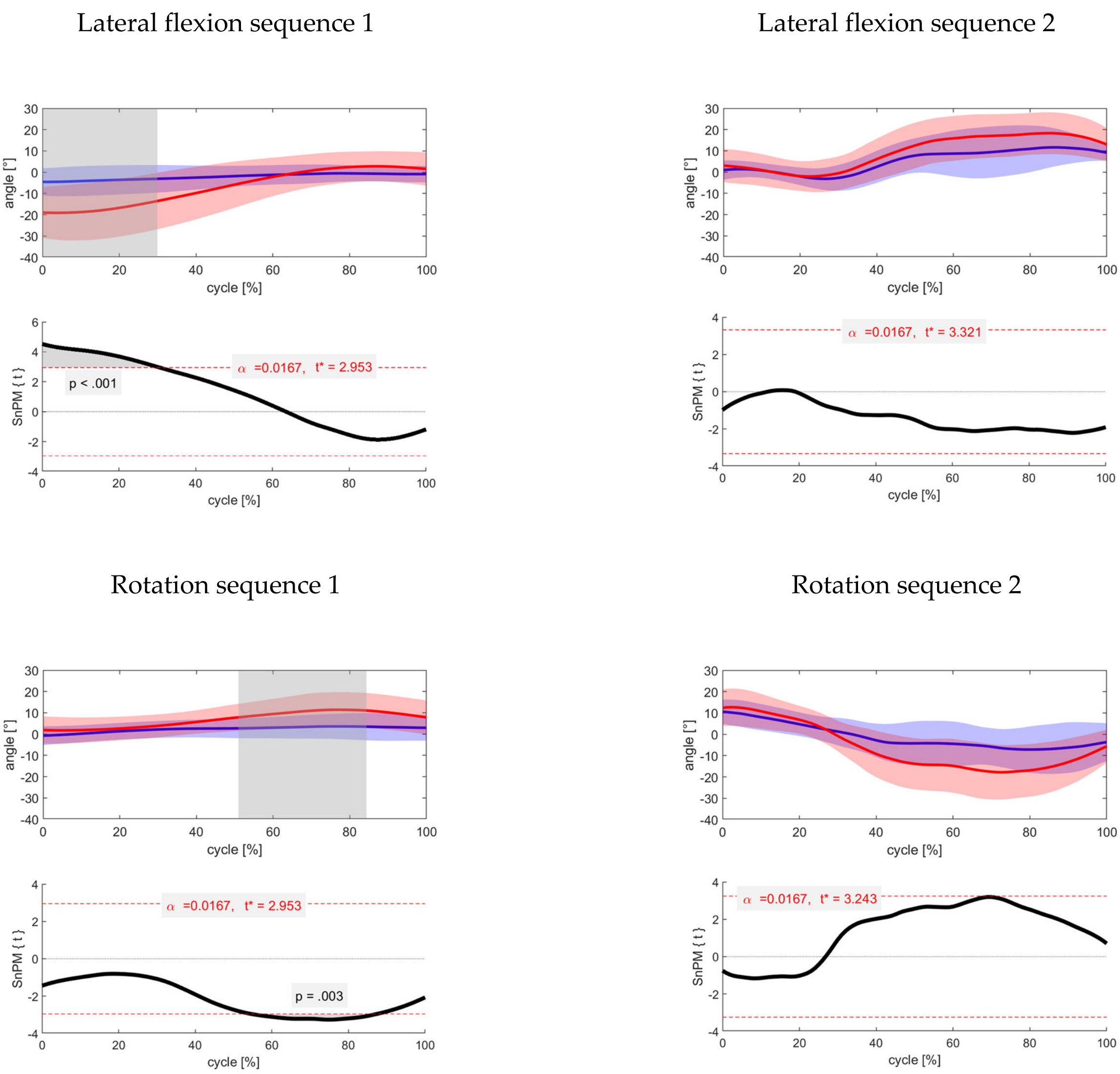


Figure 4. SPM analysis of body postures. Blue line = nurse 1; red line = nurse 2; non-parametric t-test.

3.3. Turning Acceleration

The following results of the turning acceleration are divided into the analysis of the maximum data and the analysis of the time course.

3.3.1. Maximum Turning Acceleration

Tables 5 and 6 show the results of the left- and right-sided turning maneuver with the maximum acceleration in X (transverse), Y (sagittal), and Z (longitudinal) directions (cf. Figure 2) for participants' shoulder and hip, separated for both sequences as well as for both nurses.

Table 5. Comparison of maximum turning acceleration in left-sided positioning between nurses and sequences; two-way ANOVA, main effects (ME), interaction of nurse x sequence (IA), and between-subject effect (BE) (N = 12) *†.

X, Y, Z	Seq	Nurse 1 (m/s^2)	Nurse 2 (m/s^2)	ME		IA	BE (n)	BE (Seq)
				p	η_p^2	p	η_p^2	η_p^2
Hand on shoulder X	1	1.17 ± 0.35 (0.75–1.92)	1.91 ± 1.79 (0.68–7.19)	**<0.001**	**0.710**	0.666	0.065	0.704
	2	4.80 ± 0.75 (3.33–5.75)	5.25 ± 1.31 (3.05–6.86)					

Table 5. *Cont.*

X, Y, Z	Seq	Nurse 1 (m/s^2)	Nurse 2 (m/s^2)	ME		IA	BE (n)	BE (Seq)
Hand on shoulder Y	1	1.19 ± 0.50 (0.66–2.50)	1.83 ± 0.86 (0.72–3.38)	**<0.001**	**0.651**	0.143	0.174	0.616
	2	4.04 ± 1.41 (2.50–7.82)	5.91 ± 2.28 (2.87–9.09)					
Hand on shoulder Z	1	1.19 ± 0.50 (0.94–2.67)	2.18 ± 0.54 (1.42–3.41)	**<0.001**	**0.661**	0.064	0.216	0.615
	2	3.08 ± 0.85 (1.98–4.38)	4.07 ± 0.60 (3.43–5.16)					
Hand on hip X	1	5.08 ± 1.64 (2.83–7.36)	3.21 ± 1.66 (0.99–5.90)	**0015**	**0.210**	0.118	0.166	0.008
	2	4.67 ± 0.94 (3.02–6.43)	4.10 ± 1.33 (2.19–6.01)					
Hand on hip Y	1	6.23 ± 2.07 (2.14–9.73)	4.82 ± 3.39 (1.18–12.36)	**0.016**	**0.208**	0.358	0.038	0.169
	2	3.77 ± 1.39 (2.35–6.24)	3.53 ± 1.14 (1.53–5.55)					
Hand on hip Z	1	5.12 ± 2.00 (2.68–8.53)	2.50 ± 1.35 (0.93–5.91)	**<0.001**	**0.354**	**<0.001**	0.044	0.031
	2	3.64 ± 0.93 (2.19–5.41)	5.02 ± 1.60 (3.23–8.87)					

* seq = sequence; n = nurse; m/s^2 = meter per square second; † = values are mean ± SD (range); results in bold type indicate significance at the $\alpha \leq 0.05$ level.

Table 6. Comparison of maximum turning acceleration in right-sided positioning between nurses and sequences; two-way ANOVA, main effects (ME), interaction of nurse x sequence (IA), and between-subject effect (BE) (N = 7) *†.

X, Y, Z	Seq	Nurse 1 (m/s^2)	Nurse 2 (m/s^2)	ME		IA	BE (n)	BE (Seq)
				p	ηp^2	p	ηp^2	ηp^2
Hand on shoulder X	1	1.00 ± 0.40 (0.60–1.68)	1.19 ± 0.52 (0.55–2.07)	**<0.001**	**0.828**	0.982	0.012	0.828
	2	5.10 ± 1.58 (2.97–7.56)	5.32 ± 1.08 (3.27–6.66)					
Hand on shoulder Y	1	0.94 ± 0.18 (0.68–1.23)	1.55 ± 0.58 (0.74–2.66)	**<0.001**	**0.718**	0.463	0.136	0.703
	2	4.41 ± 2.48 (1.71–7.79)	5.80 ± 0.91 (5.12–7.69)					
Hand on shoulder Z	1	1.51 ± 0.49 (0.80–2.13)	1.80 ± 0.74 (1.01–3.28)	**<0.001**	**0.755**	0.602	0.127	0.745
	2	3.23 ± 0.53 (2.27–3.90)	3.75 ± 0.53 (3.10–4.47)					
Hand on hip X	1	1.00 ± 0.40 (0.60–1.68)	1.19 ± 0.52 (0.55–2.07)	**0.038**	**0.291**	0.088	0.199	0.030
	2	5.10 ± 1.58 (2.97–7.56)	5.32 ± 1.08 (3.27–6.66)					
Hand on hip Y	1	0.94 ± 0.18 (0.68–1.23)	1.55 ± 0.58 (0.74–2.66)	0.085	0.237	0.766	0.168	0.095
	2	4.41 ± 2.48 (1.71–7.79)	5.80 ± 0.91 (5.12–7.69)					
Hand on hip Z	1	1.51 ± 0.49 (0.80–2.13)	1.80 ± 0.74 (1.01–3.28)	**0.018**	**0.339**	**0.013**	0.171	0.002
	2	3.23 ± 0.53 (2.27–3.90)	3.75 ± 0.53 (3.10–4.47)					

* seq = sequence; n = nurse; m/s^2 = meter per square second; † = values are mean ± SD (range). Results in bold type indicate significance at the $\alpha \leq 0.05$ level.

Participants' Shoulder Acceleration

For left-sided positioning, we found higher shoulder accelerations for turning participants' shoulders in all three directions in nurse 2 compared to nurse 1. The statistical analysis showed a significant between-subject effect of nurses in participants' shoulder sagittal plane (p = 0.004, ηp^2 = 0.174) and in the shoulder longitudinal plane (p = 0.001, ηp^2 = 0.216). Both nurses showed higher overall shoulder accelerations for participants' shoulders in all three directions in sequence 2. The statistical analysis showed significant between-subject effects for sequences in participants' shoulder transverse plane (p < 0.001, ηp^2 = 0.704), sagittal plane (p < 0.001, ηp^2 = 0.616), and longitudinal plane (p < 0.001, ηp^2 = 0.615).

For right-sided positioning, we found higher shoulder accelerations for turning participants' shoulders in all three directions in nurse 2 compared to nurse 1. Both nurses showed higher overall shoulder accelerations for participants' shoulders in all three directions in sequence 2. The statistical analysis showed significant between-subject effects for sequences in participants' shoulders in the transverse plane ($p < 0.001$, $\eta p^2 = 0.828$), sagittal plane ($p < 0.001$, $\eta p^2 = 0.703$), and longitudinal plane ($p < 0.001$, $\eta p^2 = 0.745$).

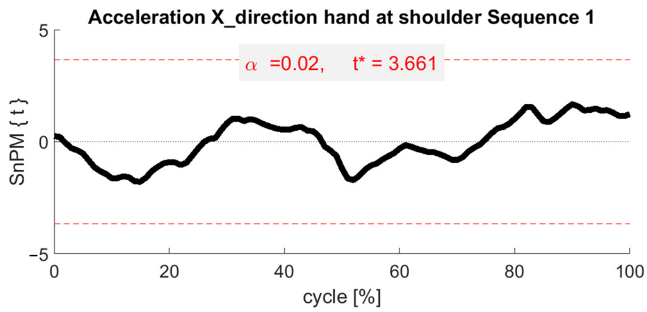
Participants' Hip Acceleration

For left-sided positioning, we found higher hip accelerations in the transverse and sagittal plane for nurse 1 in both sequences compared to nurse 2. In the longitudinal plane, nurse 1 produced higher participant hip accelerations in sequence 1, and nurse 2 produced higher participant hip accelerations in sequence 2. Statistical analysis showed significant between-subject effects for nurses in the transverse plane ($p = 0.005$, $\eta p^2 = 0.166$) and for sequences in the sagittal plane ($p = 0.004$, $\eta p^2 = 0.169$).

For right-sided positioning, we found higher participant hip accelerations in all three directions in nurse 2 compared to nurse 1. Statistical analysis showed significant between-subject effects for nurses in the transverse plane ($p = 0.022$, $\eta p^2 = 0.199$), sagittal plane ($p = 0.038$, $\eta p^2 = 0.168$), and longitudinal plane ($p = 0.036$, $\eta p^2 = 0.171$). Furthermore, both nurses produced higher participant hip accelerations in sequence 2 compared to sequence 1 but without any statistically significant between-subject effects.

3.3.2. SPM Analysis for Turning Acceleration

To check for differences in acceleration in all three directions of the nurses' hands at participants' hips and shoulders over time during the sequences between nurses, a secondary analysis (SPM analysis) was conducted. The SPM–Hotelling's T2 test revealed a significant difference for the nurses' hands at participants' shoulders in sequence 1 between 7% and 25% as well as between 26% and 27% of the cycle (Figure 5). No other differences occurred for nurses' hands at participants' shoulders nor at participants' hips in both sequences.



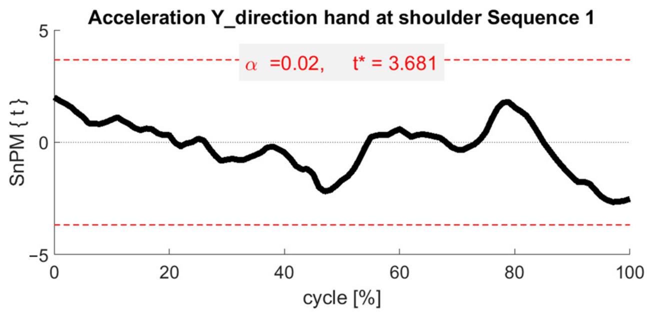


Figure 5. *Cont.*

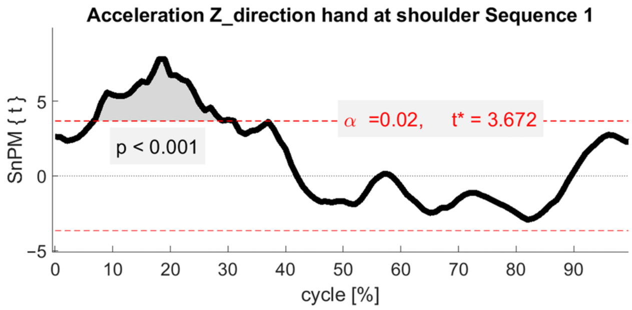


Figure 5. Acceleration of the hand on the shoulder in sequence 1.

Similar to the analysis of the angles, a Shapiro–Wilk test revealed a non-normal distribution for the accelerations, therefore a non-parametric SnPM {t} post-hoc analysis with a Bonferroni correction for nurses' hands at participants' shoulders in sequence 1 was conducted. The analysis showed a significant difference in the acceleration in the longitudinal plane between 7% and 27% of sequence 1.

3.4. Perceived Exertion

Table 7 shows the results of the left-sided positioning for the perceived exertion, separated for both nurses. Overall, nurse 2 showed lower mean values in perceived exertion. Statistical analysis revealed significant differences between both nurses for left-sided overall perceived exertion ($p = 0.033$) with lower mean values of 1.25 (95% CI: 0–2) for nurse 2, perceived exertion of the shoulders ($p = 0.006$) with lower mean values of 1.53 (95% CI: 1–2) for nurse 2, perceived exertion of the arms ($p = 0.014$) with lower mean values of 1.33 (95% CI: 0–2) for nurse 2, as well as perceived exertion of the legs ($p < 0.001$) with lower mean values of 4.00 (95% CI: 3–5) for nurse 2. For right-sided positioning, both nurses showed comparable values, except for perceived exertion of the legs. Nurse 2 showed significantly lower mean values of 3.62 ($p < 0.001$, 95% CI: 2–5).

Table 7. Comparison of perceived exertion in left- and right-sided positioning between nurses; Mann–Whitney U test, main effects (ME) *†.

Borg Score	Nurse 1	Nurse 2	ME				
			U	*p*	*MD*	*SE*	*95% CI*
Left-sided (N = 12)							
overall	10 ± 1 (8–12)	9 ± 1 (7–10)	**35.00**	**0.033**	**1.25**	**0.50**	**0–2**
neck	8 ± 1 (7–10)	8 ± 1 (7–9)	66.00	0.755	0.22	0.35	−1–1
shoulder	9 ± 1 (7–11)	8 ± 1 (6–9)	**25.00**	**0.006**	**1.53**	**0.45**	**1–2**
arms	11 ± 1 (8–12)	9 ± 1 (7–11)	**30.50**	**0.014**	**1.33**	**0.52**	**0–2**
upper back	9 ± 1 (7–11)	9 ± 1 (7–11)	63.00	0.630	0.28	0.52	−1–1
lower back	11 ± 1 (8–12)	10 ± 1 (8–12)	68.00	0.843	0.14	0.54	−1–1
legs	10 ± 1 (8–12)	6 ± 0 (6–6)	**0.00**	**< 0.001**	**4.00**	**0.37**	**3–5**
Right-sided (N = 7)							
overall	9 ± 1 (7–11)	9 ± 2 (7–10)	21.50	0.710	0.29	0.64	−1–2
neck	8 ± 1 (6–9)	8 ± 0 (7–9)	23.50	0.902	−0.19	0.44	−1–1
shoulder	9 ± 1 (7–10)	8 ± 1 (7–10)	15.50	0.259	0.62	0.57	−1–2
arms	10 ± 2 (7–12)	10 ± 1 (8–11)	18.00	0.456	0.48	0.69	−1–2
upper back	9 ± 2 (6–11)	9 ± 1 (8–11)	21.50	0.710	−0.43	0.67	−2–1
lower back	10 ± 1 (8–12)	10 ± 1 (10–11)	22.00	0.805	−0.29	0.57	−2–1
legs	10 ± 2 (7–11)	6 ± 0 (6–6)	**0.00**	**0.001**	**3.62**	**0.55**	**2–5**

* MD = mean difference; SE = standard error; 95% CI = 95% confidence interval. † = Values are mean ± SD (range). Results in bold type indicate significance at the $\alpha \leq 0.05$ level.

4. Discussion and Conclusions

This case study aimed to assess the impact of nurses' upper body posture during manual patient handling techniques, particularly while executing a standardized 30° repositioning maneuver. The study sought to uncover potential variations in posture between the two participating nurses while conducting the positioning maneuvers and correlate these findings with the acceleration experienced by the repositioned patient participants. In the context of patient participant repositioning, differences in the duration of positioning emerged, notably during left-sided repositioning. This variability indicated nuanced differences in the techniques employed by the nurses, despite both being instructed to adhere to ergonomic principles aimed at minimizing physical strain [37,39]. Typically, patient repositioning techniques are tailored to individual patient needs, often within the constraints of nurses' time availability [12]. It was noteworthy that nurse 1 exhibited shorter positioning times compared to nurse 2, potentially leading to reduced stress on body regions due to shorter exposure to challenging postures [5]. However, the study highlighted the fine balance required to avoid rapid movements during repositioning, which can contribute to musculoskeletal discomfort [48].

The subsequent analysis of upper body posture revealed that both nurses' upper-body flexion, lateral flexion, and rotation fell within ergonomic limits if these postures were sustained for short durations. Prolonged maintenance of extreme postures, as highlighted by ergonomic standards [16,17], can result in musculoskeletal overload. It became evident that nurse 2 adopted more pronounced and potentially discomforting postures in all dimensions, implying an increased risk of overloading. Variations in maximum flexion, lateral flexion, and rotation between the two nurses underscored the divergent techniques they employed. The examination of sequence effects and acceleration further illuminated the intricacies of the repositioning process. Sequence 2 emerged as more physically demanding, attributed to the extended duration of sustained extreme upper body postures. Acceleration analysis demonstrated that nurse 2 initiated higher accelerations in both shoulder and hip regions, aligning with her more pronounced upper body postures. This finding was consistent with nurse 1 showcasing a more economical and less physically demanding technique.

Interestingly, nurse 2, despite demonstrating more extreme angular positions, reported lower subjective perceived exertion across various body regions. This seemingly contradictory finding could be attributed to differences in perceived exertion between novice and experienced individuals although both participants were trained nurses, but with different years of experience, as suggested by existing literature [49].

Addressing these challenges in patient repositioning, the study highlighted the limitations of existing assistive aids, such as pillows and blankets, which offer only partial relief to nurses' discomfort and may not consistently improve patient comfort. Specialized systems catering to patient support are often one-sided in their functionality, underscoring the need for comprehensive, innovative solutions that prioritize ergonomic care for both nurses and patients [50,51].

Considering these findings, recommendations for future studies include a more extensive investigation into the physical burdens that nurses face while handling bed-bound patients, especially in cases of higher patient weights. Adhering to upper-body posture guidelines outlined by ISO and DIN EN standards could potentially alleviate the risk of musculoskeletal strain [16,17]. Moreover, technical training and ergonomic supervision have shown promising results in reducing work-related injuries and discomfort among nurses [32,33,37,39].

To conclude, this study illuminated the intricate interplay between repositioning techniques, nurses' upper body postures, and the resulting acceleration experienced by patient participants, as well as nurses' perceived exertion. By adhering to ergonomic guidelines and embracing technical training, the nursing community can work toward safer and more effective patient-handling practices. Innovative tools and ongoing research stand to contribute significantly to enhancing patient care and alleviating the physical strain faced by nurses in their vital roles.

Limitations

The interpretation of the study's outcomes should consider its inherent limitations. Chiefly, a noteworthy technical restriction pertains to the constrained number of participating nurses—a mere two trained individuals. This restricted representation potentially curtails the findings' applicability to a broader healthcare professional population. However, according to the actual situation in care settings, explorative feasibility studies like ours should give first insights into potential requirements for future studies. Augmented generalizability and more robust conclusions mandate an expanded, more heterogeneous sample size. Nevertheless, it has to be noted that the results were derived by a total of six positioning trials per person (both sides triple; N = 114 trials). The study's outcomes, while aligned with extant literature concerning the challenges nurses face in manual patient handling, might be somewhat circumscribed due to the diminutive sample size, affecting comparative analyses and correlations with previous research endeavors. Furthermore, the exclusive focus on bilateral 30° turning maneuvers might not comprehensively encompass the array of techniques nurses encounter in practical settings. This selected scope limits the spectrum of scenarios under scrutiny.

Another noteworthy limitation pertains to the reliance on self-reported perceived exertion data garnered from nurses during the manual repositioning procedures. The integration of subjective measurements introduces inherent biases and reporting disparities. Objective measurements, such as electromyography or force sensors, would proffer more precise and reliable data regarding the physical demands placed on nurses. Employing these objective metrics would enhance the fidelity of the collected data. Furthermore, the external validity of the study is restricted by the study's specific context and sample composition. Healthy individuals serving as representatives for bed-bound patients might not comprehensively replicate the multifaceted challenges nurses encounter when dealing with a diverse array of patient conditions. Thus, careful consideration is requisite when extrapolating the results to different healthcare scenarios.

Nevertheless, acknowledging these limitations, the study provides invaluable insights into the challenges confronted by nurses in the realm of manual patient handling, especially in manual patient repositioning processes, underscoring the importance of patient-centered and ergonomic practices. Subsequent research endeavors, enriched by larger and more diverse participant samples, enhanced methodological frameworks, and the incorporation of objective measurements, stand to build upon these insights. Such future investigations hold the potential to propel our understanding of nurse well-being and the quality of patient care in the context of manual handling, especially in the repositioning process, to new heights.

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Institutional Review Board Statement: The ethical approval was obtained from the local ethics committee of the University of Hamburg (2019_259). All participants gave their written informed consent. The study followed the Declaration of the Helsinki (version of 2013) and was registered at the German Clinical Trials Register (DRKS00024389; 24.03.2021).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The original contributions presented in the study are included in the article, further inquiries can be directed to the corresponding author.

Conflicts of Interest: Author Andreas Argubi-Wollesen was employed by the company ExoIQ GmbH. The remaining authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest. The funders had no role in the design of the study; in the collection, analyses, or interpretation of data; in the writing of the manuscript; or in the decision to publish the results.

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The Effects of Concussions on Static Postural Stability

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Abstract: Concussions among the athletic population are extremely common, which could lead to postural instabilities. The purpose of this study was to assess the effect of concussions on postural stability in young healthy adults. The static postural stability of thirty volunteers (age 21.63 ± 2.50 years; height 1.70 ± 0.14 m; weight 75.00 ± 15.58 kg; 17 with a history of concussions) was assessed using a force platform during three tests: baseline stability test, clinical test of sensory interaction and balance test, and unilateral stability test. Postural sway variables during each test were statistically analyzed using an independent *t*-test between the concussion group (CONC) and no concussion (NO CONC) groups. Two secondary analyses were performed with the CONC group: individuals who had one concussion (ONCE) vs. who had multiple concussions (MULTIPLE) and individuals who had their last concussion in 2023–2018 (RECENT) and in 2017–2011 (OLD). The CONC, MULTIPLE, and RECENT groups demonstrated greater postural sway than the NO CONC, SINGLE, and OLD groups. Concussions cause postural decrements in young healthy adults compared to their counterparts with no history of concussions. The results of the study exhibit that concussions could lead to imbalances, which is decisive in athletes' performance and injury risk during play.

Keywords: concussions; static; balance; CTSIB; force plate; mTBI; postural stability

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1. Introduction

Concussions are defined as a complex pathophysiological process affecting the brain, induced by traumatic biomechanical forces [1]. Concussions could be identified as a significant intrinsic factor that could affect an athlete's postural stability. Postural control must happen in order to maintain a myriad of activities and postures [2], and thus, the deficits can result from a sensory, motor, or cognitive impairment [2]. McLeod [3] found that ~67–77% of the people who had a concussion experienced dizziness, which resulted in a risk factor for a longer recovery. Other symptoms of concussions include facial and neck pain, blurred vision, sleep disturbance, headache, memory and concentration [1,2]. However, the non-modifiable and modifiable risk factors of concussions are not well understood [4]. Concussions can affect everyday life, whether that be in sports, recreation, or work [5], and can affect someone's performance and injury risk.

Having a concussion diagnosed immediately after a blow or hit to the head allows for an active management approach, which can result in faster recovery and prevent a secondary injury [5]. In McLeod's [3] literature review, the authors found that 3–10 days post-injury, the concussed individuals had balance impairments. Research shows that most people who have suffered a concussion recover completely, but about 10–15% of individuals who have had a concussion do not recover completely and suffer from post-concussion syndrome (PCS) [6]. More specifically, some research specifies that it can take up to 3 to 6 months to recover [7]. Roughly 40% of concussed people have impairments that affect their daily life [8–10], while 25% still cannot return to work one-year post-injury [8]. Furthermore, some people have persistent symptoms that require active rehabilitation [5]. The outcome of concussions depends on the severity of the concussion. Clinically, the individuals with concussions are categorized as grade 1, grade 2, and grade 3, depending on their symptoms. In this categorization, amnesia, confusion, and loss of consciousness are the main symptoms to consider. There are different guidelines that exist for this categorization, such as the Colorado Sports Concussion Grading Scale, the American Academy of Neurology Concussion Grading Scale, and the Cantu Sports Concussion Grading Scale. In all grading scales, transient confusion is considered grade 1, and loss of consciousness is considered a grade 3 [3].

Lifetime concussion symptoms lead to an increased risk of falls and motor disturbances [11]. Decreased postural control and neurocognitive function are both side effects of someone who has had a concussion [12]. As well as over 23% of concussed people complain about being dizzy [4]. Dizziness can be a result of either inner ear disorders and central nervous system disorders [4]. Some of these disorders include benign paroxysmal positional vertigo and post traumatic migraine [4]. The central nervous system (CNS) plays a key role in postural equilibrium [13]. For the CNS to maintain equilibrium and execute coordinated and appropriate musculoskeletal responses, it must process and integrate information from the somatosensory (proprioceptive), vestibular, and visual systems [13]. The feedback received from the peripheries must be processed in the CNS to send messages to the muscles in the extremities to contract appropriately for the body to maintain postural stability [13]. Thus far, the results from studies that investigated the effects of concussions on balance are controversial. Lee [14] found that postural stability was not affected by a concussion. Guskiewicz [13] found that when the task became more complex by altering the somatosensory, vestibular, or visual feedback during the trial, the balance impairment increased in the individuals who had a mild traumatic brain injury (mTBI). (The terms concussion mild traumatic brain injury (mTBI) are used interchangeably in this paper, as they are in other research articles such as Junn [1] and Lefevre-Dognin 2020 [5]).

Guskiewicz [6] and colleagues attributed their findings to a sensory interaction problem by receiving various combinations of information from the visual, vestibular, and somatosensory systems. Kunker [15] found that when a participant with a history of a concussion relied on visual cues with their eyes open during testing, they had an increased displacement of COP (center of pressure) in the mediolateral direction, increased COP area, and reduced path length when compared to individuals who never had a concussion. Whereas Lee [14] reported no functional deficits of postural stability performances in the individuals with a mTBI while using the force platform to assess balance. With the mixed results of past research, this study aimed to investigate the effect of previous concussions on postural stability in young healthy adults. It was hypothesized that any history of concussions would have a negative effect on postural stability. Additionally, it was hypothesized that multiple concussions and time since the last concussion would cause greater balance decrements compared to single and old concussions.

2. Materials and Methods

A total of thirty healthy collegiate volunteers (age 21.63 ± 2.50 years; height 1.70 ± 0.14 m; weight 75.00 ± 15.58 kg; 9 males; 2 left leg dominants; 17 with a history of concussions) without current or recent history of neurological, visual, vestibular, or musculoskeletal abnormalities were recruited for the study. Participants with a history of lower extremity fractures, surgery, back pain, or joint sprains within the last three months were excluded from the study. The study was approved by the institutional review board (IRB-AY22-23-301). The sample size was calculated using G*Power statistical software (3.1.9.6) with an effect size of 0.25, a desired power of 0.8, and an alpha level of 0.05.

The study consisted of a single session for each participant, conducted over the course of one day. Participants were instructed to refrain from vigorous leg workouts the day before testing. Upon arrival at the laboratory, informed consent was obtained, followed by completing a Physical Activity Readiness Questionnaire (PARQ+) to inquire about any existing pathology. To get the concussion history, the participants answered an online-based survey, which asked whether they had any concussions, how many concussions they had, how many concussions were diagnosed by a medical professional (e.g., medical doctor, nurse, athletic trainer, etc.), when their last concussion was, and the activity(ies) led to their concussion(s). Then, the participants' demographics and anthropometric data were recorded, and the dominant leg was determined using the ball-kick test (i.e., asking the participants, "If I gave you a ball to kick, which leg you would use") [16]. After recording these, they were familiarized with the study protocols, including standing on the force plate in a standardized body position. During each trial, the participants were advised to stand in the middle of the force plate, feet shoulder-width apart, keeping the arms along the body, looking forward with gaze fixed, without talking. The participants were allowed to practice standing on the force plate during familiarization.

Following familiarization, the testing started. The protocol included three tests: baseline stability test, Clinical Test of Sensory Interaction and Balance (CTSIB) test, and unilateral stability test administered in the same order for every participant. Static postural stability was assessed using an AMTI force platform (Advanced Mechanical Technology, Inc., Watertown, MA, USA) at a rate of 100 Hz and analyzed using bioanalysis software version 2.2 (Watertown, MA, USA) [17]. During the baseline stability test, the participants' static postural stability was assessed while standing on the force plate, on both feet, and with their eyes opened. Three 20-s trials were recorded for this test. The CTSIB test included four conditions: standing on the force plate on both feet and eyes opened (firm BL EO condition), standing on the force plate on both feet and eyes closed (firm BL EC condition), standing on a foam pad placed on the force plate on both feet and eyes opened (foam BL EO condition), and standing on a foam pad placed on the force plate on both feet and eyes closed (foam BL EC condition) (Figure 1). During this test, three trials were recorded for each condition, each 20 s long. The unilateral stability test consisted of four conditions: standing on the force plate on the dominant leg with eyes opened (DOM EO condition), standing on the force plate on the non-dominant leg with eyes opened (ND EO condition), standing on the force plate on the dominant leg with eyes closed (DOM EC condition), and standing on the force plate on the non-dominant leg with eyes closed (ND EC condition) (Figure 1). During this test, three trials were recorded for each condition, each 10 s long.

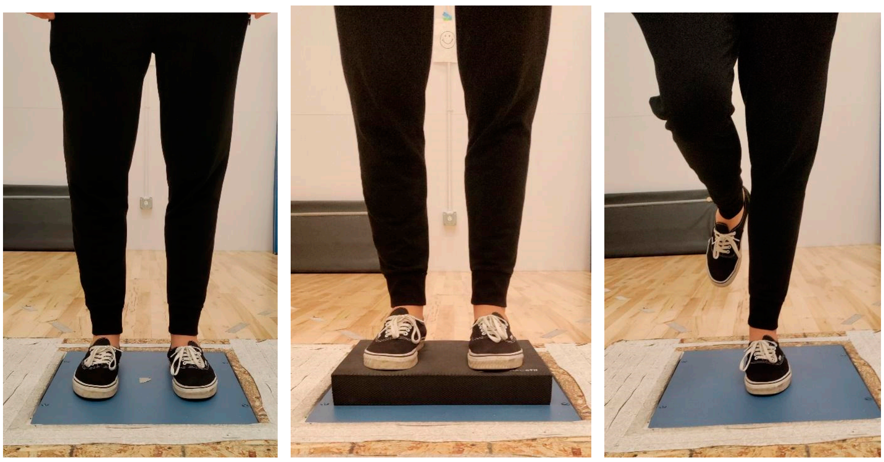


Figure 1. (**Left**): A participant standing on the force platform on both feet during the clinical test of sensory interaction and balance (CTSIB) test; (**Middle**): A participant standing on a foam pad placed on the force platform on both feet during CTSIB test; (**Right**): A participant standing on the force platform on one foot during unilateral stability test.

As the postural sway variables of interest, the maximum COP excursion in the anterior-posterior direction (COP-X max), minimum COP excursion in the anterior-posterior direction (COP-X min), maximum COP excursion in the medial-lateral direction (COP-Y max), minimum COP excursion in the medial-lateral direction (COP-Y min), average COP displacement in the anterior-posterior direction (Avg. Displacement along X), average COP displacement in the medial-lateral direction (Avg. Displacement along Y), and 95% ellipsoid area (95 EA) were chosen. Lower values for the postural sway variables indicate greater postural stability and vice versa.

Participants who had at least one medically diagnosed concussion were put in the concussions (CONC) group, and the others were placed in the no-concussion (NO CONC) group. Postural sway data from each test [baseline stability test, CTSIB test, and unilateral stability test] were analyzed separately using an independent *t*-test for CONC vs. NO CONC groups. Furthermore, two secondary analyses were performed with the CONC group. In one analysis, the CONC group was divided into two groups: the individuals who had one concussion (ONCE) and the individuals who had multiple (>1) concussions (MULTIPLE). In the other analysis, the participants were divided into two groups: those who had their last concussion in the years 2023–2018 (RECENT) and those who had their last concussion in the years 2017–2011 (OLD). In both these analyses, postural sway variables were analyzed using an independent *t*-test for ONCE vs. MULTIPLE and RECENT vs. OLD groups. Statistical analyses were performed using JASP (Version 0.18.1; Amsterdam, The Netherlands) at an apriori alpha level of 0.05.

3. Results

Out of the 30 participants, 17 had a history of medically-diagnosed concussions, while 13 did not. Eight participants had a history of one concussion, while nine had multiple concussions (five with two concussions, two with three concussions, and two with four concussions). The participants have had their last concussion in the years ranging from 2023–2011. Nine of them had their last concussion in the years 2023–2018 (two in 2018, one in 2019, two in 2020, two in 2021, one in 2022, and one in 2023), while the others had their last concussion in the years 2017–2011 (one in 2011, one in 2013, one in 2014, one in 2015, one in 2016, and three in 2017). As the circumstances lead to concussions, the participants listed sporting events (13 participants), falling/slipping/tripping (3 participants), and heavy objects falling onto the head (1 participant).

3.1. Baseline Stability Test

There was no significant difference in any sway variable between the CONC and NO CONC groups, ONCE and MULTIPLE groups, or RECENT and OLD groups.

3.2. Clinical Test of Sensory Interaction and Balance (CTSIB) Test

The independent *t*-test of CONC vs. NO CONC groups revealed a significant difference in postural sway between groups in COP-X max ($p = 0.036$; $t = 2.21$; $d = 0.81$), Avg. Displacement along X ($p = 0.016$; $t = 2.57$; $d = 0.95$), and 95 EA ($p = 0.043$; $t = 2.12$; $d = 0.78$) sway variables during the foam BL EO condition (Figure 2). The descriptive statistics for all three variables demonstrated higher values for the CONC group than the NO CONC group. No significant differences between CONC vs. NO CONC groups were observed in firm BL EO, firm BL EC, or foam BL EC conditions.

The secondary analysis of ONCE vs. MULTIPLE groups revealed a significant difference in postural sway between groups in COP-Y max ($p = 0.027$; $t = 2.46$; $d = 1.20$) and 95 EA ($p = 0.037$; $t = 2.29$; $d = 1.11$) sway variables during the foam BL EO condition. The descriptive statistics for both variables demonstrated higher values for the ONCE group than the MULTIPLE group. No significant differences between ONCE vs. MULTIPLE groups were observed in firm BL EO, firm BL EC, or foam BL EC conditions. Furthermore, the secondary analysis of RECENT vs. OLD groups revealed a significant difference in postural sway between groups in COP-X max ($p = 0.019$; $t = 2.63$; $d = 1.28$) in the foam BL EC condition with higher values for the RECENT group than the OLD group.

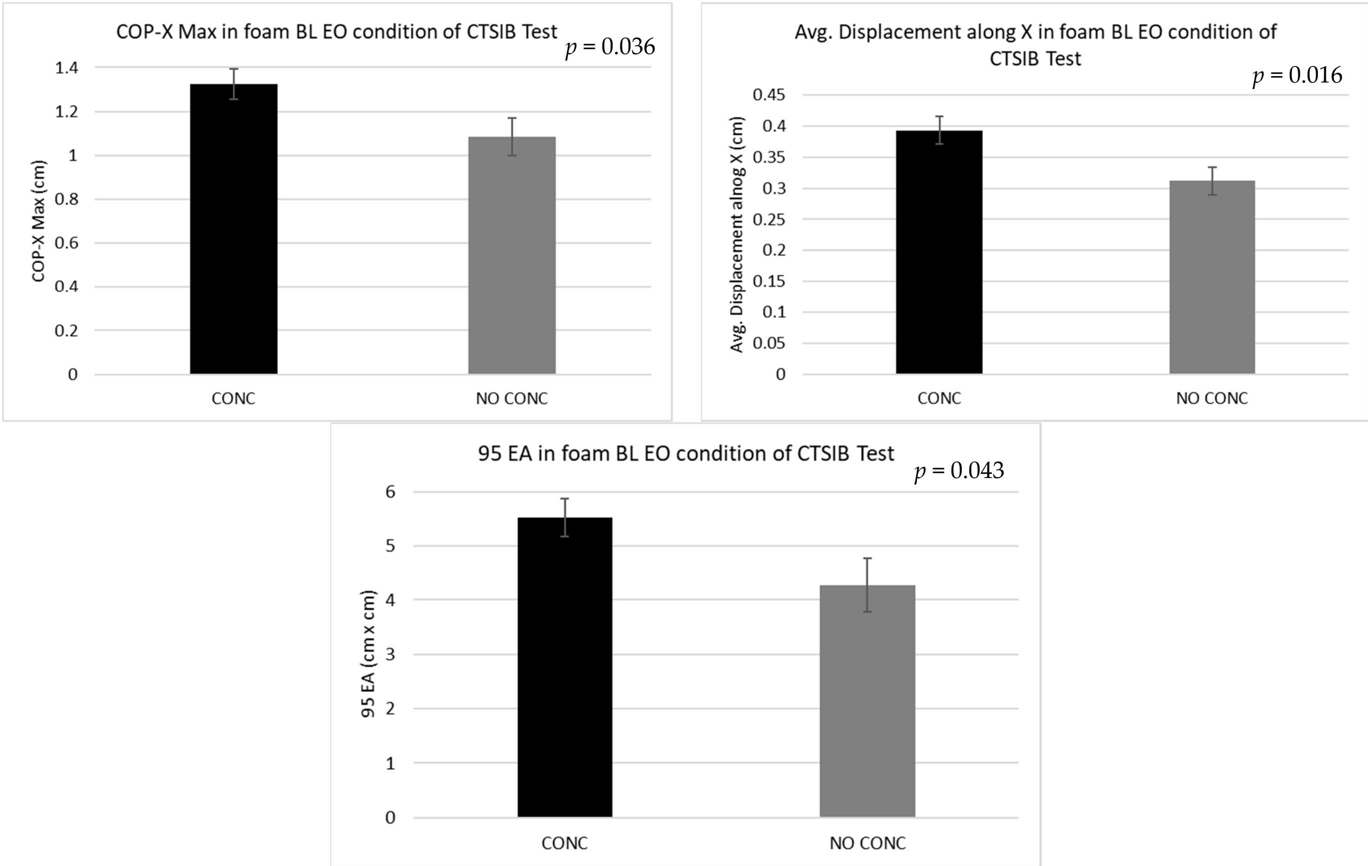


Figure 2. Significant postural sway variables between the individuals with a history of concussion (CONC) and those without concussion (NO CONC) when standing on a foam pad placed on the force platform on both feet with eyes opened (foam BL EO) condition of Clinical Test of Sensory Interaction and Balance (CTSIB) Test. (**Top left**) COP-X max; maximum center of pressure (COP) excursion in the anterior-posterior direction. (**Top right**) Avg. Displacement along X; average COP displacement in the anterior-posterior direction. (**Bottom**) 95 EA; 95% ellipsoid area. Bars represent standard errors.

3.3. Unilateral Stability Test

During this test, significant differences in sway between CONC and NO CONC groups were evident in the DOM EC and ND EC conditions. In DOM EC condition, COP-X max ($p = 0.002$; $t = 3.41$; $d = 1.26$), COP-Y max ($p = 0.045$; $t = 2.09$; $d = 0.77$), Avg. Displacement along X ($p = 0.011$; $t = 2.72$; $d = 1.00$), Avg. Displacement along Y ($p = 0.032$; $t = 2.26$; $d = 0.83$), and 95 EA ($p = 0.024$; $t = 2.38$; $d = 0.88$) demonstrated significant differences with greater postural sway in the CONC group compared to the NO CONC group (Figure 3). Similarly, in ND EC condition, COP-Y max ($p = 0.042$; $t = 2.14$; $d = 0.79$), COP-Y min ($p = 0.046$; $t = 2.09$; $d = 0.77$), Avg. Displacement along Y ($p = 0.009$; $t = 2.80$; $d = 1.03$), and 95 EA ($p = 0.043$; $t = 2.12$; $d = 0.78$) demonstrated significant differences with greater postural sway in the CONC group compared to the NO CONC group (Figure 4). The descriptive statistics for all variables demonstrated higher values for the CONC group than the NO CONC group in both conditions. No significant differences between CONC vs. NO CONC groups were observed in DOM EO or ND EO conditions.

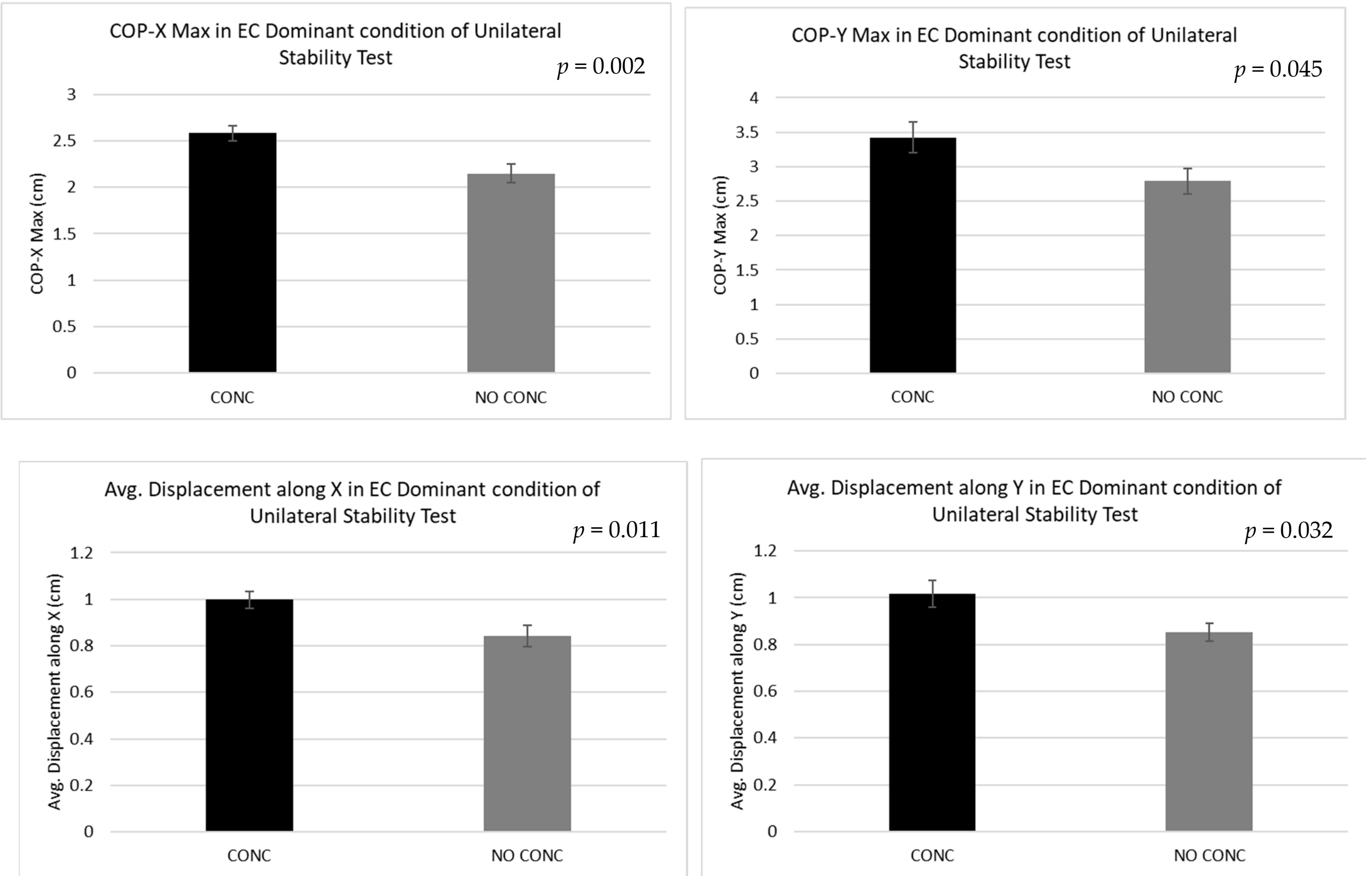


Figure 3. Significant postural sway variables between the individuals with a history of concussion (CONC) and those without concussion (NO CONC) in eyes closed (EC) dominant condition of the unilateral stability test. (**Top left**) COP-X max; maximum center of pressure (COP) excursion in the anterior-posterior direction. (**Top right**) COP-Y max; maximum COP excursion in the medial-lateral direction. (**Bottom left**) Avg. Displacement along X; average COP displacement in the anterior-posterior direction. (**Bottom right**) Avg. Displacement along Y; average COP displacement in the medial-lateral direction. Bars represent standard errors.

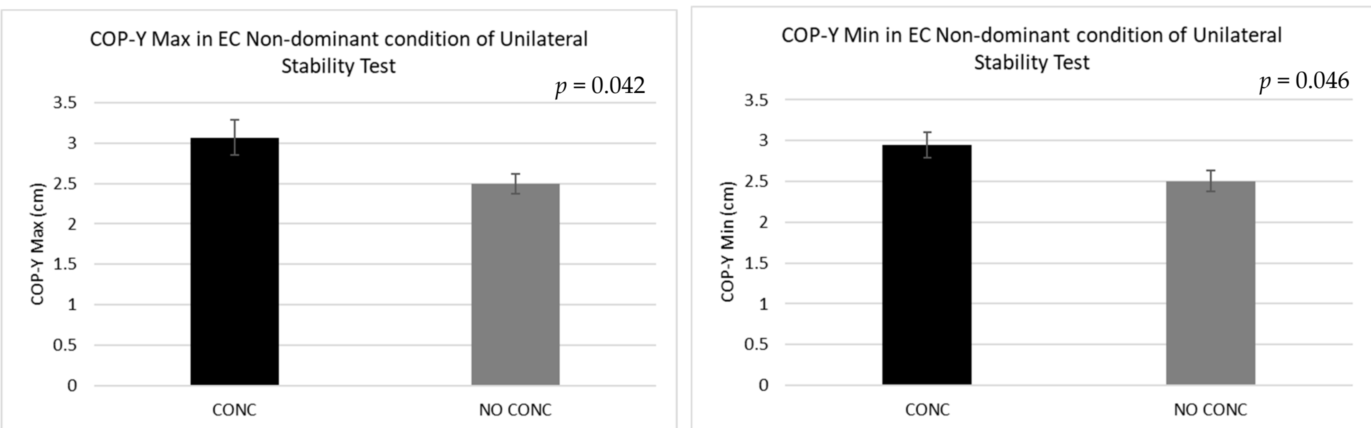
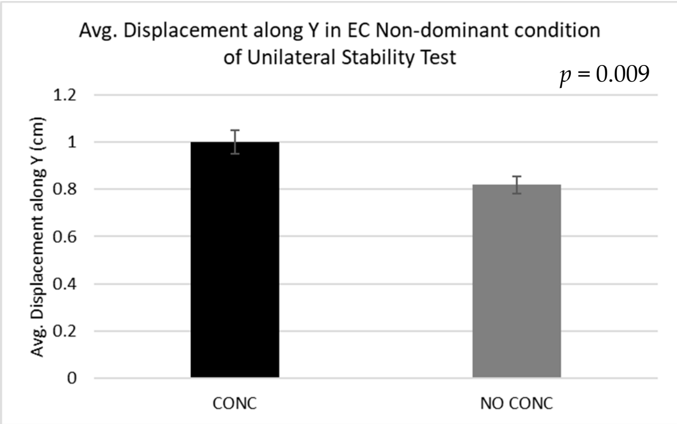


Figure 4. *Cont.*



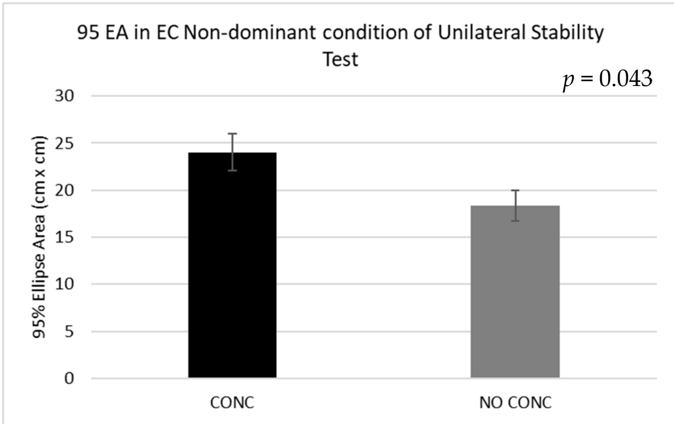


Figure 4. Significant postural sway variables between the individuals with a history of concussion (CONC) and those without concussion (NO CONC) in eyes closed (EC) non-dominant condition of the unilateral stability test. (**Top left**) COP-Y max; maximum center of pressure (COP) excursion in the medial-lateral direction. (**Top right**) COP-Y min; minimum COP excursion in the medial-lateral direction. (**Bottom left**) Avg. Displacement along Y; average COP displacement in the medial-lateral direction. (**Bottom right**) 95 EA; 95% ellipsoid area. Bars represent standard errors.

The secondary analysis of ONCE vs. MULTIPLE groups revealed a significant difference in postural sway between groups in the COP-Y average ($p = 0.019$; $t = 2.62$; $d = 1.27$) during the ND EO condition. The descriptive statistics demonstrated higher values for the ONCE group than the MULTIPLE group. No significant differences between ONCE vs. MULTIPLE groups were observed in the DOM EO, DOM EC, and ND EC conditions.

4. Discussion

The purpose of this study was to assess the effect of concussions on postural stability in young healthy adults. It was hypothesized that a history of concussions would have a negative effect on postural stability. Additionally, it was hypothesized that experiencing multiple concussions and recent concussions would cause greater balance decrements compared to single and old concussions. Overall, the results demonstrated poor postural stability in the group with a concussion history compared to the group that does not have a concussion history.

4.1. Effects of Concussion History on Postural Stability

In the analyses of postural sway variables between the participants with a concussion history (CONC) and without a concussion history (NO CONC), the results revealed that the CONC group had higher postural sway than the NO CONC group, indicating greater postural decrements. More specifically, the CONC group demonstrated greater COP-X max, avg. displacement along X, and 95 EA in the foam BL EO condition of CTSIB. The CONC group also demonstrated greater COP-X max, COP-Y max, avg. displacement along X, avg. displacement along Y, and 95 EA in DOM EC condition, as well as COP-Y max, COP-Y min, avg. displacement along Y, and 95 EA in ND EC condition of unilateral stability test. In essence, the CONC group showed a greater postural sway in both anterior-posterior and medial-lateral directions compared to the NO CONC group. These findings agree with the researchers' original hypothesis that concussions would have a negative effect on postural stability, as well as with the previous studies [3,15,18].

Postural stability decrements following concussions can occur due to multiple reasons. Postural stability in humans is mainly maintained by three main systems: the afferent system, the CNS, and the efferent system. The afferent system consists of the three visual, vestibular, and somatosensory systems, with healthy individuals' somatosensory system contributing to 70% of the postural control in a well-lit room on a firm surface [19]. The CNS consists of higher brain centers, while the efferent system consists of the musculoskeletal system [20]. Coordination between these systems is required for balance, and any effect on these three systems can affect someone's postural stability. Concussions (and the majority of other traumatic brain injuries) are well-known to affect the vestibular system [21,22]. The vestibular organ located in the inner ear is mainly comprised of three semi-circular canals that detect angular acceleration of the head and the two maculae (saccule and utricle) that detect linear movement as well as the effects of gravitational force on the body. Moreover, endolymph and hair cells in semicircular canals, along with the hair cells in maculae, detect the direction of body movements [23]. Thus, any disruption to the anatomy of this highly sensitive vestibular system (especially endolymph, hair cells, and natural orientation of the structures) affects the person's postural stability. As concussions result following a severe collision of the head against another surface, it interferes with the vestibular system, which explains the headache, dizziness, lightheadedness, nausea/vomiting, and tinnitus following concussions. Furthermore, the symptoms of photophobia and blurred vision could suggest a disturbance to the individual's visual system, affecting the contribution of their visual system in postural control. The visual system, consisting of eyes, optical tracks, and optic nerves gathers information on the surrounding environment to maintain balance and is considered the fastest sensory system out of the three and hence capable of reacting to acute perturbations [24]. Thus, interruptions to the function of visual systems affect someone's ability in postural stability [25].

In the present study, there were no significant differences in balance between CONC vs. NO CONC groups during the baseline stability test. This was not unexpected since the participants were young healthy adults (age 21.63 $\pm$ 2.50 years) who have excellent postural control mechanisms and were shown to recover soon after concussions [26]. Moreover, in the baseline stability test, the participants were standing on a firm surface (force plate), on both feet, with their eyes open, which is not a challenging condition for balance maintenance. The significant differences in balance between CONC vs. NO CONC groups were only observed in foam BL EO condition, and no differences were observed in firm BL EO, firm BL EC, and foam BL EC conditions of CTSIB. In "foam" conditions of CTSIB, the individuals stand on a foam pad placed on the force plate, which is an unstable surface for the participants. In EO conditions, the participants were advised to keep their gaze fixed at a specific point. Thus, the participants sway due to the unstable surface, but the gaze is fixed, leading to conflicting feedback from somatosensory and visual systems, causing postural instability. In healthy individuals, their vestibular system continues to provide correct information even when erroneous or conflicting information is received via other sensory systems [27]. Previous research suggests that the impact of concussions appears to be most prominent on the vestibular system [20,21]. Therefore, the vestibular system of the participants in the CONC group could be already negatively affected by past concussions, leading to compromised corrections made by their vestibular system. This is also supported by the results that the CONC group performed poorly when the tasks required a greater reliance on the vestibular system. During the unilateral stability test, declined performance was observed only when visual input was prevented (in ND EC and DOM EC conditions, but not in DOM EO and ND EO conditions). Further, when there was a significant reduction in their base of support (the unilateral stability test), a greater contribution of the vestibular feedback was required to maintain balance, which could have led to significant findings in UL EC conditions.

4.2. Effects of Multiple Concussions on Postural Stability

In the analyses of postural sway variables between the participants who had a single concussion (ONCE) and multiple concussions (MULTIPLE), the results revealed that the ONCE group had higher postural sway compared to the MULTIPLE group, indicating greater postural decrements. More specifically, the ONCE group demonstrated greater COP-Y max and 95 EA in foam BL EO condition of CTSIB, as well as average displacement along Y in ND EO condition of unilateral stability test. This finding that the participants with a history of one concussion (ONCE) had a greater postural sway in both anterior-posterior and medial-lateral directions compared to the group who had a history of multiple concussions (MULTIPLE) contradicts the researchers' original hypothesis that multiple concussions would have a greater negative effect on postural stability compared to a single concussion.

These findings could have occurred due to multiple reasons. One reason could be testing the young healthy individuals who are shown to recover from concussions faster and better [26]. Additionally, in the present study, the severity of the concussion(s) and post-concussion management/therapy that the study participants underwent were not taken into consideration, which could have affected the results. For example, an individual who sustained a grade 3 concussion once could potentially have higher balance decrements compared to an individual who had two grade 1 concussions. Recent evidence indicates that early initiation of clinical interventions, including vestibular therapy following concussion, positively impacted the recovery time [28,29] and functional outcomes [30]. Moreover, in the current study, the timing of concussion(s) was not taken into consideration. Therefore, a participant who had a single concussion recently will have more effects on postural stability compared to someone who had multiple concussions a few years ago. The authors assume the results would differ if the severity and timing of concussions were taken into consideration when categorizing participants into ONCE and MULTIPLE groups. In addition to these, the individual participants' activity level, training history, and type of footwear worn could have contributed to the findings.

4.3. Effects of Timing of Concussions on Postural Stability

In the analyses of postural sway variables between the participants who had their last concussion in the years 2023–2018 (RECENT) and those who had their last concussion in the years 2017–2011 (OLD), the results revealed that the RECENT group had higher postural sway compared to the OLD group, indicating greater postural decrements. More specifically, the RECENT group demonstrated greater COP-X max in foam BL EC condition of CTSIB, showing greater postural sway in the anterior-posterior direction compared to the OLD group. This finding agrees with the researchers' original hypothesis that recent concussions would have a greater negative effect on postural stability compared to old concussions. This finding also agrees with the previous research that showed the persisting effects of concussions [3,31,32].

As mentioned above, concussions disrupt the anatomy and physiology of vestibular and visual systems, leading to postural instability. This disruption is pronounced during the immediate three weeks after concussion and is considered to resolve by three months post-concussion [31]. However, this recovery depends on the person and multiple other factors, such as age, activity level, previous history of concussions, the severity of the concussion, co-existing neurological disorders, stress, and pre-mature return to work/play, where some individuals were shown to have lingering effects until a few years post-concussion [31]. As such, they could have postural decrements lasting from immediately after the concussion to a few years until the vestibular system returns to normal. Additionally, concussions cause post traumatic vestibular migraines, positional vertigo, and spatial disorientation, which can have a significant effect on someone's postural stability [21].

The findings of this study have a few practical implications. Since it is evident that concussions cause postural instabilities, it would be a decisive factor when determining an athlete's return to play following a concussion. Premature return to play after a concussion is known to increase their risk of PCS and possibly fatal second impact syndrome, it is crucial to determine the appropriate time to return [33]. As the timing of full recovery is still unclear, balance and gait assessment should be a part of the concussion assessment. Moreover, since disruption of the vestibular structure and function is the major reason for the postural issues following concussions, vestibular rehabilitation therapy (VRT) could improve the recovery time [28,29]. Additionally, it has been shown that when VRT is combined with cervical therapy, it allows the athletes to be able to return to play within eight weeks of their concussion [29].

This study is not without limitations. Not considering the severity and timing of each concussion was a limitation, which narrowed the possibility of further assessing the effects of multiple concussions on postural stability. As the sample size of the present study was small ($n = 30$; 17 had a history of concussions), the participants were not categorized according to the severity or the timing of the concussion. Thus, having a smaller sample in the CONC group could also have affected the secondary analyses (MULTIPLE vs. ONCE and RECENT vs. OLD comparisons). Hence, future studies could be focused on recruiting a larger sample with a concussion history and considering potential confounding factors, including the severity and timing of concussions and clinical interventions such as VRT in analyses. Furthermore, this study recruited young healthy adults, who are known to have a rapid and excellent recovery; geriatric and clinical populations would yield different results. Additionally, future studies could focus on investigating the timing of full recovery following concussions.

5. Conclusions

In conclusion, the present study demonstrated concussions cause postural decrements in young healthy adults compared to their counterparts who did not have a history of concussions. The secondary analyses showed greater instability among the individuals who had recent concussions compared to the old concussions, as well as in the individuals who had one concussion compared to their counterparts who had multiple concussions. The results of the present study demonstrate that concussions could lead to impaired balance, which is decisive in athletes' performance and injury risk during play. Moreover, the results show that the postural effects of concussion could last for years, and therefore, it is mandatory to perform a balance and gait assessment before returning to play. As the timing of full recovery is yet debatable, further studies are warranted to ascertain the perfect time to return.

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Institutional Review Board Statement: The study was conducted in accordance with the Declaration of Helsinki and approved by the Institutional Review Board (or Ethics Committee) of Weber State University (protocol code IRB-AY22-23-301 and 5 February 2023).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The data presented in this study are available on request from the corresponding author due to privacy reasons.

Conflicts of Interest: Author Paul Brewer is employed by the company Diverge Neurovision. The remaining authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Injury Incidence in Traineras: Analysis of Traditional Rowing by Competitive Level and Gender

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Abstract: The growing interest in "Traineras", a traditional competitive rowing modality prevalent in Northern Spain, underscores the need for a comprehensive analysis of the injury incidence associated with this sporting practice. Despite rowing's significance in the international sports arena and its inclusion since the beginnings of the modern Olympic Games, research into injuries in this sport, especially in traditional modalities such as Traineras, has been limited. This study aimed to identify and describe the predominant injuries among Traineras rowers, analyzing their epidemiology, characteristics, affected body regions, and diagnoses, further differentiated by competitive level and gender. A retrospective survey completed by 773 rowers (24% women, 76% men) participating in various leagues (ACT, ARC1, ARC2, LGT1, LGT2, ETE, and LGT-F) during the season revealed that 68.2% suffered from at least one injury, predominantly due to overuse (91.1% in men, 83.1% in women). The most affected regions were the lower back and shoulders, with the main diagnoses being muscle cramps and tendinitis, showing statistically significant differences between sexes. The findings of this study not only provide a deeper understanding of the etiology and origin of injuries in this sport but also lay the groundwork for developing specific injury prevention plans, thereby contributing to the safety and optimal performance of athletes.

Keywords: sports injuries; low back; injury incidence; overuse; performance level

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1. Introduction

Rowing sports are categorized into three main disciplines: outdoor flat-water rowing, open water disciplines [1], and indoor variants. Traditional rowing can be considered an open water discipline and often employs a fixed seat, in contrast to the more widespread Olympic sliding-seat rowing, which consequently has been subject to more extensive research [2–4]. Given their similarities as sports in relation to the phases and paddling cycle, comparisons between traditional and Olympic rowing have been common in the literature [1,2,5].

Traditional rowing typically involves athletes using a single oar gripped with both hands and oriented either to the port (right) or starboard (left) side. The Olympic category includes events such as the eight (a boat for eight sweep rowers with a coxswain for direction), the four (for four sweep rowers), and the pair (for two sweep rowers), with options for coxed or coxless boats [6]. Despite the diversity of disciplines, all rowing forms require a cyclical movement pattern where the synchronization of leg and arm work is essential for maximizing stroke efficiency and achieving peak performance [7]. Although fixed-seat rowing modalities are widespread throughout the Spanish peninsula [8], Traineras rowing can be considered the modality that has acquired most importance in recent years [9].

In Olympic rowing, all athletes need to perform over 2000 m, women and men, while in Traineras rowing, 5556 m, or three sea miles, must be completed in the shortest possible time during a regatta for males [1–4,10], and half of this distance, or 2778 m, needs to be completed by women [11]. Moreover, approximately between 5′30″–7′ effort time (200–240 strokes) is required in Olympic rowing depending on the boat type and weather conditions, while 19′–20′ (700–800 strokes) are needed in a Traineras male rowing competition [1,2,10], and 11′–12′ (350–450 strokes) in women regattas, time that can be extended when the sea conditions become rough. Another characteristic that makes traditional rowing different from Olympic rowing is the *ciaboga* maneuver, in which male Traineras turn around three times and women once in each regatta, which means that the boat loses speed that need to be recovered after each *ciaboga* [2], instead, in Olympic rowing boats go straight from the beginning to the finish line. Strength-endurance capacity becomes crucial for performance in both, Olympic rowing [12] and traditional rowing [1]. Among the differences while Olympic rowing takes place in calm waters such as rivers or lakes, traditional rowing modalities such as Traineras are mostly played on the sea [1,2,10], and their main characteristic is that the seat is fixed, so rowers do not use a sliding seat to propel on the sea, and instead a more pronounced flex-extension of the trunk is required [1,10].

Among all rowing modalities, Olympic rowing is the most researched worldwide [13], and it can be considered the main reference for many fixed-sit rowing modalities. In Olympic rowing research, different aspects have been analyzed e.g., physiological variables [6,7,14], physical training [6,15–18], nutrition [17,19,20], anthropometry [21–23], psychology [24,25], coaching [24,26], or technical and biomechanical aspects [27,28], seeking to improve athletes' performance. Likewise, we find how fixed-sit rowing has also been analyzed from different perspectives to seek athletes' performance [1,8,29]. e.g., form perspectives such as physiology [1,2,5,30–32], physical training [33–35], nutrition [2,36–38], anthropometry [2,3,39,40], leadership and behavioral based investigations [41], psychology [42,43] or biomechanical aspects [2,4].

Despite the investigations carried out to improve athletes' performance, everything changes when a sports injury occurs, and injuries across sports have been consistently reported [44]. For this reason, sports injuries and their prevention have become a crucial subject of study [45]. The early specialization and the high demand to improve athletes' performance have contributed to a significant increase in chronic and overuse injuries suffered by different level athletes [46].

Even though a recent investigation shows that team sports have a higher injury rate in Olympic sports than individual sports [47], for example, Olympic rowing is not considered a very harmful sport since it belongs to the group with the lowest incidence of injuries [48–53]. Moreover, due to their importance to athletes' performance and health, the epidemiology of injuries in rowing has been widely analyzed [52], and several investigations have described the incidence of injuries in Olympic rowing [49,54–56]. These main injuries are shown in an updated systematic review of injury epidemiology in Olympic rowing [57]. Within the scientific literature related to injuries in rowing, many articles have focused on athletes' injuries according to rowers' competitive level or age, with several investigations gathering the main injuries in elite rowing [51,55,58–61] and many others describing the main injuries age-wise [62], junior [63], collegiate rowers [64,65], or amateur rowers [66]. Likewise, high school rowers [65] and master rowers' injuries [50] also haven been analyzed. Several investigations show female rowers' injuries [57,67], while some compared these sport injuries between men and women rowers [59,68], showing a wide map of the main injuries in Olympic rowing.

Among the characteristics or types of injuries of Olympic rowers, mainly overuse, acute (traumatic), and chronic (overuse) injuries have been described [50,51,56,57,62]. Apart from the type, the injury region in the body has also been studied, showing that the highest injury incidence described are low back [56,68–80] and rib stress [67,81,82]. However, other injuries, such as wrist and forearm, knee, iliotibial band, shoulder, hip, and skin conditions [54,56,60] are common in rowing. Among the main injury diagnoses, tenosynovitis, spondylolysis, disk injuries, or stress fractures have been described [49,56,60,62], thus completing the specific injury pattern in this sport. Beyond the characteristics of the cyclic movement in rowing, biomechanics and rowers' injuries have also been associated in different publications, since the biomechanical analysis of the technique of each sport is connected to the aetiology of the injuries and therefore crucial to be able to subsequently prevent or reduce the effects of these injuries [56,60,69,83]. Moreover, due to the importance of not being injured during an athletic season, injury prevention plans have been released for team sports [45] and in rowing [60,83] to try to reduce the injury rate.

Despite the number of studies that analyze the etiology and injury incidence in Olympic rowing, fixed-sit rowing injuries have been scarcely analyzed [84,85]. A recent investigation in Maltese fixed-sit rowing, compared the musculoskeletal injuries in this rowing modality with a group of Olympic rowers, and results showed that for both, the low back and shoulder were the body regions with the highest injury incidence [84]. In the Spanish context, Mediterranean fixed-sit rowers' incidence of injury was described, and ankle, low back, and shoulder were the main injured body regions, while sprains and tendinitis were the highest described diagnoses, and overuse injuries occurred the most during, especially during the training period [85]. However, there is currently no research that analyzes the injuries in Traineras rowing. Therefore, due to its importance to athletes' performance and health, the main objectives of this research will be to describe the characteristics, body region, and incidence of injuries in traditional rowing, creating thus specific injury incidence patterns in Traineras rowing related to gender and competitive level of rowers.

2. Materials and Methods

2.1. Participants

A retrospective cohort study was carried out with Traineras rowers competing in different competitive leagues: elite (ACT-Eusko Label in men and Euskotren in women) and non-elite (ARC1, ARC2, LGT1, LGT2 in men and ETE, LGT-F in women). The survey was completed by 714 athletes, 389 men and 107 women suffered one or more injuries, while 166 men and 51 women suffered no injuries. The 389 men had a total of 597 injuries, 122 were from elite rowers and 475 were from non-elite rowers. The 107 women had a total of 178 injuries, 49 were from elite rowers and 129 from non-elite rowers. The participants were selected by convenience, and the only inclusion criteria was that they needed to be competing in any of the Traineras leagues during that season. Approval was obtained from the University of Alicante's Ethics Committee (protocol code UA-2023-06-14_1), and the study was conducted following the ethical principles of the Declaration of Helsinki.

2.2. Questionnaire

The online survey was adapted from a previously validated questionnaire [85], and the tool was readapted with some general and specific questions related to Traineras discipline and divided into different sections. In the first section there is an introduction to the study and informed consent. The second section details demographics and questions in relation to the trajectory of the sportsmen and sportswomen, such as years of experience, competitive level, or Traineras league in which they were competing at that season. In the third section, questions regarding rowers' injuries, answering first whether they had suffered or not from an injury, are discussed with the option to describe and specify in detail the suffered injuries.

2.3. Procedure

Different rowing clubs and coaches were contacted via email and phone during the competitive season and were provided with detailed information about the study, the objective, and the justification of the investigation. Through direct contact with the coaches, they explained and administered the questionnaire to their rowers. The survey was created using Google Docs technology [86], and it was tested before being released. Athletes received the necessary information about the study before completing the survey and provided written informed consent at the beginning of it. The participating rowers were provided with an email address to be able to ask any questions that they might have while filling out the questionnaire.

2.4. Statistical Analysis

In the current study, the Z-test for the comparison of proportions was employed to analyze the difference between two proportions or percentages in a sample. It will calculate the test statistic for the difference in proportions by approximating the normal distribution. The Z-test is a robust statistical tool that allows us to determine whether the observed differences between proportions are statistically significant or merely a result of chance. The null hypothesis is that the two proportions are equal, while the alternative hypothesis can be two-tailed, left-tailed, or right-tailed. The formula to calculate the Z-test statistic is:

$$Z = \frac{p_1 - p_2}{SEDP} = \frac{p_1 - p_2}{\sqrt{\frac{p_1(1-p_1)}{n_1} + \frac{p_2(1-p_2)}{n_2}}}$$

This test is based on the normal approximation of the binomial distribution. We aim to compare two proportions, p_1 and p_2, observed in two different groups of sizes, n_1 and n_2, respectively. The Z-statistic follows a normal distribution. The confidence interval is obtained using the formula, where SEDP corresponds to the standard error of the difference in proportions as calculated in the previous formula. If the p-value corresponding to the Z-test statistic is less than the chosen significance level (0.05 for a 95% confidence level), then the null hypothesis can be rejected.

3. Results

A total of 775 injuries were described, 597 by men and 178 by women rowers. Within men rowers, 122 injuries were described by elite athletes, while 475 injuries were reported by non-elite athletes. Among women athletes, 49 injuries were reported by elite athletes, while 129 were reported by non-elite athletes (Table 1).

3.1. Characteristics of Injuries According to Competitive Level and Sex

Table 1 shows a list with characteristics of injuries and different comparisons: the differences for the total sample of injuries between men and women, the differences according to the competitive level, among elite and non-elite rowers, and the differences within each group in injury characteristics. Within the first comparison group among men and women injuries, statistically significant differences were found in high volume and competitive periods ($p < 0.01$), in the mode, with traumatism and overuse injuries ($p < 0.01$), at the severity, in 4–7 days and < 21 days ($p < 0.05$), and for starboard rowers ($p < 0.05$).

According to competitive level, men showed statistical differences ($p < 0.05$) in eight characteristics (high volume and competitive period, new injury or recurrent, in the injury severity 8–21 days, in athletes that row at bow and with those rowing in starboard and both sides), while elite and non-elite women's injuries showed statistical differences ($p < 0.05$) in the injury mode, traumatism, and overuse ($p < 0.05$).

Within each analyzed group (period, moment, type, mode, severity, rower position, and boat side), Table 1 shows how there are statistically significant differences within period, timing, type, and mode of injury in the overall male and female samples and at the elite and sub-elite levels for both men and women. However, injury severity does not show any significance for any of the groups, and rowing position only shows significant differences for men, showing a higher number of injuries for rowers rowing in the middle of the boat, in the total men's sample, and in the men's non-elite group. Finally, in relation to the side of the boat on which they row, a higher number of injuries is shown for rowers rowing on the port side in non-elite male rowers and elite female rowers, compared to starboard and those rowers that row on both sides.

Table 1. Characteristics of injuries in elite and non-elite male and female rowers.

	Male Rowers				Female Rowers				
	All (n = 597)	Elite (n = 122)	Non-Elite (n = 475)	p	All (n = 178)	Elite (n = 49)	Non-Elite (n = 129)	p	p
	% (n)	% (n)	% (n)		% (n)	% (n)	% (n)		
Period									
High volumes	78.9 (471) †	88.5 (108) †	76.4 (363) †	0.003 #	68.0 (121) †	63.3 (31) †	69.8 (90) †	0.406	0.003 *
Competitive period	21.1 (126)	11.5 (14)	23.6 (112)	0.003 #	32.0 (57)	36.7 (18)	30.2 (39)	0.406	0.003 *
Moment									
Training	90.5 (540) †	91.0 (111) †	90.3 (429) †	0.823	87.1 (155) †	81.6 (40) †	89.1 (115) †	0.182	0.194
Competition	9.5 (57)	9.0 (11)	9.7 (46)	0.823	12.9 (23)	18.4 (9)	10.9 (14)	0.182	0.194
Type									
New injury	73.9 (441) †	82.0 (100) †	71.8 (341) †	0.022 #	67.4 (120) †	61.2 (30) †	69.8 (90) †	0.277	0.091
Recurrent	26.1 (156)	18.0 (22)	28.2 (134)	0.022 #	32.6 (58)	38.8 (19)	30.2 (39)	0.277	0.091
Mode									
Traumatism	8.9 (53)	9.8 (12)	8.6 (41)	0.677	16.9 (30)	6.1 (3)	20.9 (27)	0.018 #	0.003 *
Overuse	91.1 (544) †	90.2 (110) †	91.4 (434) †	0.677	83.1 (148) †	93.9 (46) †	79.1 (102) †	0.018 #	0.003 *
Severity									
1–3 days	48.6 (290)	41.0 (50)	50.5 (240)	0.060	47.8 (85)	51.0 (25)	46.5 (60)	0.591	0.847
4–7 days	21.9 (131)	18.9 (23)	22.7 (108)	0.355	14.0 (25)	14.3 (7)	14.0 (18)	0.955	0.021 *
8–21 days	17.3 (103)	24.6 (30)	15.4 (73)	0.016 #	18.5 (33)	18.4 (9)	18.6 (24)	0.971	0.692
>21 days	12.2 (73)	15.6 (19)	11.4 (54)	0.206	19.7 (35)	16.3 (8)	20.9 (27)	0.490	0.012 *
Rower position									
Stern	20.9 (125)	23.0 (28)	20.4 (97)	0.540	21.3 (38)	16.3 (8)	23.3 (30)	0.314	0.906
Middle	33.7 (201) †	35.2 (43)	33.3 (158) †	0.679	34.3 (61)	40.8 (20)	31.8 (41)	0.257	0.882
Bow	20.3 (121)	13.1 (16)	22.1 (105)	0.028 #	17.4 (31)	18.4 (9)	17.1 (22)	0.837	0.400
Versatile	25.1 (150)	28.7 (35)	24.2 (115)	0.309	27.0 (48)	24.5 (12)	27.9 (36)	0.646	0.621
Boat side									
Port	39.4 (235)	41.0 (50)	38.9 (185) †	0.681	43.8 (78) †	53.1 (26) †	40.3 (52)	0.126	0.288
Starboard	35.7 (213)	43.4 (53)	33.7 (160)	0.045 #	27.5 (49)	26.5 (13)	27.9 (36)	0.854	0.044 *
Both	25.0 (149) †	15.6 (19) †	27.4 (130)	0.007 #	28.7 (51)	20.4 (10)	31.8 (41)	0.134	0.323

Note: * Significant differences between male and female rowers ($p < 0.050$); # significant difference between elite and non-elite rowers ($p < 0.050$); † significant differences in each group ($p < 0.050$).

3.2. Body Region of Injuries

Table 2 shows the different body regions of injuries for men and women rowers at elite and non-elite competitive levels. Among all the body regions, lower back ($p < 0.01$), shoulder ($p < 0.05$), and ankle ($p < 0.01$) injuries showed statistical differences between the total sample of men and women rowers, while the rest of the body regions showed no such significance. When it comes to the differences between elite and non-elite rowers, back ($p < 0.05$), ribs ($p < 0.01$), and leg ($p < 0.05$) injuries were statistically different for men, and pyramidal injuries ($p < 0.05$) resulted in being higher for elite women rowers. Finally, the lower back was found to be the most significantly recurrent injury in male rowers for their total sample, as well as for the elite and non-elite groups.

Table 2. Body region of injuries in elite and non-elite male and female rowers.

	Male Rowers				Female Rowers				
	All (*n* = 597)	**Elite** (*n* = 122)	**Non-Elite** (*n* = 475)	*p*	**All** (*n* = 178)	**Elite** (*n* = 49)	**Non-Elite** (*n* = 129)	*p*	*p*
	% (*n*)	% (*n*)	% (*n*)		% (*n*)	% (*n*)	% (*n*)		
Lower back	35.2 (210) †	36.1 (440) †	34.9 (166) †	0.818	19.1 (340)	20.4 (100)	18.6 (240)	0.785	<0.001 *
Shoulder	11.2 (67)	10.7 (13)	11.4 (54)	0.824	16.9 (30)	22.4 (11)	14.7 (19)	0.219	0.046 *
Back	8.7 (52)	4.1 (5)	9.9 (47)	0.043 #	7.3 (13)	10.2 (5)	6.2 (8)	0.359	0.552
Knee	5.7 (34)	6.6 (8)	5.5 (26)	0.645	9.6 (17)	4.1 (2)	11.6 (15)	0.126	0.069
Ribs	5.4 (32)	12.3 (15)	3.6 (17)	<0.001 #	6.2 (11)	6.1 (3)	6.2 (8)	0.984	0.675
Pelvis	3.7 (22)	0.8 (1)	4.4 (21)	0.060	3.4 (6)	0.0 (0)	4.7 (6)	0.125	0.844
Forearm	3.5 (21)	4.1 (5)	3.4 (16)	0.696	3.9 (7)	6.1 (3)	3.1 (4)	0.354	0.795
Wrist	3.5 (21)	5.7 (7)	2.9 (14)	0.136	1.1 (2)	2.0 (1)	0.8 (1)	0.474	0.099
Neck	2.8 (17)	0.8 (1)	3.4 (16)	0.131	1.7 (3)	0.0 (0)	2.3 (3)	0.282	0.391
Leg	2.8 (17)	0.0 (0)	3.6 (17)	0.034 #	0.6 (1)	0.0 (0)	0.8 (1)	0.537	0.076
Elbow	2.3 (14)	3.3 (4)	2.1 (10)	0.445	2.2 (4)	2.0 (1)	2.3 (3)	0.909	0.939
Arm	2.2 (13)	4.1 (5)	1.7 (8)	0.103	4.5 (8)	6.1 (3)	3.9 (5)	0.518	0.095
Hip	2.2 (13)	1.6 (2)	2.3 (11)	0.648	2.8 (5)	2.0 (1)	3.1 (4)	0.702	0.623
Hand	2.0 (12)	3.3 (4)	1.7 (8)	0.263	1.7 (3)	0.0 (0)	2.3 (3)	0.282	0.783
Psoas	2.0 (12)	0.8 (1)	2.3 (11)	0.294	3.9 (7)	4.1 (2)	3.9 (5)	0.950	0.145
Thigh	1.7 (10)	0.0 (0)	2.1 (10)	0.106	1.7 (3)	2.0 (1)	1.6 (2)	0.820	0.992
Abdomen	1.2 (7)	0.8 (1)	1.3 (6)	0.685	2.8 (5)	2.0 (1)	3.1 (4)	0.702	0.121
Ankle	1.0 (6)	2.5 (3)	0.6 (3)	0.071	3.9 (7)	2.0 (1)	4.7 (6)	0.424	0.008 *
Foot	0.8 (5)	0.8 (1)	0.8 (4)	0.981	1.1 (2)	2.0 (1)	0.8 (1)	0.474	0.723
Abductor	0.7 (4)	0.8 (1)	0.6 (3)	0.820	0.6 (1)	0.0 (0)	0.8 (1)	0.537	0.874
Clavicle	0.5 (3)	0.8 (1)	0.4 (2)	0.579	0.6 (1)	0.0 (0)	0.0 (0)	-	0.923
Pyramidal	0.3 (2)	0.0 (0)	0.4 (2)	0.473	1.1 (2)	4.1 (2)	0.0 (0)	0.021 #	0.198
Fingers	0.2 (1)	0.0 (0)	0.2 (1)	0.612	2.2 (4)	0.0 (0)	3.1 (4)	0.213	0.002
Twin	0.2 (1)	0.0 (0)	0.2 (1)	0.612	0.6 (1)	0.0 (0)	0.8 (1)	0.537	0.363
Chest	0.2 (1)	0.0 (0)	0.2 (1)	0.612	0.6 (1)	2.0 (1)	0.8 (1)	0.474	0.363

Note: * Significant differences between male and female rowers ($p < 0.050$); # significant difference between elite and non-elite rowers ($p < 0.050$); † significant differences in each group ($p < 0.050$).

3.3. Diagnosis of Injuries

Table 3 shows the diagnosis of the suffered injuries, with muscular cramps ($p < 0.01$) and tendinitis ($p < 0.01$) being the highest diagnosed injuries for the total sample of men and women and being higher in female rowers, while sciatica ($p < 0.05$) was higher diagnosed for male rowers. In relation to the competitive level, significant differences were found in muscular cramps, Tendinitis, Sprain, fissure, and vertebral displacement for male's rowers. Muscle micro-tears were significantly higher in elite women than in non-elite women ($p < 0.05$). Finally, in the comparison of different diagnoses, muscle cramps and tendinitis were statistically significant and higher than the rest of the diagnoses for the total sample of men and women rowers ($p < 0.05$). Likewise, when it comes to the different groups, women and men who are non-elite rowers showed that muscle cramps and tendinitis were higher statistically ($p < 0.05$). On the other hand, in elite rowers, even though the number of diagnoses was also higher, only muscle cramps were found to show significant differences ($p < 0.05$). While this significance was not found for elite male rowers, a diagnosis of tendinitis was not significantly higher for both elite men and women.

Table 3. Diagnosis of injuries in elite and non-elite male and female rowers.

	Male Rowers				Female Rowers				
	All (*n* = 597)	**Elite (*n* = 122)**	**Non-Elite (*n* = 475)**	***p***	**All (*n* = 178)**	**Elite (*n* = 49)**	**Non-Elite (*n* = 129)**	***p***	***p***
	% (*n*)	**% (*n*)**	**% (*n*)**		**% (*n*)**	**% (*n*)**	**% (*n*)**		
Muscle Cramp	48.2 (288) †	32.8 (40)	52.2 (248) †	<0.001 #	73.0 (41) †	51.0 (25) †	37.2 (48) †	0.094	<0.001 *
Tendinitis	19.1 (114) †	27.9 (34)	16.8 (80) †	0.006 #	30.0 (17) †	14.3 (7)	17.8 (23) †	0.573	0.003 *
Muscle Tear	5.5 (33)	4.9 (6)	5.7 (27)	0.741	10.0 (6)	4.1 (2)	6.2 (8)	0.583	0.202
Muscle micro-tears	3.7 (22)	5.7 (7)	3.2 (15)	0.177	4.0 (2)	6.1 (3)	0.8 (1)	0.032 #	0.103
Overload	3.2 (19)	3.3 (4)	3.2 (15)	0.946	11.0 (6)	2.0 (1)	7.8 (10)	0.158	0.849
Sprain	3.0 (18)	5.7 (7)	2.3 (11)	0.049 #	5.0 (3)	0.0 (0)	3.9 (5)	0.162	0.298
Sciatica	2.5 (15)	2.5 (3)	2.5 (12)	0.966	0.0 (0)	0.0 (0)	0.0 (0)	-	0.033 *
Fracture	1.8 (11)	0.8 (1)	2.1 (10)	0.346	4.0 (2)	0.0 (0)	3.1 (4)	0.213	0.600
Contusion	1.3 (8)	1.6 (2)	1.3 (6)	0.747	3.0 (2)	4.1 (2)	0.8 (1)	0.126	0.679
Fissure	1.3 (8)	4.1 (5)	0.6 (3)	0.003 #	5.0 (3)	2.0 (1)	3.1 (4)	0.702	0.812
Herniated disc	1.3 (8)	1.6 (2)	1.3 (6)	0.747	1.0 (1)	0.0 (0)	0.8 (1)	0.537	0.251
Displacement	1.2 (7)	0.0 (0)	1.5 (7)	0.177	7.0 (4)	2.0 (1)	4.7 (6)	0.424	0.303
Dislocation	1.2 (7)	1.6 (2)	1.1 (5)	0.591	7.0 (4)	6.1 (3)	3.1 (4)	0.354	0.303
Vertebral displacement	1.0 (6)	3.3 (4)	0.4 (2)	0.005 #	0.0 (0)	0.0 (0)	0.0 (0)	-	0.179
Inflammation	0.7 (4)	0.8 (1)	0.6 (3)	0.820	1.0 (1)	0.0 (0)	0.8 (1)	0.537	0.587
Meniscus wear	0.5 (3)	0.0 (0)	0.6 (3)	0.379	2.0 (1)	2.0 (1)	0.8 (1)	0.474	0.836
Superficial injury	0.5 (3)	0.8 (1)	0.4 (2)	0.579	3.0 (2)	0.0 (0)	2.3 (3)	0.282	0.502
Protrusion	0.5 (3)	0.8 (1)	0.4 (2)	0.579	0.0 (0)	0.0 (0)	0.0 (0)	-	0.343
Joint impingement	0.3 (2)	0.0 (0)	0.4 (2)	0.473	3.0 (2)	0.0 (0)	2.3 (3)	0.282	0.298
Bursitis	0.3 (2)	0.8 (1)	0.2 (1)	0.299	1.0 (1)	0.0 (0)	0.8 (1)	0.537	0.968
Irritation	0.3 (2)	0.0 (0)	0.4 (2)	0.473	2.0 (1)	0.0 (0)	1.6 (2)	0.381	0.584
Low back pain	0.3 (2)	0.0 (0)	0.4 (2)	0.473	0.0 (0)	0.0 (0)	0.0 (0)	-	0.439
Meniscus tear	0.0 (0)	0.0 (0)	0.0 (0)	-	2.0 (1)	2.0 (1)	0.8 (1)	0.474	0.052
Femoroacetabular impingement	0.2 (1)	0.0 (0)	0.2 (1)	0.612	0.0 (0)	0.0 (0)	0.0 (0)	-	0.585
Muscle strain	0.2 (1)	0.0 (0)	0.2 (1)	0.612	0.0 (0)	0.0 (0)	0.0 (0)	-	0.585
Spondylitis	0.2 (1)	0.0 (0)	0.2 (1)	0.612	1.0 (1)	0.0 (0)	0.8 (1)	0.537	0.699
Plantar fasciitis	0.2 (1)	0.0 (0)	0.2 (1)	0.612	0.0 (0)	0.0 (0)	0.0 (0)	-	0.585
Pilonidal fistula	0.2 (1)	0.0 (0)	0.2 (1)	0.612	0.0 (0)	0.0 (0)	0.0 (0)	-	0.585
Tendon impingement	0.2 (1)	0.0 (0)	0.2 (1)	0.612	0.0 (0)	0.0 (0)	0.0 (0)	-	0.585
Burns	0.2 (1)	0.0 (0)	0.2 (1)	0.612	0.0 (0)	0.0 (0)	0.0 (0)	-	0.585
Rheumatism	0.2 (1)	0.0 (0)	0.2 (1)	0.612	0.0 (0)	0.0 (0)	0.0 (0)	-	0.585
Sacralgia	0.2 (1)	0.8 (1)	0.0 (0)	0.048	0.0 (0)	0.0 (0)	0.0 (0)	-	0.585
Calcifying myositis	0.0 (0)	0.0 (0)	0.0 (0)	-	1.0 (1)	2.0 (1)	0.0 (0)	0.104	0.170
Other	0.5 (3)	0.0 (0)	0.6 (3)	0.379	2.0 (1)	2.0 (1)	0.8 (1)	0.474	0.836

Note: * Significant differences between male and female rowers ($p < 0.050$); # significant difference between elite and non-elite rowers ($p < 0.050$); † significant differences in each group ($p < 0.050$).

4. Discussion

The main objective of this study was to analyze and describe the incidence of injury patterns in Traineras rowing. The main finding of this research was that the majority of injuries occurred during high-volume training periods. Most of them are new injuries for men and women. The most injured body regions in this rowing modality were the low back and shoulders, while the main diagnoses were muscle cramps and tendinitis for most of the rowers.

To our knowledge, investigations carried out on fixed-seat rowing injuries are practically non-existent in the scientific literature about Traineras. However, due to their similarity, research about Traineras has been compared to the Olympic rowing modality [1–3,5]. Therefore, despite the substantial heterogeneity of injury-reported methodologies in Olympic rowing research [57], our results will be compared to the existing literature related to injuries in Olympic rowing and also to the existing literature about injuries in other fixed-seat rowing modalities.

The results obtained showed that, irrespective of competitive level and gender, overuse injuries were much higher than those caused by trauma ($p < 0.01$) in Traineras rowing, results that are in line with previous research that described injuries in Olympic rowing in general [56]. In elite rowers [50,63] junior, senior, and master rowers [62], and female rowers [57]. Likewise, the result of this study fits well with Mediterranean fixed-sit rowing injuries [85]. These results make sense since, given the cyclic character of these rowing modalities and not physical contact, most injuries are the result of chronic overuse [56]. Within injury characteristics, our results showed that training periods and high-volume periods were statistically associated with a higher incidence of injury, in accordance with a previous investigation [85], and in the same line, new injuries were significantly higher than recurrent injuries.

The body region has been the most analyzed variable when trying to understand the prevalence of injuries in sports in general and also in rowing. The results described that the main injured body region was the low back, a similar result with previous investigations in Olympic rowing [64,66,69,71] and with an investigation carried out in Maltese traditional fixed-sit rowing [84]. However, the results were only partially in line with previous research carried out in Spanish Mediterranean rowers since the main injured body region was the ankle due to sprains over low back pain injuries, which were in second position [85]. The results showed that the second injured body region in Traineras rowing were the shoulders, only partially in line with results in previous investigations in Olympic rowing, since the following injured regions after the low back were the knee [51] and chest [49,60]. Related to rib stress fractures [55,58,67,82], and then shoulders were described in less number. In fact, the shoulder is considered a less common injury in Olympic rowing [60]. However, the results are in line with previous investigations in other fixed-seat rowing modalities, such as Mediterranean rowing [85] and Maltese traditional rowing [84], since shoulders were described as the most injured body regions. From these results, it could be interpreted that the fact that the flexion-extension of the trunk is more pronounced than in Olympic rowing and trunk rotation is needed significantly in the last part of the stroke influences the injury to the shoulders.

Within injuries per body region, the results showed that there were statistical differences between men and women in low back ($p < 0.01$). showing a higher incidence for male rowers. in the total sample and in both. elite and non-elite rowers. Our results showed that injuries in the low back were about 35% of the total for men, while in women, being the highest injured body region as well, they did not reach 20%. These results are in line with previous investigations in Olympic rowing, since the low back was the body region that showed the most injuries in elite [50,63], subelite [51,70], and international master rowers [51]. On the other hand, our results showed that the second body region with the most injuries was the shoulder, being statistically higher for women than men ($p < 0.05$). While almost 17% of the injuries were in the shoulders for women, in men, this body region collected 11.2% of the total injuries. Previous literature showed differences among men and women in Olympic rowing., e.g., comparing lightweight and open-weight rowers, and have demonstrated conflicting results [59] but no investigation in fixed-seat rowing have shown such a comparation.

Lastly, when it comes to the diagnosis of injuries, it should be noted the importance of the right diagnosis since it helps with better treatment and prevention [87]. Our results showed that muscle cramps were the most commonly diagnosed in Traineras rowing, showing statistical differences between men and women ($p < 0.01$), which was not in line with previous investigations in Olympic rowing [54]. On the other hand, our results showed that the second-ranked diagnosis for men and women rowers was tendinitis, being statistically higher in women and according to competitive level in men ($p < 0.05$). These results partially coincide with previous research in senior international Olympic rowers [50], since tendinitis was one of the main diagnoses of injury [62]. Other than that, our results are partially in line with the previous research about Mediterranean traditional rowing injuries, since tendinitis was one of the main diagnoses, after sprains and fractures [85]. Our results showed that sciatica diagnosis was statistically higher for men Traineras rowers than for women ($p < 0.05$), despite the low number of this type of diagnosis that were found. However, these results should be treated with caution since the heterogeneity of injury reported methodologies in rowing sport research is a fact [57], and further investigation would be needed taking into account the previously used methodologies.

Based on this first approach, the main injuries, their characteristics, and affected body regions are described, making this useful information for athletes, coaches, and physiotherapists. In addition, different professionals are involved in the performance of this sport. Future studies are needed to better understand the incidence of injuries in Traineras rowing, e.g., the training and biomechanical aspects that influence those injuries.

5. Conclusions

In conclusion, this study shows that the low back is the main body region affected by an injury in this sport, and shoulders are highly injured, finding differences between men and women. Moreover, most of the injuries are due to overuse and related to muscle cramps and tendinitis, partially in line with previous investigations carried out in different rowing modalities. This information will be helpful in creating a specific injury prevention plan to help athletes and coaches decrease the incidence of injuries, therefore helping them to improve continuity and reach their highest performance.

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Smartphone-Based Video Analysis for Guiding Shoulder Therapeutic Exercises: Concurrent Validity for Movement Quality Control

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Abstract: Neuromuscular re-education through therapeutic exercise has a determinant role in chronic shoulder pain rehabilitation. Smartphones are an interesting strategy to extend the rehabilitation to a home-based scenario as it can increase the attraction and involvement of users by providing feedback. Objective: To analyze the concurrent validity of a smartphone's application based on 2D video analysis against the gold-standard 3D optoelectronic system for assessing movement quality during upper limb therapeutic exercises. Methods: Fifteen young adults were evaluated while executing two different shoulder exercises with a smartphone's 2D video and a 3D optoelectronic system simultaneously in two conditions: (1) with the supervision and instructions of a physiotherapist (guided exercise), and (2) without the feedback of the physiotherapist (non-guided exercise). The data obtained during the guided and non-guided exercises were compared to calculate the movement quality index based on the approximation of the non-guided exercise to the guided exercise for the head, trunk, and shoulder's range of movement. The agreement of the movement quality index assessed with the smartphone application and the optoelectronic system was carried out through Bland–Altman analysis. Results: The Bland–Altman analysis indicates the range of agreement and bias tendency. This tendency demonstrates that the percentage of difference between the two methods increases as the movement quality index decreases. Conclusions: There is agreement between the movement quality evaluated by a gold-standard method and the developed application, although the proposed method appears to have less sensitivity for evaluating movements with lower quality index.

Keywords: shoulder rehabilitation; kinematic parameters; concurrent validity; 2D video system; smartphones; therapeutic exercises

1. Introduction

Chronic shoulder pain is the third most common pain condition presented in primary health care [1–3]. Exercise-based physiotherapy is the first line of approach [3–5]. In the clinical context, kinematic re-education in ensured through the supervision of qualified physiotherapists in promoting adequate muscle recruitment and movement patterns during the therapeutic exercise [6]. Continuous monitoring by a specialized professional is essential for greater health gains, but this monitorization may be limited when the rehabilitation process is extended to home. Recent studies have highlighted both the advantages and challenges associated with home-based rehabilitation [4,7,8]. Home-based rehabilitation offers logistical benefits, optimizing resources, reducing travel time, and providing flexibility in appointment scheduling [9]. This rehabilitation modality has shown potential in enhancing users' performance in daily activities, improving functional capacity, and enhancing overall quality of life. The home environment, rich in context-dependent learning opportunities and the use of familiar objects, increases the likelihood of transferring acquired skills to daily living activities [10]. However, the effectiveness of home-based rehabilitation may be influenced by several factors. The absence of direct clinical oversight, despite efforts to include continuous monitoring, pain education, and feedback in the home setting, could impact the overall success of the intervention. The adherence to prescribed home-based exercise programs has been posed as one of the main challenges in home-based rehabilitation. The non-monitorization of critical parameters such as frequency, intensity, sets, repetitions, rest time, and exercise speed can contribute to this fact [4,11–13]. Therefore, further research to optimize home-based implementation is required.

To address these limitations, among the various proposed solutions, one that shows high potential is the use of smartphones. Smartphones could be an interesting strategy as they can increase the attraction and involvement of users in the rehabilitation process. Given that it is a device that is easily used in daily life and available anywhere, it allows the bridging of gaps related to time, space, and costs and, more importantly, allows remote monitoring by the physiotherapist [4,10,14–16]. Several studies pointed to the use of smartphones to support rehabilitation as being very promising [17] since it would increase the effectiveness of exercise-based physiotherapy interventions in this setting and, thus, encourage health gains [4].

Two-dimensional (2D) systems that are incorporated in smartphone's cameras are simple to use, easily accessible, and affordable. However, their performance when compared to other systems, such as three-dimensional (3D) systems, is not yet well established [18]. Motion capture systems, widely used to quantify human movement, and 3D motion capture systems are considered the gold standard in human movement analysis in terms of accuracy and reliability [18]. However, 3D analysis methods are expensive and cannot be used in in home-based settings. On the other hand, the overall performance of 2D motion capture systems is not yet well established, which could justify the lack of broad utilization of these systems in both research and clinical contexts [18].

Given the aforementioned, there is a clear need to provide satisfactory evidence for the validity and reliability of smartphones' 2D video system as a tool to guide movement rehabilitation through movement quality control. Considering its portability, real-time data responsiveness, and the standardization of its usage, this smartphone application featuring a 2D camera system has the potential to offer benefits to the scientific community, patients, and rehabilitation professionals, ultimately enhancing treatment adherence and overall quality of life. Thus, the present study aims to validate a smartphone application which, through video recording monitors, supervises the execution of shoulder therapeutic exercises and gives the user feedback regarding the movement quality.

2. Materials and Methods

2.1. Study Design

This is a validation study, classified as observational, cross-sectional, and analytical.

2.2. Sample

The target population of this study included Center for Rehabilitation Research's employees aged between 18 and 35 years [19]. Musculoskeletal and neurological conditions that influence exercise performance, history of persistent pain associated with the shoulder complex, and extreme obesity (BMI greater than 40 kg/m^2) were exclusion criteria. Only young adults who consented to be contacted were assessed.

The final sample consisted of 15 participants. Ethical approval was obtained by the institutional Ethics Committee (CE0108C). All participants provided their written informed consent before the data collection began, according to the Declaration of Helsinki.

2.3. Instruments

2.3.1. Sample Selection and Characterization

A questionnaire via Google Forms was used to collect data to characterize the population and the criteria required for participation in the study. This questionnaire included topics related to demographic data (age, height, weight, dominant limb), general health (infectious, systemic, neurological, and/or musculoskeletal diseases) and shoulder pain (presence and frequency of shoulder pain). In addition, it featured a section for the participant to give their consent to be contacted to carry out the physical assessment and the study protocol.

The measuring tape of COMED® (COMED SAS, Strasbourg, France) has inelastic and flexible characteristics. It was used to measure the height (m) of the participants, being 200 cm in length and bearing graduation every 1 mm [20].

The TANITA scale, model BC-543 Inner Scan TM (Monitoring Your Health, Amsterdam, The Netherlands), was used to assess total body mass (kg) and body mass index (BMI) [21]. Its dimensions are $30 \times 30 \times 3$ cm^3, accounting for a mass of 2.52 kg. It has a maximum capacity of 150 kg and an accuracy of 0.1 kg per kg.

The International Physical Activity Questionnaire (IPAQ) was used to characterize the level of physical activity of the participants. This version was validated for the Portuguese population together with the coordinating group in Portugal, Mota, and Sardinha [22]. The questionnaire features a value referred to as the criterion validity considering the accelerometer data of $r = 0.49$ and a Cronbach α of 0.96 [22].

To assess the participants' potential interest in using the smartphone app developed, two research questions were made: "Would you find this application useful?" and "Would you see yourself using this application?". The participants were asked to answer "Yes" or "No".

2.3.2. Kinematic Data

The joint position of the shoulder, forearm, wrist, head, and trunk segments were assessed using an optoelectronic system, the Qualisys Motion Capture System (Qualisys AB, Göteborg, Sweden) [23], for concurrent validation. The spatial position of the reflector markers, placed on the participant, were collected using twelve infrared cameras, eight Oqus 500 and four Miqus M3, connected to the Qualisys USB Analog Acquisition interface, at a sampling frequency of 100 Hz. Qualisys Track Manager software version 2021.2 (Qualisys AB, Göteborg, Sweden) [23] was used to display, acquire, and analyze kinematic data.

The smartphone-based video analysis was conducted by a smartphone application, designed for Android devices, that leverages the power of machine learning to facilitate shoulder rehabilitation exercises. The app uses the device's 2D camera to capture video footage of patients performing their exercises. This footage is then processed in real-time using the MediaPipe library [24], a state-of-the-art machine learning solution for computer vision tasks, which recommends a minimum resolution of 640×480 pixels and 30 frames per second. Employing pose detection, a computer vision technique that identifies and tracks human body parts, the app analyzes users' movements during shoulder rehabilitation exercises. Despite the 2D nature of the camera, the app extracts valuable information such as shoulder position, arm angles, movement patterns, and gesture recognition from the detected poses. By analyzing the spatial relationships and temporal dynamics of key body landmarks, the application intelligently interprets the user's movements, offering a nuanced understanding of shoulder rehabilitation exercises beyond the limitations of a traditional 2D perspective. The key feature of the app is its ability to provide immediate feedback to the user. By analyzing the video footage, the app calculates a metric that reflects how similar the performance of the exercise is when non-guided compared to the exercise guided. This score is based on various factors such as the accuracy of the movements and the completion of the exercise.

MediaPipe is a powerful tool that uses machine learning to process video footage in real-time. One of the key features of MediaPipe is its ability to detect and track landmarks on the human body. These landmarks are specific points of interest that are identified in each frame of the video footage. When a patient performs an exercise, the app uses MediaPipe to identify these landmarks on the patient's body. Once the landmarks are identified, it tracks their movement across multiple (video) frames. This allows the app to analyze the patient's movements and calculate a score representing movement quality. For this study, landmarks from the shoulder, forearm, wrist, head, and trunk segment were considered.

2.4. Procedures

2.4.1. Sample Selection and Characterization

Data collection took place at a biomechanical laboratory, the Center for Rehabilitation Research of the School of Health at the Polytechnic Institute of Porto, between 5 and 20 August 2023, in a controlled environment. To avoid inter-rater error, each researcher was always responsible for the same task. Prior to data collection, anthropometric measures, body mass, height, and body mass index (BMI) were recorded for each participant. The participants maintained the orthostatic position on the scale, with bare feet and their upper limbs along the body, facing forward [25].

To identify the level of physical activity, participants were asked to complete the IPAQ.

2.4.2. Data Acquisition

The collections were carried out in one moment.

For the collection of kinematic data with Qualisys Motion Capture System, 32 reflective markers were placed bilaterally in the anatomical references (identified by manual palpation) needed to identify the movement of the assessed body segments: left and right anterior head (L/RALH), left and right posterior head (L/RPLH), left and right lateral part of the acromion (L/RCAJ), deepest point of incisura jugularis (SJN), xiphoid process (SXS), left and right styloid apophysis of the ulna (L/RULN) and radius (L/RRAD), left and right anterior superior iliac spine (L/RIAS), left and right lateral prominence of the greater trochanter (L/RFTC), spinous process of the seventh cervical vertebrae (CV7), second thoracic vertebrae (TV2), midpoint between the inferior angles of the most caudal points of the two scapulae (TV7), first lumbar vertebrae (LV1), fifth lumbar vertebrae (LV2), left and right posterior superior iliac spine (L/RIPS), left and right lateral (L/RLELB) and medial (L/RMELB) epicondyle of the humerus, left and right dorsal second metacarpal head (L/RLH), and left and right fifth metacarpal head (L/RMH) [26–30]. The marker setup is presented in Tables 1 and 2.

After placing the reflective markers, participants were instructed on the execution of two therapeutic exercises, namely, a diagonal shoulder exercise (D1) [31] and a multi-joint exercise, including shoulder external rotation at 90° of shoulder abduction (M90) [32] by a specialized physiotherapist. A description of the exercises used are shown in Table 3. Before data collection, sufficient time was given until the participants became familiar with the experimental setting.

Initially, each participant performed the exercises under the supervision of the physiotherapist to collect data from the movements that served as the basis for the comparative analysis—guided exercises (GE). The order in which the exercises were performed was randomized for each participant. After this collection, each participant watched a demonstration video on the application of the exercises to be performed. Subsequently, each exercise was performed unilaterally with the right upper limb three times, without supervision—non-guided exercises (NGE). A resting time of one minute between repetitions was established to prevent fatigue. While conducting the exercises, the optoelectronic system (considered the gold-standard equipment) recorded the 3D kinematics, and simultaneously, a 2D video was captured using a smartphone.

Table 1. Anterior view of marker setup.

Anterior View	
Marker Name	**Description**
L/RALH	Left/Right anterior head
L/RCAJ	Left/Right acromion
SJN	Deepest point of incisura jugularis
SXS	Xiphoid process, the most caudal point of the sternum
L/RIAS	Left/Right anterior superior iliac spine
L/RFTC	Most lateral prominence of the greater trochanter
L/RRAD	Left/Right distal radius
L/RULN	Left/Right distal ulna

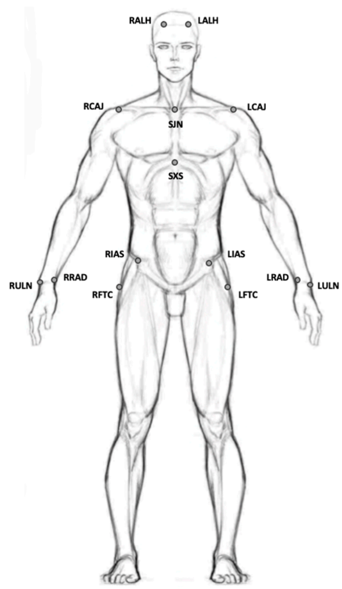

Table 2. Posterior view of marker setup.

Posterior View	
Marker Name	**Description**
L/RPLH	Left/Right posterior head
CV7	Spinous process of the seventh cervical vertebrae
TV2	Second thoracic vertebrae
TV7	Midpoint between the inferior angles of the most caudal points of the two scapulae
LV1	First lumbar vertebrae
LV3	Third lumbar vertebrae
LV5	Fifth lumbar vertebrae
L/RIPS	Left/Right posterior superior iliac spine
L/RLELB	Left/Right lateral elbow
L/RMELB	Left/Right medial elbow
L/RLH	Left/Right dorsal 2nd metacarpal head
L/RMH	Left/Right dorsal 5th metacarpal head

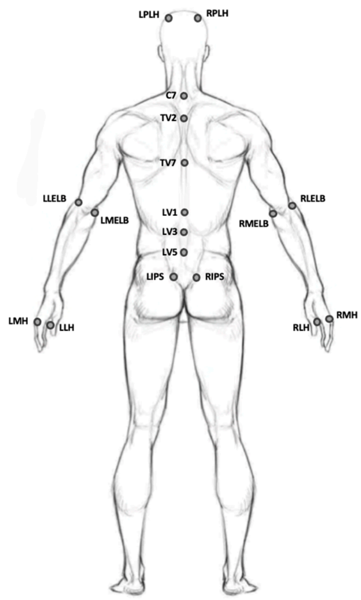

Table 3. Description of the diagonal shoulder exercise (D1) and multi-joint exercise, including shoulder external rotation at 90° of shoulder abduction (M90) used for the kinematic analysis.

Exercise	Initial Position	Final Position	Description
Diagonal Shoulder Exercise (D1)			The participant is standing, looking forwards with trunk and pelvis in a neutral position. The hand of the right upper limb should be at the level of the hip of the opposite lower limb and rotated inwards. The participant was asked to elevate the upper limb, both in the sagittal and frontal planes, and simultaneously rotate the right hand outwards. In the end, the patient should return to the starting position.
Multi-joint Exercise, Including Shoulder External Rotation at 90° of Shoulder Abduction (M90)			The participant is seated with knees bent at 90° flexion and feet on the floor, looking straight ahead and with the trunk and pelvis in a neutral position. The right elbow should be at 90° flexion, and the hand should be pointing forwards. The participant was asked to rotate the trunk to the right side and simultaneously rotate the right hand, producing the maximum shoulder external rotation. For 3 s, the participant was asked to bring the scapulas together, avoiding tilting the trunk. In the end, the patient should return to the starting position.

After performing all the proposed exercises, the researcher showed each participant a video demonstration of the application, exposing all the functionalities and its layout. Then, the potential interest of the participants in using the smartphone app was questioned.

To analyze the movement variation in Qualisys Motion Capture System, angles were calculated for each segment between two lines formed by anatomical markers. For the analysis of the head segment in both exercises, the markers left and right anterior head (L/RALH) and left and right acromion (L/RCAJ) were considered. Regarding the trunk segment, the markers left and right acromion (L/RCAJ) and left and right anterior superior spine (L/RIAS) were considered for both exercises. For the analysis of the shoulder segment, in the D1 exercise, the markers right acromion (RCAJ), right anterior superior spine (RIAS), and right lateral elbow (RLELB) were used. In the M90 exercise, we contemplated the right acromion (RCAJ), right lateral elbow (RLELB), and right distal ulna (RULN) markers for the analysis of the shoulder segment. Before and after calculating each angle, the "fit to 2nd degree curve" filter was applied.

For the smartphone app kinematic analysis, the recorded videos from guided exercises (GE) were directly compared to the non-guided exercises (NGE) for each participant. This analysis was performed (to all video frames) by calculating a similarity score between the landmarks from head, trunk, and shoulder segments of both videos. For the relevant landmarks, the distance between both videos was calculated using the cosine similarity. After comparing both videos, the average similarity was calculated, and if a predefined threshold was reached, that frame was considered similar. To calculate the final score representing movement quality, the similar number of frames was measured against the total, e.g., for a specific landmark, if we had an exercise video with 100 frames and 80 were deemed as similar, the final score would be 80%.

For the optoelectronic systems data, the final score was determined based on a comparison of the range of movement of guided exercises (GE) and non-guided exercises (NGE), which represents the percentage of approximation of the range of motion (ROM) obtained for each segment.

The percentage of approximation of the range of movement (ROM) obtained for each segment with the guided exercises against the non-guided exercises was designed as movement quality index and was assessed by using the following Formula (1):

$$Movement\ quality\ index\ (\%) = 1 - \frac{(ROM_{GE} - ROM_{NGE)}}{ROM_{GE}} \times 100 \quad (1)$$

After the collection and processing of the data provided by the instruments, they were exported to a Microsoft Excel for Microsoft 365 spreadsheet [33].

2.4.3. Statistical Analysis

For the statistical analysis, the software Predictive Analytics Software Statistics version 28 (SPSS IBM Corporation, Armonk, NY, USA) was used, with a significance level of 0.05 and a confidence interval of 95%.

Since the sample size was less than 30, the distribution of normality could not be assured. Thus, the Shapiro–Wilk test was used to test normality since the number of the sample was less than 50. As the variables did not follow a normal distribution, the median, the 25 and 75 percentiles, and the percentage values were used for the descriptive analysis.

Bland–Altman analysis was used to test the agreement between methods and to identify possible bias tendency. The percentage of the differences between the two methods were plotted against averaged values of the two methods. Separate Bland–Altman plots were created for head, trunk, and shoulder segments for both exercises. A linear regression analysis was calculated to quantify the bias tendency.

3. Results

3.1. Sociodemographic and Clinical Characterisation

As a result of the questionnaire distribution, 19 responses were obtained, of which 4 were excluded. Thus, the final sample consisted of 15 participants. The exclusion criteria are described in Figure 1.

After analyzing the quantitative variables of age and BMI, the median values (P25; P75) of each parameter were used to characterize the population. These values are described in Table 4.

Regarding the variable of gender, IPAQ and "Last shoulder pain episode" percentage values (%) of each parameter were used to characterize the population and are described in Table 5.

Also, regarding upper limb dominance, only one participant reported predominantly using the left upper limb; the other participants were right-handed. Only one participant reported having a diagnosed health condition, which in this case was diabetes mellitus 1.

Table 4. Characterization of the participants regarding age and BMI (Body Mass Index). Descriptive values of median (P25; P75) and *p*-values from the Shapiro–Wilk test are presented.

	Median (P25; P75)	**Valor *p***
Age (years)	25.00 (23; 27)	0.001
IMC (kg/m^2)	23.14 (22.04; 27.34)	0.004

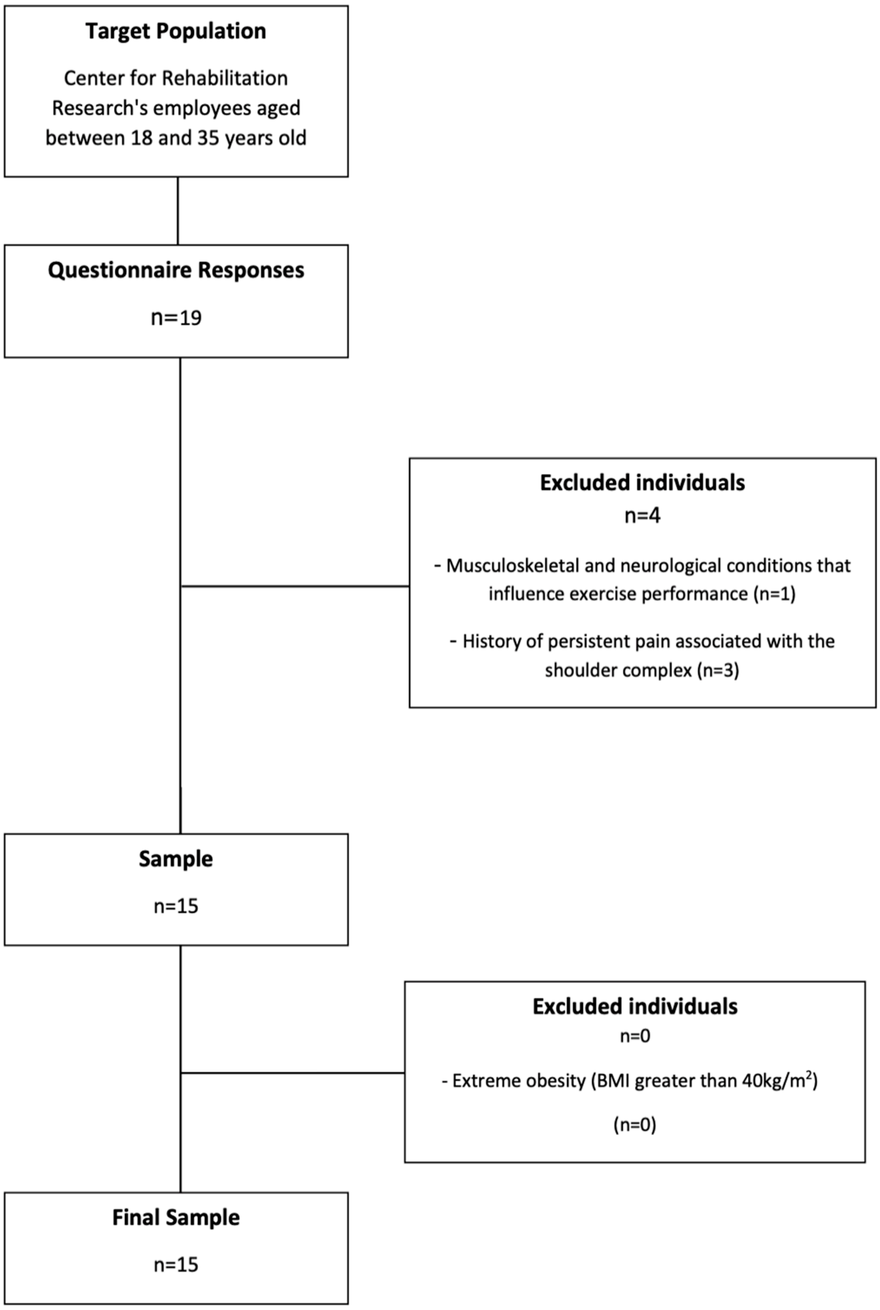

Figure 1. Sample constitution diagram.

Table 5. Characterization of the sample according to gender, the level of physical activity, and the variable "Last shoulder pain episode". Descriptive percentage values are presented below.

Gender	Women	60%
	Men	40%
Physical Activity Level	Low	20%
	Moderate	26.7%
	High	53.3%

Table 5. *Cont.*

"Last Time You Had Shoulder Pain"	Never	66.6%
	More than 6 months ago	26.7%
	Less than 6 months ago	6.7%

3.2. Concurrent Validity

As depicted in Figure 2 and Table 6, the extent of agreement varies according to the exercise and the specific body segment under consideration. To illustrate, when focusing on the head segment during the D1 exercise, a range of limits of agreement spanning from −9.63 to 71.40 can be observed, and an even wider range from −48.94 to 106.8 during the M90 exercise (Figure 2, Table 6). Bland–Altman analysis of this segment in both exercises revealed the presence of bias, which is corroborated by a significant correlation obtained from the regression analysis ($p < 0.05$). This correlation indicates that the difference (%) between the two methods increases as the movement quality index decreases. Turning our attention to the trunk segment, a more limited variation in the range of limits of agreement can be noted with an evident bias (Figure 2). It is noteworthy, however, that the average values of the movement quality index in the trunk segment were consistently high across both exercises. In the shoulder segment during the D1 exercise, a narrower range of limits of agreement and a low bias value can be observed (Figure 2). Conversely, in the M90 exercise within the same segment, wider limits of agreement and higher bias values can be noted (Figure 2). Notably, no discernible bias trend was found in this segment, as evidenced by the absence of a significant correlation ($p > 0.05$), which sets it apart from the other segments.

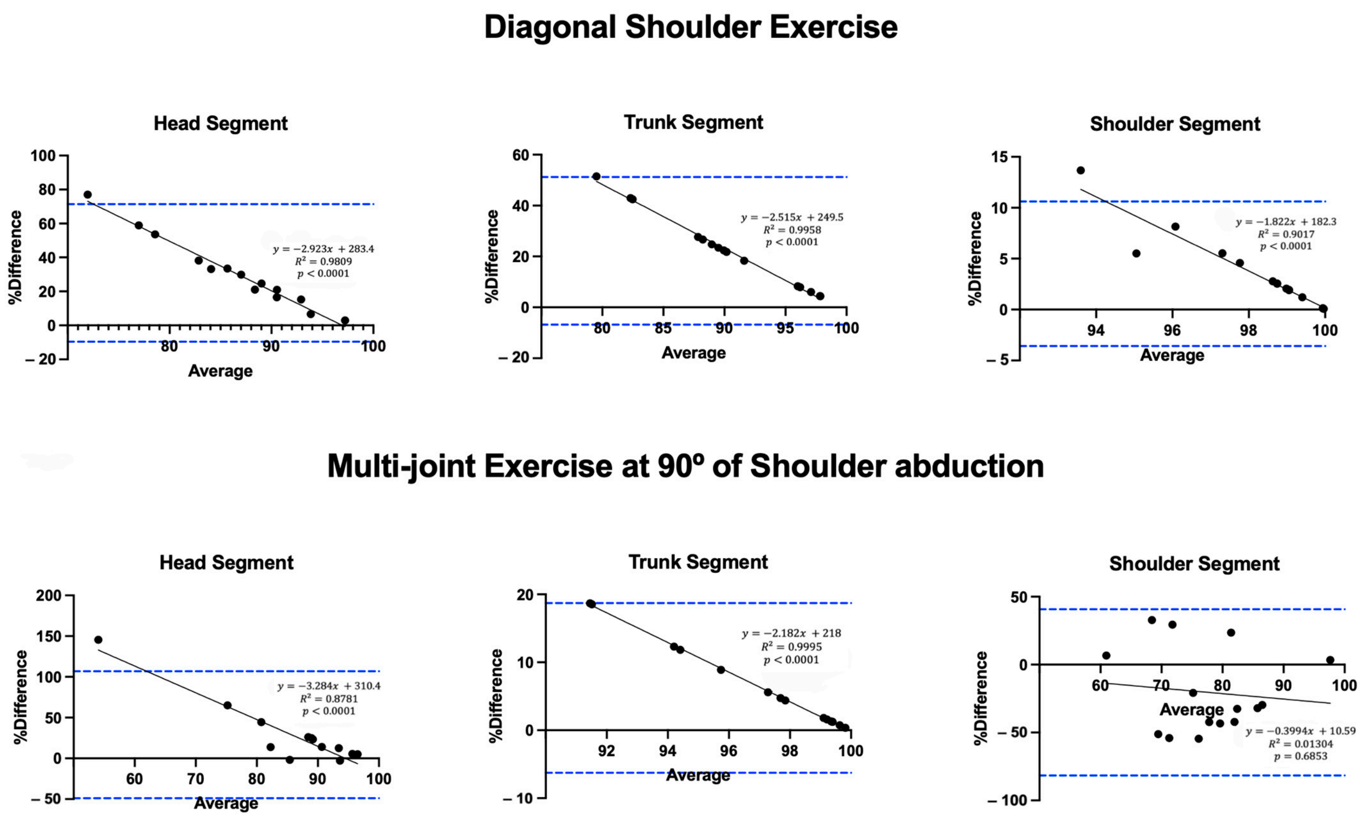

Figure 2. Bland–Altman analysis and linear regression of diagonal shoulder exercise (D1) and multi-joint exercise at 90° of shoulder abduction (M90). The blue lines indicate the upper and lower limits of agreement (LoA).

Table 6. Data from Bland–Altman analysis for all kinematic variables.

Segments	Bias		SD of Bias		LoA	
	D1	**M90**	**D1**	**M90**	**D1**	**M90**
Head	30.88	28.95	20.67	39.74	[−9.63; 71.40]	[−48.94; 106.8]
Trunk	22.20	6.22	14.80	6.37	[−6.80; 51.21]	[−6.27; 18.71]
Shoulder	3.51	−20.46	3.62	31.23	[−3.58; 10.62]	[−81.67; 40.74]

Standard deviation (SD); limits of agreement (LoA).

4. Discussion

The purpose of this study was to assess the concurrent validity of a smartphone application employing a 2D video system for supervising shoulder complex exercises in comparison to a gold-standard optoelectronic 3D system. Our hypothesis centered on the feasibility of employing an affordable and portable system as a kinematic tool with reasonable accuracy for guiding home-based rehabilitation exercises when compared to the gold standard.

Our findings indicate that agreement varies depending on the body segment and exercise in question. Starting with the analysis of the HEAD segment, we observed a wide range of limits of agreement in both exercises, representing significant variation in the percentage difference between the two methods. The Bland–Altman analysis also reveals a consistent bias trend. As the movement quality index of each subject decreased, the percentage difference between the two methods increased. However, upon closer examination of the Bland–Altman plot for the HEAD segment during the M90 exercise, it becomes evident that an agreement between methods exists when the movement quality index exceeds 90%, as evidenced by mean differences approaching zero [34]. This result may suggest that there is an alignment between the application and the 3D system when participants perform the guided and non-guided exercises almost identically. Future studies should assess the application in populations more prone to compensations, such as those with associated symptoms, as our participants generally demonstrated good movement quality.

Regarding the trunk segment, which plays a pivotal role in both exercises, we observed narrower ranges of limits of agreement and bias values compared to the head segment. The bias trend was again evident in the Bland–Altman analysis and confirmed by the regression analysis. However, in the context of the D1 exercise and considering the analysis of the trunk segment, the average variation in the movement quality index ranged from 70% to 100%, and for the M90 exercise, it ranged from 90% to 100%. This finding suggests that, overall, participants displayed consistent performance in both guided and non-guided versions of the exercise. Consequently, the observed trend associated with the trunk segment in both exercises may have limited clinical significance, as the small variation in the average movement quality control (x-axis) indicated an optimal kinematic relationship between GE and NGE.

In the shoulder segment during exercise D1, a narrower range of limits of agreement was noted, indicating reduced disparities between the 2D and 3D kinematic analyses. This phenomenon may be linked to the lower bias values observed in this segment and exercise. Furthermore, in this exercise, the shoulder segment displayed a narrower range of variation, with the movement quality index varying between 90% and 100%. The diagonal movement performed by the shoulder complex is recognized as a functional motion commonly employed in numerous daily activities [31,35]. This may explain the participants' higher proficiency in executing this movement without direct supervision from a physiotherapist, closely mirroring their performance in the guided exercise. This suggests that the observed bias trend may not have clinical significance. However, in the M90 exercise, the shoulder segment, despite showing higher values of limits of agreement and bias compared to the same segment in the D1 exercise, did not exhibit the consistent bias observed in the other segments and showed no statistical differences between methods

($p > 0.05$). This result reveals that for the principal segment of this exercise—the shoulder—the application demonstrates kinematic agreement with the 3D system.

Globally, the results of the present study seem to suggest that the application could be a valuable tool for supporting home-based rehabilitation in individuals with extensive movement experience and postural control. Future studies are required involving populations with low physical activity levels, limited movement awareness, or even associated chronic pain to test the suitability of this system for these subjects [36].

Some limitations were identified in this study. The fact that the sample was recruited voluntarily led to the existence of selection bias, resulting in a small sample size while hindering the representativeness of the population and decreasing the statistical power, thus compromising the external validity of this study. Regarding the questionnaires used to characterize the sample, all were subject to memory bias since some questions referred to past events.

For future studies, we suggest the recruitment of a larger sample and it would also be relevant to use more stringent inclusion and exclusion criteria to control confounding variables and allow data reproducibility. Given postural variability, it would be interesting to include specific populations (e.g., those with shoulder pain and those without), different tasks, and data collection protocols [18].

To our knowledge, very few studies have validated 2D video systems integrated into smartphones against 3D optoelectronic systems. A study conducted by Ramkumar et al. (2018) [37] aimed to validate a mobile technology for assessing shoulder range of motion. The study involved a comparison between a motion-based machine learning software and a goniometer for only four specific arcs of shoulder motion. While the study showed promising levels of accuracy, it is important to note that it did not employ gold-standard equipment, such as a 3D system. This methodology could potentially have an impact on the study's findings and results. Most existing research, as indicated by several systematic reviews [38,39] primarily examines tasks such as running or mechanical lifting, rather than those directly related to upper limb functionality. This underscores the innovative nature of our study, and it can represent a starting point for further research into this matter.

5. Conclusions

The smartphone application designed for supervising home-based rehabilitation exercises may be a valuable tool for guiding therapeutic exercise in specific populations, especially those with substantial movement experience and a heightened sense of body awareness and postural control. The results of the present study demonstrate an agreement between the movement quality evaluated by a gold-standard method and the developed application, although the proposed method appears to have less sensitivity for evaluating movements with lower quality index. However, considering that high levels of movement quality index were obtained in the present study, future studies involving shoulder pain conditions with lower levels of movement quality control are required.

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Institutional Review Board Statement: The study was conducted in accordance with the Declaration of Helsinki and approved by the Ethics Committee of the School of Health of the Polytechnic Institute of Porto (protocol code CE0108C and date of approval 15 March 2023) for studies involving humans.

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: Data are contained within the article.

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RObotic-Assisted Rehabilitation of Lower Limbs for Orthopedic Patients (ROAR-O): A Randomized Controlled Trial

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Abstract: Osteoarthritis is a common chronic condition in the elderly population and, with falls, represents a major public health problem. Patients with hip or knee osteoarthritis often have poor balance, which is considered an important risk factor for falls. In recent years, there has been increasing research supporting the use of robotic rehabilitation to improve function after total knee and hip replacement. The aim of this study is to investigate the effects of robotic balance rehabilitation on elderly patients who have undergone hip and knee replacement, with the aim of reducing the risk of falls and improving balance and walking, as well as motor function, fatigue, and overall quality of life. Twenty-four elderly patients with knee or hip replacement underwent robotic balance treatment with the Hunova® platform or conventional treatment three times a week for four weeks. Patients underwent an assessment of balance, walking, autonomy, quality of life and fatigue. Patients who underwent rehabilitation with Hunova® showed an improvement in dynamic balance ($p = 0.0039$) and walking ($p = 0.001$) and a reduction in both motor ($p = 0.001$) and cognitive ($p = 0.05$) fatigue. The study found that specific treatment for balance disorders in these patients could improve balance and reduce the risk of falling.

Keywords: technological rehabilitation; balance; osteoarthritis; elderly

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1. Introduction

Osteoarthritis, often known as OA, is a chronic condition characterised by a degenerative process that causes the joint to lose more and more structural components. It causes the gradual wear and tear of the articular cartilage, which are the tissues that lines the ends of bones, which in turn causes reactive neoformation of bone tissue, resulting in joint restriction and discomfort [1]. Several studies have shown that more than thirty percent of adults over the age of sixty-five, i.e., approximately one in three, experience a fall on a yearly basis [2–4]. The risk of falling may even increase by more than 50 percent with increasing age or in the presence of specific pathologies associated with ageing; in fact, OA affects almost all people in their 70 s, with the incidence reaching its highest point between the ages of 75 and 79 [2,3].

Falls are a common cause of illness and mortality, and often compromise people's level of independence, which can lead to an early need for assistance. In 2022, it was estimated that 28.6 per cent of people over 65 years in Italy fall during a year; of these, 43 per cent fall more than once, and 60 per cent of falls occur at home [4,5]. Falls and osteoarthritis are both major issues that affect public health. Falls are the second leading cause of death from unintentional injuries worldwide, and an estimated 37.3 million people require medical care each year because of their injuries [6]. Fractures are a major cause of death and morbidity, as well as a strain on the socioeconomic system [7,8]. The elderly have an increased risk of suffering from hip and knee osteoarthritis [9], as well as hip and knee fractures [10].

Patients suffering from knee and hip OA often experience pain, muscle weakness, impaired joint proprioception, and poor balance, all of which are important risk factors for falls [11–13]. There is a correlation between fall risk and measurements in several domains, such as assessment of cognitive and physical functioning, gender, age, comorbidities, medication use, fear of falling, and environmental factors [14]. Age and gender are two aspects of a person's life that cannot be changed, and are among the elements that contribute to the likelihood of falling.

Muscle function, balance control, and gait quality are three of the most significant determinants of fall risk; however, effective fall prevention interventions can improve these factors [15,16]. Fractures, requiring hospitalisation, are a common consequence of falls, which also have an important negative effect on both motor and cognitive function. In particular, for elderly patients undergoing hip or knee replacement surgery, the rehabilitation procedure for recovering the sense of balance and ability to walk is of paramount importance for improving person's autonomy and independence. In addition, to have a good recovery after hip or knee replacement surgery, it is essential to engage in rehabilitation activities that improve walking ability and sense of balance while reducing the risk of falling. In the past, post-surgical therapies for knee replacement have emphasized 'conventional' treatment protocols. These protocols included activities to improve range of motion, stretching, and development of strength and endurance. Nevertheless, this rehabilitation strategy did not significantly improve either the patient's level of discomfort or limb function [17]. On the other hand, more recent research has included more targeted balance therapies, both bipodal and monopodal, with an emphasis on the fact that a better outcome was achieved by introducing rehabilitative activities that stress the sensory systems involved in balance [18].

The most effective form of exercise-based therapy is one that combines strength training with other types of exercise, particularly functional and balance activities. For an exercise routine to be effective in reducing an individual's number of stumbles and falls, it must be condensed in terms of both duration and intensity [19]. In addition, the use of perturbation-based therapies [20] and stepping interventions helps to prevent the occurrence of falls [21].

A higher level of intensity, objectivity, and standardisation of treatment procedures, as well as the measurement of results, are possible with technological and robotic rehabilitation [22–24]. In recent years, research supporting the use of robotic rehabilitation to improve function after total knee and hip replacement has increased [25–28]. In particular, the use of exoskeletal robots and end-effectors for gait recovery is gaining popularity [29,30].

Considering the literature on the topic [31–33], the purpose of the study is to investigate the effects of robotic balance rehabilitation on elderly patients who have undergone hip and knee replacements, with the aim of reducing the risk of falls and improving balance and walking, as well as motor function, fatigue, and overall quality of life.

2. Materials and Methods

This is an interventional, randomised, controlled pilot study. Patients from the Post-Acute Rehabilitation Unit of the Fondazione Policlinico Universitario A. Gemelli IRCCS in Rome were evaluated between September 2022 and June 2023.

2.1. Inclusion and Exclusion Criteria

Patients had to meet the following inclusion requirements to be included in the study: (i) ages ≥ 55; (ii) results of total or partial prosthetic hip or knee replacement surgery; (iii) latency from the acute event between 5 days and 3 months; (iv) cognitive ability such that they could follow simple instructions and understand physiotherapist instructions, as measured by the Token Test (score $\geq$ 26.5); (v) ability to walk without assistance or with minimal help; and (vi) understanding and signing of informed consent.

In contrast, patients with: (i) systemic, neurological, cardiac pathologies that made walking dangerous or caused motor deficits; (ii) oncological conditions; (iii) plantar ulcers; or (iv) partial or total amputation of foot segments were excluded.

Patients who met the participation requirements were randomly assigned to one of two groups: the experimental group, referred to as G-Hun, or the conventional group, referred to as G-Conv. In addition to the traditional treatment that was offered by the clinical practice, G-Hun patients participated in a specialised rehabilitation treatment for balance using the Hunova® robotic platform (Movendo Technology srl, Genoa, Italy). In contrast, G-Conv patients were only required to receive traditional treatment on a daily basis.

2.2. Clinical and Technological Assessment

At baseline (T0) and after 4 weeks of treatment (T1), all patients underwent clinical and instrumental assessment.

Clinical evaluation was performed using scales assessing motor performance and balance, walking, autonomy, quality of life and fatigue.

The Berg Balance Scale (BBS), the Time Up&Go (TUG), the Italian Knee Injury and Osteoarthritis Outcome Score (KOOS-I) and the Italian Hip disability and Osteoarthritis Outcome Score (HOOS-I), the Short Physical Performance Battery (SPPB) and the Motricity Index-Lower Limb (MI-LL) were used to assess motor performance and balance.

The BBS is an instrument used to assess a patient's ability or inability to maintain balance during a series of tasks, both static and dynamic. It is a scale consisting of 14 tasks, for each of which a value ranging from 0 to 4 can be assigned, where 0 indicates the inability to perform or complete the proposed task and 4 the highest level of functionality. A score below 45 indicates an increased risk of falling [34]. The TUG is a test to assess the risk of falling in the elderly. It is performed by recording the speed with which a patient is able to get up from a chair, walk a distance of 3 metres, turn around, return to the chair and sit down. In the elderly population, a time $\geq$ 12 indicates a risk of falling [35]. The HOOS and KOOS are two extremely similar self-administered questionnaires investigating symptoms, at the hip joint level in the former, and at the knee level in the latter. They consist of 40 and 42 items, respectively; both are divided into five subscales (symptoms, pain, activities of daily living, physical function, sports and recreational activities, and quality of life). All items can be scored from 0 (no difficulty) to 4 (high difficulty). For each subscale, the result is calculated as a percentage, where higher percentages correspond to a better condition [36,37]. The SPPB is a test that assesses balance, lower limb strength and functional capacity in the elderly. Three specific domains are assessed, i.e., the ability to stand for 10 s with feet in three different positions (side-by-side, semi-tandem and tandem); gait, which is assessed by recording the time taken by the patient to walk 3 or 4 metres (the fastest recorded); and the time taken by the patient to get up from a chair five times [38,39]. The MI-LL is a test used to assess motor impairment. Ankle dorsiflexion, knee extension, and hip flexion are assessed for each lower limb [40].

The walking assessment was carried out through the Ambulation Index (AI), the Walking Handicap Scale (WHS), the Functional Ambulation Classification (FAC), the 10-Meter Walking Test (10 MWT), and the 6-Minute Walking Test (6 MWT).

The AI is a mobility assessment scale that evaluates the time and degree of assistance needed to walk 8 metres. The patient is asked to walk 8 metres as quickly and safely as possible, while the travel time and the type of assistance needed are recorded. The travel time is used to assign a score to the patient: lower scores correspond to a higher degree of independence and activity [41]. The WHS is a scale used to assess the quality of walking in the home and social environment through six categories. Category 1 corresponds to 'walking only for exercise', while category 6 corresponds to 'walking in the social environment without limitations' [42]. The FAC is a functional gait test that assesses the patient's ability to walk and the amount of help needed. A higher score corresponds to the ability to walk independently [43]. The 10 mWT is a performance test that is used to assess how long it takes a patient to travel a predetermined distance [44]. The 6 MWT is a test that assesses walking endurance. The patient is asked to walk as far as possible in six minutes. The final score is given by the metres walked by the patient [45].

For the assessment of autonomy, quality of life, and fatigue, the modified Barthel Index (mBI), the EuroQoL-5D (EQ-5D) and the Modified Fatigue Impact Scale (MFIS) were used.

The mBI is a scale that measures autonomy in performing the activities of daily living (ADLs), such as personal hygiene, the ability to wash oneself, feed oneself, use the toilet, the ability to dress oneself, bladder and bowel control, the ability to make transfers, the use of a wheelchair, and the ability to walk and climb stairs. Higher scores correspond to greater autonomy in ADLs [46]. The EQ-5D is an instrument used to assess health-related quality of life. Overall, it assesses five dimensions: mobility, self-care, habitual activities, pain, and anxiety/depression. Higher scores correspond to a worse health status [47]. The MFIS is a scale that evaluates the influence that fatigue has on people's lives. The impact of weariness on a person's physical, cognitive, and psychosocial functioning can be evaluated with the help of this test. It consists of 21 items, each of which is assessed using a Likert scale with five points, ranging from 0 (meaning "never") to 4 (meaning "almost always"). In addition to the overall score, it is possible to extract scores for the physical subscale, the cognitive subscale, and the psychosocial subscale. Scores higher than 10 indicate increased levels of weariness [48,49].

Considering instrumental assessment, the balance assessment was performed with the use of the robotic platform (Hunova® Movendo Technology srl, Genoa, Italy). In particular, the balance assessment was performed in both static and dynamic conditions while the subject was standing. Specifically, stabilometric data were collected from the analysis of centre of pressure (CoP) trajectories. Subsequently, the following balance performance factors were calculated from the instantaneous CoP positions: CoP oscillation velocity along the antero-posterior (AP) and mid-lateral (ML) axis, CoP trajectory length, area of the 95% confidence ellipse, and the Romberg Test [ratio of the length value in the eyes closed (OC) condition to the same value in the eyes open (OA) condition]. In addition, trunk movement data were also collected.

2.3. Procedures

The Hunova® robotic platform is a medical device consisting of two electromechanical platforms, a foot platform and the seat platform [50].

G-Hun patients underwent robotic treatment for improving balance using Hunova® 3 times a week for 45 min each, in addition to conventional treatment. In particular, the technological rehabilitation performed with Hunova® mainly aimed to improve balance, both sitting and standing; static and dynamic exercises, dual-task exercises and exercises to improve trunk control were proposed. In particular, technological rehabilitation was mainly geared towards improving balance, both in a sitting and standing position.

As for the standing exercises, they were first performed in bipodal position. This position was adopted to train the patient to handle the load appropriately, to maintain the standing position both in quiet environments and during activities that require continuous adjustment of standing and trunk control, and to keep the load in the correct position. Subsequently, training was performed to strengthen the core and improve proprioception, both on a stationary platform and using different perturbation modes (elastic mode, fluid mode). As treatment progressed, exercises were performed in monopodal support, both on the operated limb and the healthy limb, to restore proper load management and proprioception, as well as to strengthen muscle tissue.

To strengthen the trunk and improve load control on the lower limbs and pelvis, exercises performed while seated focused on these areas.

To train the patient to complete several tasks simultaneously, dual-task activities were performed while sitting and standing. Cognitive–motor coordination and dual-task activities were performed.

G-Conv patients underwent only conventional rehabilitation treatment using major rehabilitation methods (e.g., neurocognitive theory, progressive neuromuscular facilitation, etc.).

2.4. Statistical Analysis

As this is a pilot study on a specific subgroup of patients, on whom the actual usefulness of the Hunova® has not yet been studied in the literature, the study was set up as a pilot study. As such, no formal sample sizing was necessary. However, based on Julious' rules (2005) of thumb for clinical pilot studies [51], 12 subjects per group were included for a total population of 24 subjects.

The division into the two groups followed a randomisation algorithm according to the random sorting procedure. The allocation sequence was generated through PASS2019 software.

The sample was described in its clinical and demographic variables using descriptive statistical techniques. Quantitative variables were summarized with mean and standard deviation (SD), and median and interquartile range (IQR) where appropriate. Qualitative variables were presented through absolute and percentage frequency tables.

The Shapiro–Wilk probability test was used to assess the normality of the distributions [52]. The within-group analysis was based on the application of the Wilcoxon Signed Rank test for each clinical and instrumental balance outcome registered at T0 and T1.

The between-group analysis was performed using The Mann–Whitney U test to compare the percentage increase calculated for each group.

Regarding the clinical outcome, the between-group differences were analyzed by comparing the changes from baseline of each clinical scale, defined as S(T1) − S(T0), where S is one of the clinical outcomes. Instead, regarding the instrumental outcome, the between-group differences were analyzed by comparing the percentage increase in each outcome, defined as $\Delta S = \frac{(S(T1)-S(T0))}{(S(T0))}$.

The between-group analysis of clinical scales were conducted by considering the differences between the scores, S(T1) − S(T0), because the minimum value of almost all scales is 0, and normalization was not thus possible.

Statistical significance for each test was set at 0.05. Statistical analysis was performed with SPSS 25 (IBM Corp., Armonk, NY, USA).

3. Results

Some 24 patients admitted to the post-acute rehabilitation unit between September 2022 and June 2023, 10 men and 14 women, with a mean age of 67.64 years (standard deviation of ±23.67 years), were included in the study. The two groups did not differ in terms of clinical and demographic characteristics, as shown in Table 1.

Table 1. Clinical and demographical characteristic of the whole sample at baseline.

		G-Hun (n = 12)	G-Conv (n = 12)	p Value
Gender, %	Male vs. Female	33.33 % vs. 66.67 %	50.00 % vs. 50.00 %	p = 0.410
Age, years	Mean ± SD	69.1 ± 24.6	65.6 ± 24.3	p = 0.887
Latency, days	Mean ± SD	5.0 ± 1.4	5.0 ± 1.6	p = 0.514
Type of prosthesis, %	Knee vs. Hip	58.33 % vs. 41.67 %	41.67 % vs. 58.33 %	p = 1.000
Affected size, %	Left vs. Right	66.67 % vs. 33.33 %	58.33 % vs. 41.67 %	p = 0.755

G-Hun: experimental group G-Conv: conventional group; SD: standard deviation.

Considering the motor, balance and walking assessments, the intragroup analysis showed statistically significant improvements in the clinical scales in both groups, as shown in Table 2. Regarding the intergroup comparison of clinical scales, however, a statistically significant difference was found in TUG (p = 0.039), 10 MWT (p = 0.001) and 6 MWT (p = 0.001) values (Figure 1).

Table 2. Intra-group and inter-group analysis of motor function, balance, gait and fatigue, autonomy and quality-of-life scales.

	G-Hun			G-Con			
	T0 Median (IQR)	**T1 Median (IQR)**	***p* Value**	**T0 Median (IQR)**	**T1 Median (IQR)**	***p* Value**	***p* Value G-Hun vs. G-Con**
Motor function, balance and gait							
MI-LL prosthetic side	61.5 (48.75–65.5)	77 (74.5–92)	***p* = 0.002**	64 (52.5–66)	87 (74.5–91.25)	***p* = 0.013**	*p* = 0.319
MI-LL non-prosthetic side	88 (82–100)	100 (92–100)	***p* = 0.016**	88 (76–94)	95.5 (91–100)	***p* = 0.017**	*p* = 0.799
TUG	31.60 (27.63–63.5)	22.93 (19.14–26.86)	***p* = 0.002**	23.85 (17.32–27.62)	12.82 (8.3–24.29)	***p* = 0.002**	***p* = 0.039**
BBS	30 (22.75–32.25)	42 (38.75–48.5)	***p* = 0.02**	28 (18–33)	38.5 (33–54)	***p* = 0.003**	*p* = 0.089
SPPB_B	1 (1–2)	2 (2–3)	***p* = 0.002**	1 (1–1)	2 (1.75–4)	***p* = 0.007**	*p* = 0.713
SPPB_W	1 (1–1)	2 (1–2)	***p* = 0.007**	1 (1–1)	1 (1–3)	***p* = 0.039**	*p* = 0.843
SPPB_STS	1 (1–1)	2 (1.75–2)	***p* = 0.004**	1 (1–1)	1 (1–4)	***p* = 0.007**	*p* = 0.843
SPPB_TOT	3 (2.75–4)	6 (5–6)	***p* = 0.002**	3 (3–3)	4 (3.75–11)	***p* = 0.011**	*p* = 0.514
HAI	5 (4.75–6)	3 (2.75–4)	***p* = 0.002**	4.5 (2–5.25)	2 (2–2.5)	***p* = 0.053**	*p* = 0.671
FAC	1.5 (1–2)	4 (3–4)	***p* = 0.002**	2 (2–3)	3 (2.75–5)	***p* = 0.002**	*p* = 0.219
WHS	2 (2–1)	3.5 (3–4)	***p* = 0.001**	2 (2–2)	3 (3–5)	***p* = 0.002**	*p* = 0.932
10 MWT	20.16 (16.86–28.61)	12.19 (11.29–18.58)	***p* = 0.001**	17.4 (14.8–18.24)	15.69 (13.79–16.38)	***p* = 0.002**	***p* = 0.001**
6 MWT	77.5 (60–106.25)	175 (133.75–212.5)	***p* = 0.002**	100 (85–1656)	150 (143.75–190)	***p* = 0.003**	***p* = 0.001**
Fatigue, autonomy, and quality of life							
mBI	50.5 (42.75–58.5)	91 (84–92.75)	***p* = 0.002**	43 (37.75–43)	64 (56–87)	***p* = 0.002**	*p* = 0.219
EQ-5D_VAS	52.5 (43.75–60)	85 (80–86.25)	***p* = 0.003**	47.5 (45–53.75)	80 (70–85)	***p* = 0.003**	*p* = 0.630
EQ-5D TOT	11 (10.5–15.25)	7 (6–7.25)	***p* = 0.002**	15 (14–15)	9 (7–11)	***p* = 0.002**	*p* = 0.713
MFIS_PHY	25 (22.5–28.25)	14 (10–16.5)	***p* = 0.002**	22 (21–22.25)	19 (16–19)	***p* = 0.002**	***p* = 0.001**
MFIS_COG	13 (12–18.25)	3 (2–3.25)	***p* = 0.003**	11 (9–16.75)	8 (6–12)	***p* = 0.002**	***p* = 0.05**
MFIS_PSY	7 (5.5–7.25)	3 (2–3.25)	***p* = 0.003**	5 (4.75–6.25)	3 (2–3)	***p* = 0.003**	*p* = 0.514
MFIS_TOT	43.5 (42–50.75)	23.5 (17.75–29.25)	***p* = 0.002**	41 (36–44)	30 (28–30)	***p* = 0.002**	***p* = 0.002**

G-Hun: experimental group G-Conv: conventional group; MI-LL: Motricity index-Lower Limb; TUG: Timed Up&Go; BBS: Berg Balance Scale; SPPB_B: Short Physical Performance Battery_Balance; SPPB_W: Short Physical Performance Battery_Walking; SPPB_STS: Short Physical Performance Battery_Sit To Stand; HAI: Hauser Ambulation index; FAC: Functional Ambulation Classification; WHS: Walking Handicap Scale; 10 MWT: 10-Meter Walking Test; 6 MWT: 6-Minute Walking Test; mBI: modified Barthel Index; EQ-5D: EuroQuoL-5 Dimensions; MFIS_PHY: Modified Fatigue impact Scale_Physical; MFIS_COG: Modified Fatigue impact Scale_Cognitive; MFIS_PSY: Modified Fatigue impact Scale_Psycosocial; in bold the significant results for $p < 0.05$.

Regarding the ratings of fatigue, autonomy and quality of life, the intra-group analysis showed significant improvements for all the rating scales used, for both G-Hun and G-Conv. Comparing the two groups, a statistically significant difference was found in MFIS (p = 0.002), particularly for the physical subscale (p = 0.001) and the cognitive subscale (p = 0.05).

Considering the subgroup of patients who underwent knee replacement, the intra-group analysis between those who underwent robotic balance rehabilitation and those who underwent conventional rehabilitation showed statistically significant differences in most of the subscales comprising the KOOS-I, with the exception of the subscale related to activities of daily living in the G-Hun (p = 0.080) and the G-Conv (p = 0.170) groups.

Otherwise, considering the subgroup of patients undergoing hip replacement, intra-group analysis showed statistically significant results in all subscales of the HOOS-I only in the G-Hun group (Table 3).

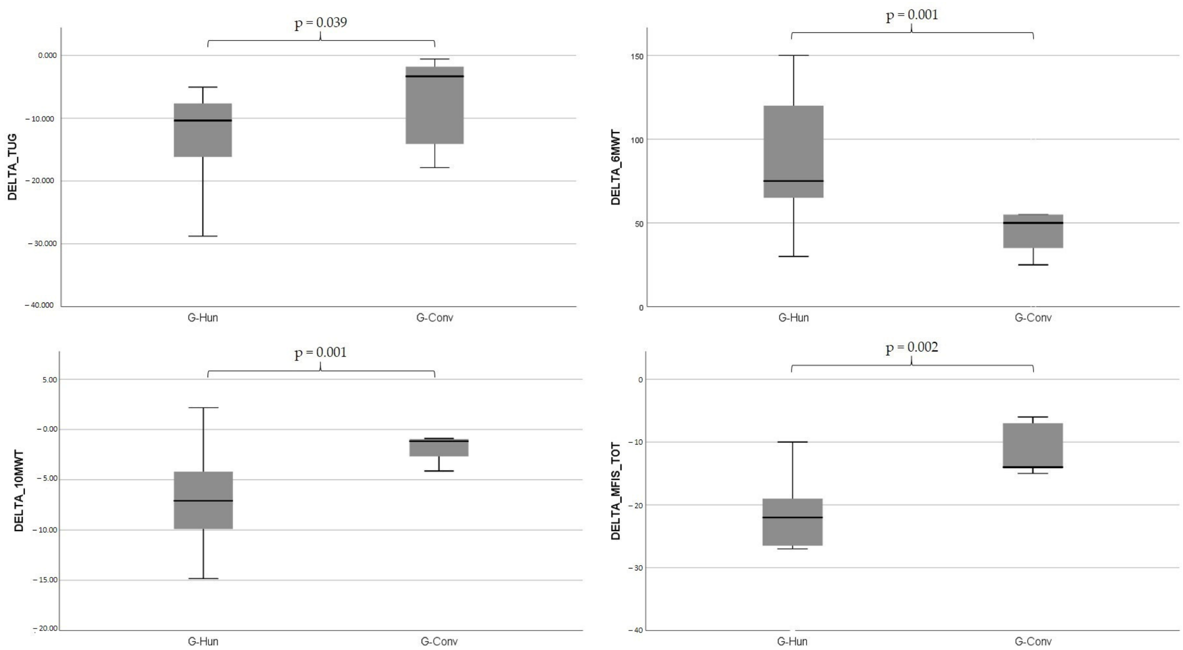

Figure 1. Comparison of dynamic balance, walking, and fatigue between G-Hun and G-Conv.

Table 3. Intra-group analysis of Italian version of Knee Injury and Osteoarthritis Outcome score and the Italian version of Hip Injury and Osteoarthritis Outcome score for both groups.

	G-Hun			G-Conv		
	T0 Median (IQR)	**T1 Median (IQR)**	***p* Value**	**T0 Median (IQR)**	**T1 Median (IQR)**	***p* Value**
KOOS-I						
KOOS_S	70 (65–75)	35 (25–45)	***p* = 0.013**	91.66 (91.66–100)	75 (75–75)	***p* = 0.046**
KOOS_P	69.45 (66.66–75)	41.67 (41.67–52.77)	***p* = 0.043**	80 (80–100)	33.34 (33.34 + 75)	***p* = 0.046**
KOOS_ADL	69.12 (67.64–79.42)	48.53 (45.49–52.95)	*p* = 0.080	70.58 (70.58–92.65)	51.18 (30.89–72.05)	*p* = 0.170
KOOS_sport	100 (100–100)	50 (52–75)	***p* = 0.024**	100 (100–100)	81.25 (45–81.25)	***p* = 0.042**
KOOS_QoL	75 (62.5–81.25)	27.5 (25–31.25)	***p* = 0.033**	75 (68.75–75)	62.25 (56.25–62.25)	***p* = 0.036**
KOOS_TOT	74.50 (74.18–78.21)	42.02 (31.45–45.10)	***p* = 0.013**	82.04 (82.04–93.57)	58.43 (45.25–70.34)	***p* = 0.016**
HOOS-I						
HOOS_S	66.67 (63.33–79.17)	33.33 (25–45.83)	***p* = 0.018**	75 (75–75)	66.67 (66.67–84.3)	*p* = 0.680
HOOS_*p*	69.95 (62.5–83.75)	52 (32.5–37.5)	***p* = 0.018**	57.5 (57.5–57.5)	52 (32.5–80)	*p* = 0.892
HOOS_ADL	70.59 (60.88–75.74)	43.75 (31.61–52.8)	***p* = 0.018**	60.3 (60.3–60.3)	54.42 (51.18–64.71)	*p* = 0.684
HOOS_sport	100 (100–100)	50 (46.88–62.5)	***p* = 0.017**	100 (43.75–81.2)	100 (100–100)	*p* = 0.180
HOOS_QoL	68.42 (65.25–75)	18.75 (11.13–31.25)	***p* = 0.018**	52 (43.75–81.2)	62.75 (62.75–62.75)	*p* = 0.684
HOOS_TOT	78.32 (70.70–80.09)	36.89 (31.93–47.63)	***p* = 0.018**	67.47 (54.70–70.67)	60.10 (58.43–72.60)	*p* = 0.684

G-Hun: experimental group G-Conv: conventional group; KOOS_S: KOOS_Symptoms; KOOS_P: KOOS_Pain; KOOS_ADL: KOOS_Activity of Daily Living; KOOS_QoL: KOOS_Quality of Life; HOOS_S: HOOS_Symptoms; HOOS_P: HOOS_Pain; HOOS_ADL: HOOS_Activity of Daily Living; HOOS_QoL: HOOS_Quality of Life; significant results at $p < 0.05$ are in bold.

On the other hand, regarding the instrumental assessment of static balance, the intra-group analysis showed no statistically significant results for either G-Hun or G-Conv. In contrast, the intergroup comparison showed statistically significant results for the Romberg index ($p = 0.047$) and the COP sway range in the mid-lateral direction with open eyes ($p = 0.026$).

Conversely, instrumental assessment of dynamic balance showed significance for all parameters considered for G-Hun, while G-Conv showed no statistically significant results. Comparison between groups showed significance for open-eye sway area ($p = 0.006$), open-eye COP path ($p = 0.002$), trunk movements ($p = 0.035$), trunk sway in the mid-lateral direction ($p = 0.001$), and mean speed of COP sway in the antero-posterior direction ($p = 0.035$) (Table 4).

Table 4. Intra-group and inter-group analysis of instrumental assessment of the static and dynamic conditions of the whole sample.

	G-Hun			G-Conv			
	T0 Mediana (IQR)	**T1 Mediana (IQR)**	***p* Value**	**T0 Mediana (IQR)**	**T1 Mediana (IQR)**	***p* Value**	***p* Value G-Huv vs. G-Conv**
Static Condition							
Area-EC [cm^2]	4.96 [4.17–6.50]	2.67 [1.87–6.08]	$p = 0.169$	7.73 [5.98–9.5]	7.73 [5.98–9.5]	$p = 0.142$	$p = 0.186$
Area-EO [cm^2]	2.50 [2.05–3.34]	1.99 [1.18–3.58]	$p = 0.959$	1.17 [0.95–3.1]	1.35 [1.17–3.1]	$p = 0.155$	$p = 0.361$
Romberg Index	0.37 [0.36–0.55]	0.86 [0.37–1.40]	$p = 0.471$	0.23 [0.14–0.33]	0.23 [0.14–0.47]	$p = 0.241$	$\mathbf{p = 0.047}$
COP path-EO [cm]	49.78 [36.02–70.64]	42.96 [31.00–61.78]	$p = 0.333$	33.53 [28.25–35.79]	35.32 [29.3–35.79]	$p = 0.177$	$p = 0.303$
COP path-EC [cm]	64.38 [49.44–118.26]	60.68 [49.56–88.22]	$p = 0.169$	62.89 [46.76–115.83]	62.89 [46.76–115.83]	$p = 0.220$	$p = 0.119$
Trunk movement-EO [deg/s^2]	0.05 [0.04–0.07]	0.05 [0.05–0.07]	$p = 0.735$	0.04 [0.04–0.06]	0.04 [0.04–0.08]	$p = 1.000$	$p = 0.569$
Trunk movement-EC [deg/s^2]	0.05 [0.05–0.12]	0.05 [0.05–0.05]	$p = 0.075$	0.06 [0.04–0.1]	0.06 [0.04–0.01]	$p = 0.237$	$p = 0.277$
Trunk sway range AP-EO [deg]	2.80 [2.12–3.22]	3.03 [2.31–6.97]	$p = 0.059$	2.12 [2.1–2.97]	2.12 [2.1–2.97]	$p = 0.256$	$p = 0.186$
Trunk sway range AP-EC [deg]	3.07 [2.42–4.16]	2.78 [2.24–3.32]	$p = 0.575$	3.21 [2.52–4.09]	3.21 [2.52–4.09]	$p = 0.242$	$p = 0.608$
Trunk sway range ML-EO [deg]	1.25 [0.83–1.54]	1.28 [0.97–2.07]	$p = 0.507$	0.81 [0.43.1.17]	0.81 [0.45–1.17]	$p = 0.158$	$p = 0.691$
Trunk sway range ML-EC [deg]	1.41 [1.26–1.90]	1.31 [0.98–2.03]	$p = 0.507$	1.21 [0.86–1.43]	1.21 [0.86–1.43]	$p = 0.189$	$p = 0.331$
COP sway range AP-EO [cm]	1.96 [1.83–2.67]	1.75 [1.36–3.87]	$p = 0.959$	2.96 [2.22–3.43]	2.96 [2.22–3.43]	$p = 0.164$	$p = 0.424$
COP sway range AP-EC [cm]	2.35 [2.21–3.01]	2.47 [2.14–2.77]	$p = 0.878$	1.77 [1.5–2.23]	1.78 [1.5–2.23]	$p = 0.157$	$p = 0.424$
COP sway range ML-EO [cm]	3.63 [2.59–4.41]	2.83 [2.31–3.48]	$p = 0.92$	3.74 [2.99–5.2]	3.74 [2.99–5.1]	$p = 0.174$	$\mathbf{p = 0.026}$
COP sway range ML-OC [cm]	1.41 [1.27–2.12]	1.73 [1.10–2.29]	$p = 0.959$	1.21 [0.65–1.49]	1.30 [1.03–1.49]	$p = 0.144$	$p = 0.691$
Ratio of axes of the ellipse-EO [%]	54.43 [48.07–57.68]	39.93 [35.21–49.86]	$p = 0.600$	62.64 [50.72–71.79]	71.79 [60.55–71.9]	$p = 0.182$	$p = 0.055$
Ratio of axes of the ellipse -EC [%]	55.96 [46.60]	48.89 [46.46–73.37]	$p = 0.646$	60.90 [48.86–81.69]	60.90 [54.77–81.69]	$p = 0.157$	$p = 0.119$
Mean speed COP AP-EO [cm/s]	1.38 [1.16–1.82]	1.29 [0.95–1.84]	$p = 0.721$	0.86 [0.83–0.98]	0.98 [0.86–1.02]	$p = 0.139$	$p = 0.361$
Mean speed COP AP-EC [cm/s]	1.90 [1.48–4.06]	1.68 [1.56–1.96]	$p = 0.874$	1.81 [1.41–4.02]	1.81 [1.41–4.02]	$p = 0.157$	$p = 0.186$
Mean speed COP ML-EO [cm/s]	0.77 [0.50–0.97]	0.66 [0.47–0.97]	$p = 0.283$	0.70 [0.38–0.71]	0.70 [0.38–0.71]	$p = 0.177$	$\mathbf{p = 0.018}$
Mean speed COP ML-EC [cm/s]	0.93 [0.66–1.54]	0.87 [0.67–1.87]	$p = 0.859$	0.99 [0.8–1.19]	1.04 [0.99–1.19]	$p = 0.128$	$p = 0.569$
Dynamic Condition							
Area-EO [cm^2]	38.47 [17.95–44.66]	3.06 [2.44–10.33]	$\mathbf{p = 0.006}$	39.50 [27.06–66.38]	39.50 [22.08–66.38]	$p = 0.177$	$\mathbf{p = 0.006}$
COP path-EO [cm]	85.00 [49.94–115.41]	28.17 [17.19–30.82]	$\mathbf{p = 0.004}$	94.26 [91.26–125.6]	94.26 [91.26–125.6]	$p = 0.240$	$\mathbf{p = 0.002}$
Trunk movement-EO [deg/s^2]	0.08 [0.08–0.16]	0.06 [0.05–0.07]	$\mathbf{p = 0.028}$	0.08 [0.07–0.11]	0.08 [0.07–0.11]	$p = 0.347$	$\mathbf{p = 0.035}$
Trunk sway range AP-EO [deg]	4.40 [2.70–5.01]	2.87 [2.37–4.20]	$\mathbf{p = 0.248}$	3.96 [3.45–4.09]	3.96 [3.45–4.09]	$p = 0.184$	$p = 0.459$
Trunk sway range ML-EO [deg]	4.97 [2.96–5.91]	1.89 [1.48–2.45]	$\mathbf{p = 0.005}$	2.37 [1.92–5]	2.37 [1.92–5]	$p = 0.664$	$\mathbf{p = 0.001}$
COP sway range AP-EO [cm]	7.16 [5.06–9.21]	3.23 [2.42–5.04]	$\mathbf{p = 0.059}$	7.82 [7.11–8.9]	7.11 [5.73–8.9]	$p = 0.378$	$p = 0.134$
COP sway range ML-EO [cm]	6.32 [4.32–7.41]	2.07 [1.55–3.81]	$\mathbf{p = 0.013}$	7.59 [5.6–10.38]	7.59 [5.6–10.38]	$p = 0.157$	$p = 0.093$
Mean speed COP AP-EO [cm/s]	2.49 [0.95–2.82]	0.56 [0.33–0.62]	$\mathbf{p = 0.012}$	2.42 [1.41–3.11]	2.42 [1.41–3.11]	$p = 0.124$	$\mathbf{p = 0.035}$
Mean speed COP ML-EO [cm/s]	1.44 [0.93–2.23]	0.59 [0.34–0.78]	$\mathbf{p = 0.021}$	1.81 [1.36–2.54]	1.81 [1.36–2.54]	$p = 0.237$	$p = 0.055$

G-Hun: experimental group G-Conv: conventional group; EO: eyes open; EC: eyes closed; COP: centre of pressure; AP: antero-posterior; ML: medio-lateral; In bold the significant results for $p < 0.05$.

4. Discussion

Evidence from the literature show that balance-specific treatment in elderly subjects undergoing knee replacement had positive effects on proprioception, postural control, balance, and coordination [53]. A meta-analysis showed that balance-specific treatment improves motor performance, i.e., balance and walking, compared to conventional (i.e., non-balance-specific) treatment alone [31]: according to the literature, traditional rehabilitation treatment alone would not be sufficient to improve balance in this type of patient [54].

So far, there is no work in the literature using robotic balance rehabilitation for patients with OA, so it is difficult to compare the results obtained from this study with those already published.

Data analysis reported that patients undergoing balance-specific treatment with the Hunova® showed a significant improvement in dynamic balance and walking, both in terms of walking speed and distance travelled, compared to the group of patients undergoing conventional rehabilitation alone. These data were also confirmed by the results of the instrumental assessment, especially in its dynamic component. In this case, data analysis showed a marked improvement in dynamic balance in the group of patients undergoing rehabilitation with Hunova®, both compared to baseline and, for some specific parameters, compared to the conventional rehabilitation group. In the latter case, in fact, patients in the Hunova® group showed a significant improvement, especially in the reduction of swaying, the reduction of swaying, and the reduction of trunk movements.

These results could be due to the fact that through the use of Hunova®, intervention training based on perturbation, proprioception, and load perception was more effective than conventional rehabilitation intervention alone.

The analysis of the results revealed two other interesting things: the first is that patients undergoing a hip replacement and rehabilitation with Hunova® showed a significant improvement in symptoms, pain, quality of life, and overall treatment efficacy compared to the group of patients undergoing the same surgery and conventional rehabilitation alone. The second is that when considering fatigue, patients in the Hunova® group showed a significant reduction in fatigue in both the motor and cognitive components. These results confirm what Castelli and colleagues [55] have already showed, albeit in a different population undergoing robotic rehabilitation with Hunova®.

This study showed that elderly patients who underwent prosthetic hip and knee replacement surgery had a significant improvement in terms of not only walking but also in the reduction of postural oscillations, and consequently showed greater stability during dynamic balance. These elements leading to improved balance performance and a reduced risk of falling. Furthermore, this study shows that fatigue, which can be a contributing factor to falls in elderly subjects, is also reduced in those patients who have undergone specific treatment for balance disorders with the Hunova® platform.

Several studies have reported that a specific rehabilitation treatment for balance and walking is effective in patients underwent surgery after OA [17,18,31,33]. To the best of our knowledge, to treat those patients, beyond conventional physical therapy, technological rehabilitation is carried out mainly by means of end-effectors and exoskeletons [25–28].

Some aspects need to be taken into account while analyzing these findings. First of all, because these are preliminary findings, further research will be required to validate the original theory. The sample size is the study's primary limitation. However, as previously mentioned, the Julious' rules for Pilot Clinical Trials [51] were used to estimate the inclusion of 12 patients each group, for a total population of 24 people. A further constraint on the research is the absence of post-protocol and post-discharge follow-up. In fact, even after being released from rehabilitation, certain longitudinal studies indicate that function may continue to improve [56].

5. Conclusions

This is the first study involving a robotic platform in the recovery of those orthopaedic patients. These preliminary findings provide a crucial foundation for more research.

Hunova® may be a useful tool for enhancing elderly persons' balance with hip or knee replacement after OA.

The risk of falls in older persons may be reduced by using this technological rehabilitation therapy, which can improve motor performance, minimize fatigue, and improve dynamic balance and ambulation.

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