ESSENTIALS OF PERFORMANCE AND HEALTH ANALYSIS IN SPORT

GYANDEV ANAND

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Table of Contents

1

Changes in the Kinematic and Kinetic Characteristics of Lunge Footwork during the Fatiguing Process

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Featured Application: 1. The kinematic and kinetic characteristics of the lunge maneuver were gradually impaired in the fatiguing process. 2. Period IV (pre-drive-off) of the stance phase showed the most significant fatigue response. 3. Improving the strength of the knee extensor is important for the lunge maneuver, particularly in period IV. 4. It is important to improve the strength of hip extensors for the lunge maneuver. 5. The analysis of waveform data is more useful for the assessment of weak areas of the body and periods of motion.

Abstract: Fatigue is a major injury risk factor. The aim of this study was to investigate the effects of fatigue on lunging during the fatiguing process. The lower extremity joint kinematics and kinetics of fifteen male collegiate badminton players were simultaneously recorded by optical motion-capture and force plate systems during lunging. In addition to statistical analyses of discrete variables, one-dimensional statistical parametric mapping (SPM (1D)) was used to analyze the waveform data. The hypotheses were that the biomechanics of lunging maneuvers would change during the fatiguing process, and the fatigue effects would differ in different periods (I–V) of the stance phase and in different joints. Results showed that the initial contact angles, peak angles, moments, power, and time needed to reach the peak angles at the hip, knee, and ankle in the sagittal plane all decreased post-fatigue. A continuous decreasing tendency was reflected in the moments and power of hip and, in particular, knee joints (mostly $p < 0.001$). Period IV showed a significant fatigue response. In conclusion, both discrete and waveform data illustrated the effects of fatigue, however, the results of SPM (1D) analysis showed both the key period and body segments affected by the fatigue response.

Keywords: badminton; knee joint; injury; one-dimensional statistical parametric mapping

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1. Introduction

Badminton, one of the most popular sports globally [1–4], is the fastest non-contact racket sport, and requires a combination of strength, speed, and stamina. While playing this sport, players must repetitively lunge, jump, and quickly change direction from a wide variety of positions. A previous epidemiology study conducted in Ireland reported that badminton was the sport with the most injuries [5]. Lower extremity injuries accounted for 43 to 86% of all injuries regardless of the nationality [6], and overuse is considered the major reason [7,8]. Ankles and knees are the most injured sites [1,7,9], particularly the knee joint [7,8]. Patellar tendinopathy [7] is the most common type of knee injury. As a specific and often-used example of footwork [10,11], the repetitive lunge is a likely cause of patellar tendinopathy, particularly for teenager players [3,12].

Moreover, epidemiology studies have also shown that a higher badminton injury rate was found at the end of a match or training [13], and higher rates have been found during training [7,14]. This could be explained by a higher intensity of training routine [7], which may induce fatigue. Because fatigue reduces the capacity of muscles to generate force, it may be an important factor causing injury. In badminton, increased ankle sprain injuries were found to occur at the end of a training or match session due to the accumulation of fatigue [6]. Due to fatigue caused by repeating the forward lunge, the activity of the vastus lateralis, vastus medialis, and biceps femoris showed a significant change [15], and knee injury risk was increased [16].

A number of studies [10,17–21] have investigated the biomechanical characteristics of the lunge, particularly its stance phase, which is defined as the period of time from initial contact to final lift-off from the ground by the dominant limb [10]. Ankle sprain [6] and patellar tendinopathy [14] may occur during the stance phase. However, the relationship between the repetitive lunge and injury is still not clear. A previous study examined the lunge under the condition of exhaustion to investigate the fatigue effect on the knee [16]. A limitation of this study is that the changes of lunge motion during the fatiguing process were ignored. Injury usually occurs at a certain time, instantaneously, and the factors inducing injury are prolonged. However, we are unaware of any studies that have investigated the biomechanical changes during the fatiguing process in badminton. Therefore, there is a lack of objective data on lower joint kinematics and kinetics of the lunge during the fatiguing process, which may provide essential insight into the understanding of the mechanism of injury.

Additionally, discrete data (related, for example, to the peak angle) has traditionally been used for statistical analysis to confirm the differences outlined above. However, it is worth noting that all kinematic and kinetic variables are continuous variables with time. Furthermore, as mentioned above, injury occurs partly due to the accumulation of changes. Thus, continuous data analysis may provide other useful information. In recent years, one-dimensional statistical parametric mapping (SPM (1D)) has been accepted as an effective method to analyze waveform data [22]. Kinematic waveform data of hip, knee, and ankle joints in three planes, measured in players with different levels, have been analyzed by SPM (1D) [18,19]. However, to the best of the authors' knowledge, no research has investigated the fatigue response by analyzing the biomechanics waveform data of the lunge.

Consequently, the purpose of this study was to investigate the changes of lunge biomechanical characteristics during the fatiguing process, from normal to fatigued states. To achieve this objective, a repetitive forehand forward lunge to exhaustion was proposed as the fatigue protocol. The fatiguing process was divided into sub-stages. In addition to the discrete data (ground reaction force (GRF), lower extremity joint angles, range of motion (RoM), moments, and power), waveform data (GRF and lower extremity joint angles, moments, and power) were also analyzed during different fatiguing stages. The hypotheses are (i) both discrete and waveform data of lunge biomechanical characteristics change significantly during the fatiguing process, especially post-fatigue; (ii) effects of fatigue vary with different periods of the stance phase; and (iii) the lower limb joints have different fatigue responses.

2. Materials and Methods

2.1. Participants

Fifteen male collegiate badminton players (age: 21.1 ± 2.2 years; height: 1.81 ± 0.04 m; weight: 72.5 ± 8.4 kg; years of badminton training: 8.9 ± 3.5 years) were recruited for the study. All participants were free from any injuries within the previous 3 months and did not take part in any high-intensity training or competitions during the two days prior to the experiment. All participants were informed of the procedures and requirements of the test, and written informed consent was obtained from each participant. In addition, a questionnaire about anthropometrics, health status, injury history, and physical activity level was completed. To minimize the potential effect of footwear, participants wore badminton shoes and socks of the same brand and series. The study was approved by the ethics committee of Beihang University (No. BM201900077).

2.2. Experimental Protocol

The test was conducted on a simulated badminton court of the biomechanics laboratory. Prior to the test, participants performed a familiarization of the forehand forward lunge and the study protocol, which included 10 min warm-up; tests of heart rate (HR), blood lactate (BL), and Borg 6–20 rating of perceived exertion (RPE); and a fatigue protocol. Considering the dependency of fatigue on the task being undertaken [23,24] and the aim of this study, repeating the forehand forward lunge until participants reached the state of exhaustion was proposed as the fatigue protocol [16]. More specifically, a forehand forward lunge cycle was defined as lunging from the starting position using the dominate limb with a sliding step, landing on the force plate positioned at the right front, hitting one shuttlecock, and then moving backward to the starting position. The degree of fatigue was estimated by values of HR [15], BL, and RPE [25], which were measured before (pre) and immediately after (post) the fatigue protocol.

A system of 9 optoelectronic cameras (Oqus 300+ Series, Qualisys AB®, Gothenburg, Sweden) and a Kistler mobile multi-component force plate (Type 9286A, Kistler, Kistler Instrument AG, Winterthur, Switzerland), integrated into the walkway, were used to record the marker position and ground reaction forces, and synchronized at 200 and 1000 Hz, respectively. According to the color atlas of skeletal (CAST) landmark definitions of the lower leg [26], twenty-eight reflective markers (diameter: 18 mm) were attached to the lower extremity for dynamic motion capture. The locations included anterior superior iliac spines, posterior superior iliac spines, thigh (markers cluster), shank (markers cluster), calcaneus, hallux, and 2nd and 5th metatarsal head of the left and right lower extremity. The markers on calcaneus, hallux, and 2nd and 5th metatarsal head were placed on the corresponding anatomical location of the badminton shoes.

2.3. Data Analysis

The kinematic and force data were obtained by the optical motion-capture system and then exported and saved as c3d files. Then, the hip, knee, and ankle joint angles, moments, power, and ground reaction force (GRF) were calculated using visual 3D software (V5, C-Motion, Bethesda, MD, USA). The raw kinematic data were filtered with a low-pass (Butterworth) filter with frequency of 20 Hz [10]. The threshold of the vertical ground reaction force (vGRF) data was set as 10 N.

The stance phase, from initial contact (heel strike) to final lift-off from the force-plate by the dominant limb, was determined by the vGRF value. During this phase, there were three vGRF peaks, namely, the initial impact peak (PF1) for heel strike transient, the secondary impact peak (PF2) for impact loading, and the third impact peak (PF3) for drive-off. Based on the classification of previous studies [10,20,21], five periods can be clearly identified in the stance phase: (I) initial contact (from heel strike to PF1); (II) impact loading (from PF1 to PF2); (III) weight acceptance (from PF2 to peak knee joint flexion angle (PAK)); (IV) pre drive-off (from PAK to PF3); and (V) drive-off (from PF3 to lift-off from ground). In addition, the fatiguing process was divided into four stages according to the slope of the HR–time curve. Six continuous lunging motions were assessed at the end of each stage, and the kinematic and kinetic data of three lunge motions were then averaged and normalized for further statistical analysis.

During the stance phase, the motion of lower extremity joints, particularly the knee, occurs primarily in the sagittal plane. This contributes to the major knee joint biomechanical characteristics in badminton research [10,21]. Based on the previous literature linked to the lunge in badminton [10,17–21,27], we analyzed the impact peak; duration of five sub-stance phases; hip, knee, and ankle joint initial contact angles; durations to peak angle; ranges of motion (RoM); peak angles; moments; and power in the sagittal plane.

2.4. Statistical Analysis

All variables of the fatiguing process were calculated during four sub-stages. The kinetic data were normalized by body weight. All discrete variables were reported as mean ± standard deviation (SD), and examined for normality using a Shapiro–Wilk test prior to statistical analysis. One-way repeated measures ANOVA was used for the analysis of the influence of fatigue on the related parameters and discrete biomechanical data. Paired *t*-tests were performed to identify the differences. All statistical procedures were performed with SPSS 25 (IBM SPSS Statistics for Window, IBM Corp., NY, USA). Additionally, the waveform data of joint angles, moments, and power in the sagittal plane were analyzed (paired *t*-test) and plotted using one-dimensional statistical parametric mapping (SPM (1D) (https://spm1d.org/)) in MATLAB (R2014b, Mathworks, Inc., Natick, MA, USA). In addition, prior to statistical analysis, normality tests were performed for the waveform data with SPM (1D). The statistical significance level was set at 0.05.

3. Results

3.1. Fatigue Protocol

After the fatigue protocol, the mean values of HR, BL, and RPE were greater than 185 beats/min, 14 mmol/L, and a score of 18, respectively. Results of repeated measurement indicated that HR, BL, and RPE were all affected by the fatigue protocol (partial η^2: 0.99, 0.876, and 0.994 respectively), and the p values were all less than 0.001 (Table 1).

Figure 1 illustrates the HR–time curve (mean) of four participants throughout the fatigue protocol. To divide the fatigue protocol into stages, time was normalized, the slope of the HR–time curve was calculated, and four stages were distinguished, namely, P1 (0–10% duration of fatiguing process (D)), P2 (10–30%D), P3 (30–60%D) and P4 (60–100%D).

Table 1. Values of heart rate (HR), blood lactate acid (BL), and rating of perceived exertion (RPE) pre- and post-fatigue protocol (mean ± SD).

	Mean ± SD			95% CI of the Difference			
	Pre	**Post**	$\eta_p{}^2$	**Lower**	**Upper**	t	p
HR (beats/min)	78.2 ± 9.4	185.8 ± 9.1	0.99	−115.013	−100.32	−32.256	<0.001
BL (mmol/L)	3 ± 2.4	14.3 ± 3.1	0.876	−14.241	−8.543	−8.801	<0.001
RPE	6 ± 0	18.4 ± 1	0.994	−13.05	−11.784	−43.176	<0.001

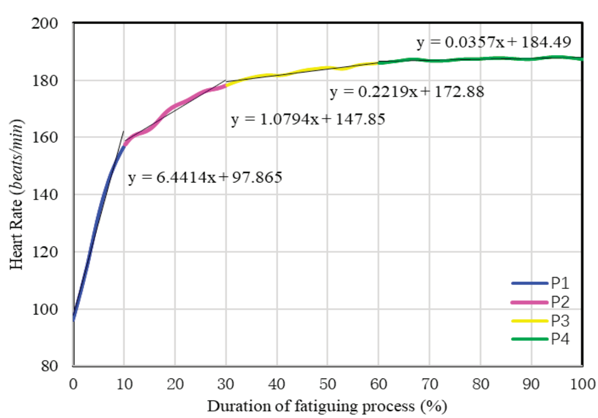


Figure 1. Heart rate (HR) of four participants during the fatiguing process (mean). According to the slope of the HR–time curve, the overall duration of the fatiguing process was divided into four stages, that is, P1, P2, P3, and P4. Duration of fatiguing process (%) vs. heart rate (beats/min).

3.2. Kinematics of Lower Extremity Joints

In the sagittal plane, Table 2 shows that the hip, knee, and ankle initial contact angle (CAH, CAK, CAA, respectively), peak angle (PAH, PAK, PAA, respectively), time to peak angle, and range of motion (RoM) all decreased during the process of the fatigue protocol (from P1 to P4).

Table 2. Kinematics variables in the four stages of fatiguing process (P1–P4) from pre- to post-fatigue protocol in the sagittal plane (mean ± SD).

	P1	P2	P3	P4	*p*
Joint angle at initial contact (°)					
CAH	46.08 ± 12.7	46.12 ± 10.32	42.2 ± 11.77	42.17 ± 11.77	*c*, *d**, *e**
CAK	13.11 ± 7.43	9.65 ± 7.55	7.82 ± 6.78	7.87 ± 6.09	*a*, *b*, *c**
CAA	10.35 ± 8.05	6.81 ± 8.25	8 ± 8.47	6.42 ± 9.11	
Peak joint angle (°)					
PAH	74.08 ± 11.94	73.3 ± 9.93	67.34 ± 11.74	67.84 ± 12.78	
PAK	69.46 ± 8.77	65.43 ± 9.44	64.17 ± 10.26	62.46 ± 8.84	*a**, *b*, *c**
PAA	−20.84 ± 5.7	−18.74 ± 5.2	−18.4 ± 4.9	−18.19 ± 6.55	
Time to peak joint angle (%)					
PAH	41.63 ± 6.59	42.13 ± 8.79	38.88 ± 7.55	36.75 ± 6.94	
PAK	45.88 ± 6.51	38.63 ± 9.12	37.25 ± 10.44	33.63 ± 9.43	*c**, *f*
PAA	14.13 ± 2.95	12.75 ± 2.82	12.38 ± 3.42	11.25 ± 2.71	*a*, *b*, *c*
Range of Motion (°)					
Hip joint	44.57 ± 10.44	39.16 ± 5.09	34.54 ± 6.85	38.67 ± 10.8	
Knee joint	59.32 ± 7.31	56.63 ± 6.84	56.74 ± 7.95	55.07 ± 6.28	*a*, *c*
Ankle joint	34.11 ± 8.09	29.82 ± 9.46	34.92 ± 9.68	31.98 ± 8.23	

Notes: CAH, CAK, and CAA: initial contact angle of hip, knee, and ankle joint, respectively; PAH, PAK, and PAA: peak joint angle of hip, knee, and ankle joint, respectively. P1–P4: the 1st, 2nd, 3rd, and 4th stages of the fatiguing process, respectively. *a*, *a**: significant differences between P1 and P2 at 0.05 and 0.01 levels, respectively; *b*: significant differences between P1 and P3 at 0.05 level; *c*: significant differences between P1 and P4 at 0.05 level; *d**: significant differences between P2 and P3 at 0.01 level; *e**: significant differences between P2 and P4 at 0.01 level; *f*: significant differences between P3 and P4 at 0.05 level.

At initial contact, a significant decrease was found for hip flexion (P1 vs. P4, $p = 0.044$; P2 vs. P3, $p = 0.009$; P2 vs. P4, $p = 0.003$) and knee flexion (P1 vs. P2, $p = 0.028$; P1 vs. P3, $p = 0.037$; P1 vs. P4, $p = 0.004$). Hip, knee, and ankle joints were all flexed (plantar-flexed) to the peak joint angle in less time. There were statistically significant differences in the time to peak value for the knee joint (P1 vs. P4, $p = 0.009$; P3 vs. P4, $p = 0.012$) and the ankle joint (P1 vs. P2, $p = 0.036$; P1 vs. P3, $p = 0.021$; P1 vs. P4, $p = 0.001$). The significance values (p) of the knee joint RoM were 0.013 (P1 vs. P2) and 0.048 (P1 vs. P4).

3.3. Five Sub-Phases of the Stance Phase

Table 3 shows that significant differences existed in the duration of the sub-stance phases II, III, IV, and V during the fatiguing process: durations were shorter for II (P1 vs. P3: $p = 0.012$), III (P1 vs. P4: $p = 0.041$; P3 vs. P4: $p = 0.005$), and V (P1 vs. P3: $p = 0.021$; P1 vs. P4: $p = 0.005$), and longer for IV (P1 vs. P2: $p = 0.022$; P1 vs. P3: $p = 0.024$; P1 vs. P4: $p = 0.002$; P2 vs. P4: $p = 0.002$; P3 vs. P4: $p = 0.036$).

3.4. Kinetics of Lower Extremity Joints

For the mean moments and power at the hip, knee, and ankle joints in the sagittal plane. Significant differences were found for moments at the hip (the 1st peak, P1 vs. P4: $p = 0.03$; the 2nd peak, P1 vs. P4: $p = 0.01$) and the knee (the 1st peak, P1 vs. P4: $p < 0.001$; the 2nd peak, P1 vs. P4: $p = 0.01$); power generation at the hip (P1 vs. P3: $p = 0.025$; P1 vs. P4: $p = 0.014$) and knee (P1 vs. P2, P1 vs. P3, P1 vs. P4: $p < 0.001$); and power consumption at the knee (P1 vs. P2: $p = 0.029$; P1 vs. P3: $p = 0.009$ and P1 vs. P4: $p < 0.001$) (details in Figure 4a).

3.5. SPM (1D) of Kinematics of Lower Extremity Joints

For the four stages of the fatiguing process (P1–P4), Figure 2a illustrates the joint angle–time curves in the sagittal plane (mean ± SD). SPM (1D) analysis results with significant differences are illustrated in Figure 2b. At the hip, significant differences exist in the initial contact (I), impact loading (II), weight acceptance (III), and pre-drive-off (IV) phases (P2 vs. P3 (I and II: $p = 0.01$; III and IV:

$p < 0.001$); P2 vs. P4 (I–IV: $p < 0.001$)). At the knee, significant differences were found mainly in the weight acceptance and pre-drive-off phases (P1 vs. P2 and P1 vs. P4: $p < 0.001$).

Table 3. Duration of five sub-phases (%) (mean ± SD).

	P1	P2	P3	P4	p
Duration of Five Sub-Phases (%)					
I (0-PF1)	3.88 ± 0.64	4 ± 0.93	4 ± 0.76	3.75 ± 0.71	
II (PF1-PF2)	11.13 ± 2.75	8.63 ± 4.44	8 ± 3.55	8.88 ± 3.31	*b*
III (PF2-PAK)	30.88 ± 6.03	26 ± 7.11	25.25 ± 9.25	21 ± 9.3	*c*, *f**
IV (PAK-PF3)	27.25 ± 10.69	36.5 ± 14.32	39.25 ± 17.17	43.38 ± 15.24	*a*, *b*, *c**, *e**, *f*
V (PF3-FO)	26.88 ± 5.57	24.88 ± 7.24	23.5 ± 7.23	23 ± 6.68	*b*, *c**

Notes: PF1: initial impact peak (the 1st peak vertical ground reaction force, vGRF); PF2: secondary impact peak (the 2nd peak vGRF); PF3: third impact peak (the 3rd peak vGRF); PAK: peak knee joint flexion angle. FO: foot off. The stance phase was divided into five sub-phases (I–V) according to impact peak and PAK. P1–P4: the 1st, 2nd, 3rd, and 4th stages of the fatiguing process, respectively. *a*, *a**: significant differences between P1 and P2 at 0.05 and 0.01 levels, respectively; *b*: significant differences between P1 and P3 at 0.05 level; *c*, *c**: significant differences between P1 and P4 at 0.05 and 0.01 levels, respectively; *e**: significant differences between P2 and P4 at 0.01 level; *f*, *f**: significant differences between P3 and P4 at 0.05 and 0.01 levels, respectively.

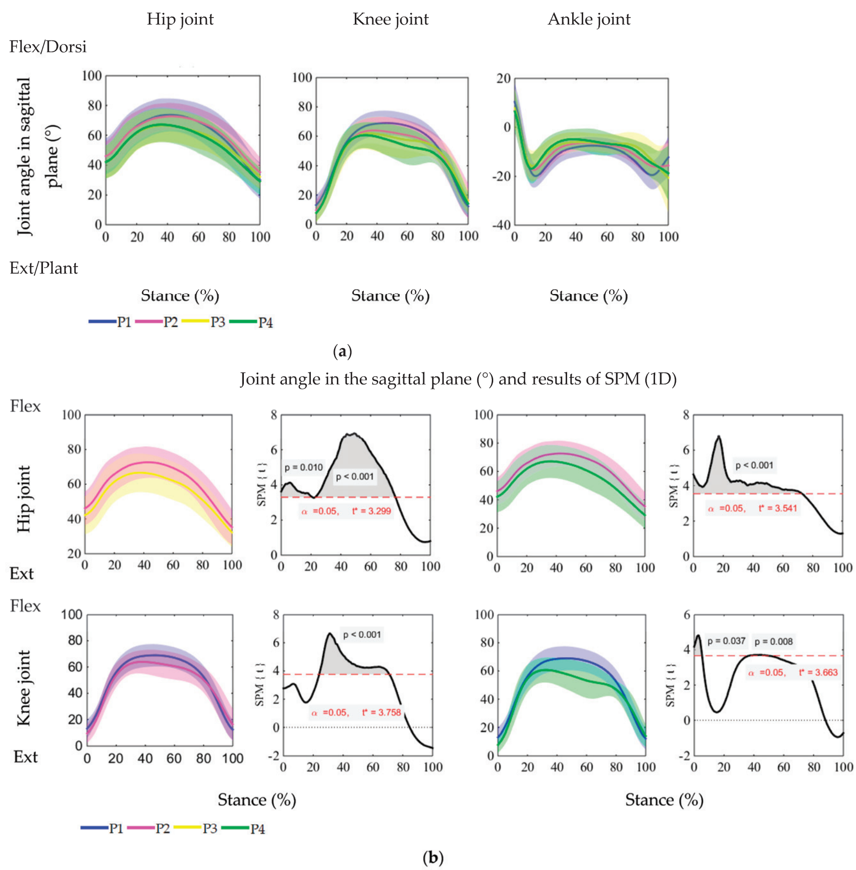


Figure 2. (**a**) Summary of the joint angles (mean ± SD) of lower limbs in the sagittal plane. (**b**) Results with significant differences of one-dimensional statistical parametric mapping (SPM (1D)) for hip and knee joint angles. Positive angles represent hip and knee flexion and ankle dorsi-flexion. P1–P4: the 1st, 2nd, 3rd, and 4th stages of the fatiguing process, respectively.

3.6. SPM (1D) of Kinetics of Lower Extremity Joints

Figure 3 illustrates the vGRF waveform data of P1, P2, P3, and P4, and the results of SPM (1D) analysis with significant differences, which are shown mainly in the pre-drive-off phases between P1 and P4 ($p < 0.001$).

Figure 4a illustrates the waveform data of the hip, knee, and ankle joint moments and power in the sagittal plane (mean ± SD) in P1, P2, P3, and P4. SPM (1D) analysis results of moments and power with significant differences are illustrated in Figure 4b,c, respectively. For hip and knee moments, significant differences exist in the partial period of the pre-drive-off phase (IV) between P1 and P3 ($p < 0.001$), and in the IV phase between P1 and P4 ($p < 0.001$). Moreover, Figure 4c illustrates that significant differences exist in the partial IV and drive-off (V) phases for hip power (P1 vs. P4: $p < 0.05$) and knee power (P1 vs. P3: $p < 0.001$; P1 vs. P4: $p < 0.001$).

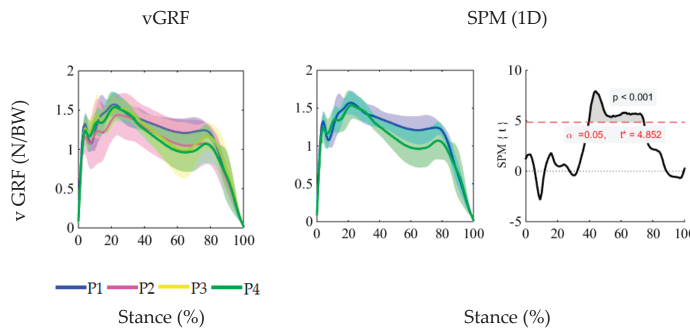
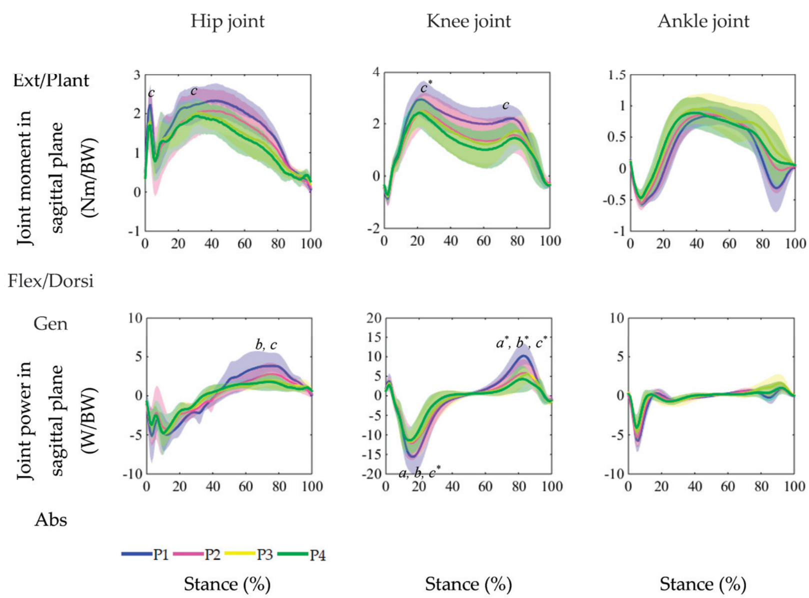
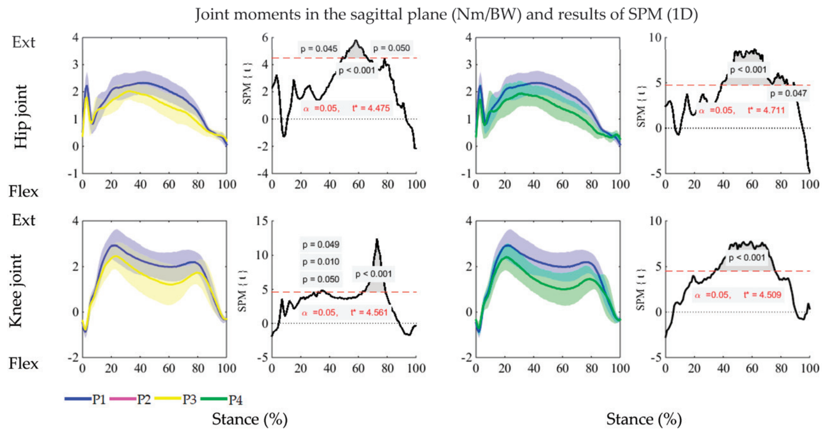


Figure 3. Vertical ground reaction force (vGRF) and results with significant differences of one-dimensional statistical parametric mapping (SPM (1D)). P1–P4: the 1st, 2nd, 3rd, and 4th stages of the fatiguing process, respectively.



(**a**)

Figure 4. *Cont.*



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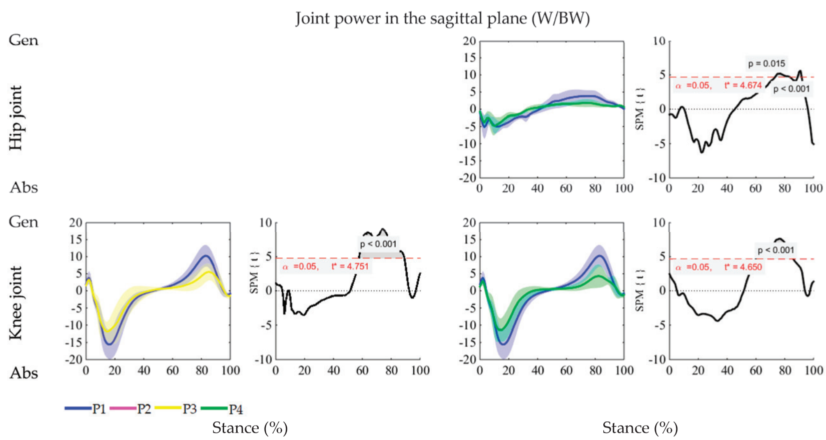

(**c**)

Figure 4. (**a**) Summary of joint moments and power (mean ± SD) of lower extremities in the sagittal plane during the lunge. (**b**) Results with significant differences of one-dimensional statistical parametric mapping (SPM (1D)) for hip and knee joint moments. (**c**) Results with significant differences of SPM (1D) for hip and knee joint power. Positive moments represent hip and knee extensor and ankle plantar flexor moments, and positive joint power indicates periods of power generation. P1–P4: the 1st, 2nd, 3rd, and 4th stages of the fatiguing process, respectively. *a*, *a**: significant differences between P1 and P2 at 0.05 and 0.01 levels, respectively; *b*: significant differences between P1 and P3 at 0.05 level; *c*, *c**: significant differences between P1 and P4 at 0.05 and 0.01 levels, respectively.

4. Discussion

An increasing number of people now play badminton. Both athletes and recreational players attempt to optimize their performance, thus increasing the risk of injury. Prevention of sports-related injuries is an important challenge. Fatigue is a major factor causing injury. Consequently, this study investigated the fatigue effects on a specific movement in badminton, i.e., the footwork associated with the lunge, which is one of the most used and integral movements [10,11].

Considering the task dependency of fatigue, a repeated forehand forward lunge, until reaching exhaustion, was proposed as the fatigue protocol, which was subdivided into four stages (P1, P2, P3, and P4) according to the slope of the heart rate (HR)–time curve. After the protocol, the mean values of HR, blood lactate (BL), and rating of perceived exertion (RPE) increased significantly at the significance level of 0.001; in particular, the mean HR was greater than 185 beats/min, the mean BL value was greater than 14 mmol/L, and the mean RPE score was greater than 18, indicating that all participants were fatigued. At P1, the first stage of the fatigue protocol, all participants were in a pre-fatigue state and, at P4, the final stage of the fatigue protocol, they were fatigued. In addition, according to previous studies [10,20,21], the first, second, and third impact peak (PF1, PF2, and PF3) and knee flexion peak angle (PAK) were used to subdivide the lunge stance phase into initial contact (I: 0-PF1)), impact loading (II: PF1-PF2), weight acceptance (III: PF2-PAK), pre drive-off (IV: PAK-PF3), and drive-off (V: PF3-end) periods. Then, statistical analyses were undertaken for both discrete and waveform kinematic and kinetic data in the sagittal plane, in which the largest movements occurred, comparing not only the pre- and post-fatigue states, but also the four stages of the fatiguing process.

The results supported the hypothesis that the biomechanical characteristics of the lunge change significantly during the fatiguing process. At the initial contact time, participants exhibited a more "erect" posture for the lunge, which is usually observed by coaches, and was shown as less hip and knee flexion, and ankle dorsi-flexion. Less dorsi-flexion at the foot strike has been found for recreational players with a relative lack of muscle power compared to national-level badminton athletes [19]. In the lateral jump performed in badminton, Herbaut et al. [27] found a decreased plantar-flexion angle at the foot strike post-fatigue. These changes may be induced by muscle fatigue caused by repeated stretch-shortening. Furthermore, the range of motion (RoM) of hip, knee, and ankle joints was decreased due to the fatiguing process.

Smaller peak joint angles were found at hip, knee, and ankle joints in the sagittal plane, particularly at the knee joint (with significant differences between P1 and the other three stages of the fatiguing process). After fatigue, a decrease in the knee peak angle was also found by Valldecabres et al. [16], however, no significant difference was found. The discrepancy may be due to the participant's sports level [24]. The decrease in the peak angle could be explained by the decrease in joint moments (details in Figure 4). There were significant differences in peak joint moments at the hip and knee extensor (P1 vs. P4: hip ($p = 0.03$), knee ($p < 0.001$)). A similar relationship was illustrated by Fu et al. [19] between professional and amateur badminton players; that is, professional players with greater muscle strength and better performance showed higher knee and ankle joint moments. Additionally, a shorter time was taken to flex to the peak angle, with significant decreases in time shown at the knee and ankle. This may be caused by impaired control due to fatigue. This can also be explained in this study by the decrease in joint moments and power.

In addition to the angles, the durations of the four sub-stance phases (II–V) changed during the process of the fatigue protocol. The shorter durations of II and III indicate that less time was taken to reach PF2 and PAK, respectively; that is, due to the fatiguing process, the participant placed his foot flat and flexed his knee more quickly. This is consistent with the increased ankle plantar flexor and knee flexor moments in the present study (Figure 4a), and with the opinion that fatigue reduces the capacity of muscles to generate force. The most significant increase was found in the IV phase (pre-drive-off), increasing from 27.25 ± 10.69% stance to 43.38 ± 15.24% stance. Significantly decreased hip and knee peak power in this phase provided sufficient support for this change. Less power was generated for players for the drive-off. Kuntz et al. [10] indicated that a hop style lunge generates higher peak vertical force during loading. In this study, three participants used the hop style at the final stage of the fatigue protocol. The change of lunge style may be a strategy to generate more power for driving-off and returning to the starting position [14].

Furthermore, considering the time continuity of biomechanical variables of the lunge motion, one-dimensional statistical parametric mapping SPM (1D) was used to analyze the biomechanical waveform data in the sagittal plane. Most of the joint angle, moment, and power waveforms decreased consistently during the process of the fatigue protocol (among P1, P2, P3, and P4). Significant differences were found in the hip and knee joint angles, mainly in the pre-drive-off phase (IV) (most p values were less than 0.001). The results support the hypothesis that the fatigue effects were different in the five periods of the lunge. Moreover, these results support the view that lunge characteristics change due to fatigue.

Taking into account the joint moments, both the hip and knee joints showed a significant response to the fatigue protocol. Significant differences were found between P1 and P3 in part of the period of IV, and between P1 and P4 in IV. The hip moments had a larger effective scope. However, there were only significant differences in the hip power between P1 and P4. For knee power, the significant differences were seen between P1 and P3, and also between P1 and P4, with a larger effective scope of the period. This indicates that the significant decrease in joint power occurred earlier and was mostly evident at the knee; that is, the fatigue responses manifested mostly at the knee joint. This result is consistent with a previous epidemiology study that reviewed musculoskeletal injuries among Malaysian badminton players, which reported that the majority of injuries sustained by players were due to overuse, primarily of the knee [7]. Another study [28] suggested that the rapidly changing eccentric/concentric work of the quadriceps in the varying degree of knee flexion was probably associated with patellar tendon injury. This may be why patellar tendinopathy is the most common injury of lower limbs among badminton players, and this result also supports our third hypothesis. Coaches and players should pay more attention to the training of the knee, and particularly the knee extensors.

Additionally, a comparison indicates that the discrete and waveform results are consistent. There are significant differences in CAH (P2 vs. P3), CAK (P2 vs. P4), PAK (P1 vs. P2; P1 vs. P4), knee peak moment (P1 vs. P4), hip peak power (P1 vs. P4), and knee peak power (P1 vs. P3; P1 vs. P4). In addition, significant differences were shown in these periods of stance phase in the discrete data. The results of SPM (1D) analysis clearly provided more information, indicating the period rather than a point of time in the fatigue response. This is also helpful in identifying the key body segments affected by fatigue. Thus, it is important for coaches and players to design a corresponding training program to improve the technique and muscle strength. Moreover, the results of SPM (1D) can provide support for monitoring of training.

Considering the key findings of this study, a few limitations should be noted when interpreting the results. First, the participants were male badminton players with at least 8 years of special badminton training. Results may differ for players of different levels, ages, and gender, thus studies of players with a range of abilities and ages, including female players, should be considered. Second, all tests were conducted on a simulated badminton court. Third, the sample size was limited. Fourth, the changing slope of the HR–time curve was used to subdivide the fatiguing process. However, it is not sufficient to explain the status of fatigue. Electromyography (EMG) data might be more suitable and could be used in future work. Finally, although joint moments and power allow further assessment of the functional contribution of the joints, EMG and musculoskeletal system simulations would help understand movement changes during the fatiguing process.

5. Conclusions

This study investigated the changes in the kinematic and kinetic characteristics of lunge footwork during the fatiguing process. To the best of our knowledge, this study was the first to subdivide the fatiguing process and analyze the changes among different sub-stages. It was also the first study to use one-dimensional statistical parametric mapping (SPM (1D)) to analyze the effect of fatigue on the lunge footwork in badminton. Statistically significant differences were found in both the discrete and waveform data. Moreover, these differences were shown not only between pre- and post-fatigue, but also among other sub-stages.

Overall, the results presented in this study confirm that period IV of the stance phase is more sensitive to fatigue than the other periods. In addition, the training program should focus on muscular strengthening of the knee extensor, particularly in period IV. It is also important to improve the strength of hip extensors. Furthermore, although the results show consistent changes between the discrete and waveform data, findings from this study highlight that results of SPM (1D) are more useful for the assessment of weak areas of the body and periods of motion. This information may contribute to the future design and development of training plans and to the monitoring of training.

Author Contributions: Y.D. designed, carried out the experiments, analyzed the data and wrote the paper, Y.F. designed the experiments, reviewed and revised the paper. All authors have read and agreed to the published version of the manuscript.

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2

Concurrent Validity and Reliability of My Jump 2 App for Measuring Vertical Jump Height in Recreationally Active Adults

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Featured Application: My Jump 2 app is a valid and reliable tool for the assessment of vertical jump in recreationally active participants. It is relatively easy to use, affordable, and portable. My Jump 2 can be used in different fields as an alternative to laboratory testing.

Abstract: This study aimed to examine the reliability, validity, and usefulness of the smartphone-based application, My Jump 2, against Optojump in recreationally active adults. Participants (18 women, 28.9 ± 5.6 years, and 26 men, 30.1 ± 10.6 years) completed squat jumps (SJ), counter-movement jumps (CMJ), and CMJ with arm swing (CMJAS) on Optojump and were simultaneously recorded using My Jump 2. To evaluate concurrent validity, jump height, calculated from flight time attained from each device, was compared for each jump type. Test-retest reliability was determined by replicating data analysis of My Jump 2 recordings on two occasions separated by two weeks. High test-retest reliability (Intraclass correlation coefficient (ICC) > 0.93) was observed for all measures in both male and female athletes. Very large correlations were observed between the My Jump 2 app and Optojump for SJ ($r = 0.95$, $p = 0.001$), CMJ ($r = 0.98$, $p = 0.001$), and CMJAS ($r = 0.98$, $p = 0.001$) in male athletes. Similar results were obtained for female recreational athletes for all jumps ($r > 0.94$, $p = 0.001$). The study results suggest that My Jump 2 is a valid, reliable, and useful tool for measuring vertical jump in recreationally active adults. Therefore, due to its simplicity and practicality, it can be used by practitioners, coaches, and recreationally-active adults to measure vertical jump performance with a simple test as SJ, CMJ, and CMJAS.

Keywords: measurement; healthy athletes; jump performance; smartphones; My Jump 2; reliability; validity

1. Introduction

Physical fitness is important for older adults to maintain their independence and enhance wellbeing [1]. Therefore, it is of great importance to measure physical fitness in adults regularly. Vertical jump tests were recognized as the most common means for assessing physical fitness in various populations [2–4]. Moreover, a fundamental step in jump training studies is a vertical jump test. It is also a common method for assessing lower limb power in a physical education class, gym, or other sports programs [5]. Furthermore, it serves as an indicator of athletes fatigue during in-season [6]. Due to its simplicity and important outcome information, vertical jump tests are broadly used by coaches, strength and conditioning professionals, and professionals in health care. The most frequently used vertical jumps are squat jump (SJ), counter-movement jump (CMJ), and drop jump (DJ).

The most commonly used instruments for measuring the vertical jump characteristics have been photoelectric cell systems, force platforms, linear position transducers, infrared cells, contact mats, and video recording [7–12]. The great majority of mentioned instruments presents good validity and reliability in measuring different jumps with the force plate considered as the "gold standard". However, most of the above-mentioned instruments are not cheap and not widely accessible for different populations. Accordingly, due to the fact that these tools are expensive and not easy for transport, practical value for measuring vertical jumps in recreationally active adults is questionable.

Technology improvements led to the integration of high-speed cameras in mobile phones. The mobile application My Jump 2 takes advantage of these cameras to record slow-motion videos of different jump tasks. It gives us information about jump height by selecting the take-off and landing frame. Its validity and reliability were previously reported in male sport science students for drop jumps [8], elderly people [13], and in professional cerebral palsy football players for SJ and CMJ [14]. To the authors' knowledge, there is only one study [15] that analyzed the concurrent validity and reliability of a My Jump app for measuring vertical jump in recreationally active adults. However, the participants were younger men (22 years), and only CMJ was evaluated. There is evidence that the reliability of jumping explosiveness in physical performance tests might vary between men and women [16]. Therefore, it is necessary to check the validity and reliability of the My Jump app with recreational male and female adults. Moreover, in the study mentioned above, My Jump has recorded videos with iPhone 5 s app at 120 fps. As mentioned earlier, the key limiting factor to the accuracy of the app is the frame rate [8]. Therefore, the 240 fps camera on iPhone X was expected to make a significant improvement in the app's performance regarding reliability and validity.

Due to smartphone apps popularity, portability, affordability, and advanced technology, it is important to check the accuracy of these apps for measuring variables related to physical performance and health. Therefore, the present study aimed to investigate the reliability, validity, and usefulness of the My Jump 2 app in comparison to reliable and validated Optojump photoelectric cells system in measuring SJ, CMJ, and CMJ with arm swing (CMJAS) in recreationally active adults. The current research covered a heterogeneous sample with a bigger age range as contrasting to the homogeneous sample in most studies. Our goal was to reassess the app validity in a more heterogeneous sample that has diverse jumping capabilities in order to overcome possible errors in measurements.

2. Materials and Methods

2.1. Participants

A total of 44 participants volunteered to partake in the research. The sample consisted of 18 women (age—28.9 ± 5.6 years; height—169.6 ± 6.2 cm; weight—60.5 ± 8.7 kg) and 26 men (age—30.1 ± 10.6 years; height—178.2 ± 16.2 cm; weight—85.9 ± 23.8 kg) who were recreationally active and had membership in the local gym in Subotica, Serbia where the testing was performed. Participants completed general health and demographic survey and were excluded if they had a history of diseases, injuries in the past six months, or physical condition that may affect testing. All participants were asked if they regularly participated in vigorous physical activity and about the type of activity. Additionally, data were collected regarding the training background and training frequency during one week. On the day of testing, they were healthy, without any heart or pulmonary disease, and injury-free. Before the testing, they were not involved in any strength, jumping, or high-intensity training for 48 h. They were informed about the testing procedures, and before the start, they signed written informed consent. The research adhered to the Declaration of Helsinki and was approved by the local ethics committee (ref. 12/1041).

2.2. Procedures

All participants were familiarized with SJ, CMJ, and CMJAS techniques one day before testing at the same place where the testing was conducted. Assistants also have introduced the participants with the proper technique before testing by video and live demonstration and the explanation of the correct technique.

Before testing, they carried out a standardized 10 min warm-up that consisted of lower-body dynamic stretches, jogging, skipping, and vertical jumps based on similar jump warm-up protocols used in previous studies [15,17]. Their body mass was measured to the nearest 0.1 kg with electronic scale TANITA BC 540 (TANITA Corp., Arlington Heights, IL, USA) and body height with a stadiometer (SECA Instruments Ltd., Hamburg, Germany) to the nearest 1 cm. The leg length and height with bended knees at about 90° were measured using a measuring tape to the nearest 1 cm. Leg length was measured from the anterior iliac spine to the tiptoe in the laying position. Height at 90° was measured vertically from the anterior iliac spine to the ground in an optimal jump performance position (the angle at approximately 90°). Then, each participant performed three SJs, three CMJs, and three CMJs free arms with the instruction to jump as high as possible. For all jumps, it was recommended that the participants leave the floor at take-off with the knees and ankles extended and land in a similarly extended position [18]. Between the trials, there was a two-minute passive rest. The highest jump of each technique was taken into analysis. The jumps were recorded with the Optojump photoelectric cell system (Optojump photocell system; Microgate, Bolzano, Italy) and with an iPhone X (Apple Inc., Cupertino, CA, USA) through My Jump 2 app at the same time. The participants repeated the testing procedure after two weeks with the same conditions and in the same order as during the first testing.

Squat jump performance [19]

Participants were instructed to start the jump in the position of 90° knee flexion with the feet shoulder-width apart and with their hands on their waist. They were asked to jump for maximum height and maintain their hand on the waist. Counter-movement was discouraged, and in case of any mistake, the jump was repeated.

Counter-movement jump performance [20]

The CMJ starting position was a standing position with a straight torso and knees fully extended with the feet shoulder-width apart. Participants were asked to keep their hands on their waist throughout the whole jump. They were instructed to perform a quick downward movement (approximately 90° of knee flexion), and afterward a fast upward movement to jump as high as possible.

Counter-movement jump free arms performance

The CMJAS technique is similar to CMJ with the exception of arm movement. Participants were instructed to swing back with their arms during downward movement and forward during upward movement.

Optojump photoelectric cell system

The Optojump system consists of two parallel bars placed approximately 1 m apart and parallel to each other (see Figure 1). The bars are equipped with 33 optical light-emitting diodes (LEDs) with continuous communication of the transmitting and the receiving bar. The LEDs are positioned 0.3 cm from the ground level and at a 3.125 cm interval. The height of the jump is calculated as follows: $h = 0.5g \times t^2$, where h is the height of the jump, g is the acceleration of gravity, t is half of the flight time.The Optojump achieved strong concurrent validity for jump height in comparison with the force platform (ICC = 0.99; 95% CI (confidence interval) = 0.97; 0.99; $p < 0.001$) and was recognised as an reliable instrument for field-based vertical jump assessments [18].

My Jump 2 app

The app My Jump 2 for iPhone X was used to calculate the jump height by manually selecting the take-off frame and landing frame (Figure 1) of the video. The app determines the jump height using the equation $h = t^2 \times 1.22625$ described by Bosco et al. [21] where *h* stands for the jump height (in meters) and *t* for flight time (in seconds). All collections were made with the same phone and by the same evaluator with no professional experience in video analysis. The evaluator was always recording from the same position (approximately 1 m height) and with the same distance from the participants (approximately 1.5 m), enabling the clear view of participants lower limbs. We used the sagittal plane because it showed that identification of the exact take-off and landing frames was more easily viewed, compared to a frontal plane view [22].

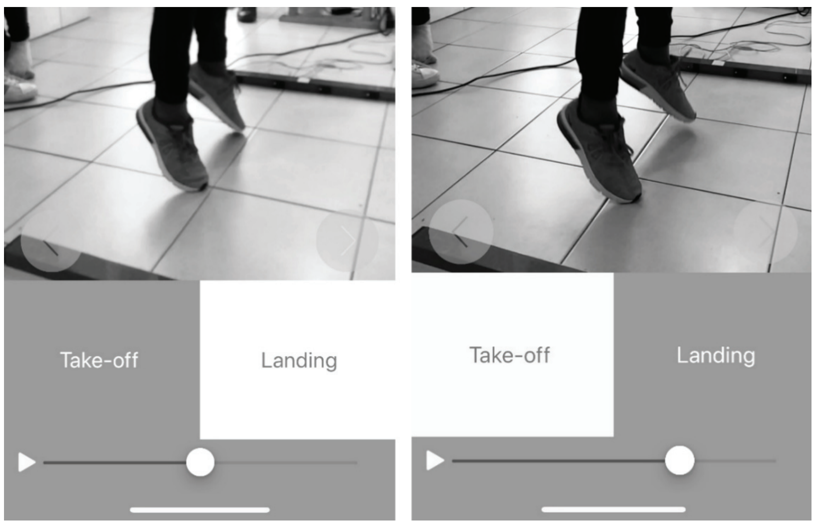

Figure 1. Take-off and landing phase frames on My Jump 2 app.

2.3. Statistical Analysis

Descriptive statistics were presented using means and standard deviations. Shapiro–Wilk test was used to check the data normality. Systematic bias between sessions and tools was evaluated using the paired samples *t*-test [18]. Standardized differences in mean (with 95% confidence intervals; CI) were calculated to determine the magnitude of the change across and between tests. According to Hopkins et al. [16], Cohen d effect size (ES) magnitudes of change were classified as trivial (>0.2), small (0.2–0.5), moderate (0.5–0.8), large (0.8–1.60), and very large (>1.60). Reliability between test-retest was analyzed using intraclass correlation coefficient (ICC), typical error (TE) expressed as coefficient of variation (CV%), and smallest worthwhile change (SWC) according to Excel spreadsheet provided by Hopkins (2007) [23]. Regarding the ICC analysis, a single measure, two-way mixed, absolute-agreement parameter was used [24]. The highest jump from each subject on both testing sessions, retrieved from the My Jump 2, was used. ICC was interpreted as <0.1 = low, <0.3 = moderate, <0.5 = high, <0.7 = very high, <0.9 = nearly perfect, and <1.0 = perfect. A good reliability was considered if following criteria was fulfilled: CV < 5% and ICC > 0.69 [25]. Test usefulness was determined based on the comparison of SWC (0.2 multiplied by the between-subject SD, based on Cohen's ES) to TE [26]. The following criteria were used to establish the usefulness of tests: "Marginal" (TE > SWC), "OK" (TE = SWC), and "Good" (TE < SWC).

The concurrent validity of the app was tested with Pearson's product-moment correlation coefficient (r). Additionally, the agreement between Optojump and My Jump 2 data was then examined graphically using Bland and Altman's plots in which the difference between both devices was plotted against the mean of the two devices [27].

3. Results

Participants' descriptive characteristics are presented in Table 1.

Table 1. Descriptive characteristics.

	Male (n = 26)	Female (n = 18)
Age (years)	30.1 ± 10.6	28.9 ± 5.6
Height (cm)	178.2 ± 16.2	169.6 ± 6.2
Weight (kg)	85.9 ± 23.8	60.5 ± 8.7
Leg length (cm)	108.1 ± 4.7	106.1 ± 4.5
Years of training	10.5 ± 7.6	9.8 ± 6.6
Training hours per week	6.2 ± 2.1	3.9 ± 1.1

Note: Values are expressed as mean ± SD.

3.1. Reliability

Similar SJ (test = 29.6 ± 6.0 cm; retest = 30.8 ± 6.6 cm), CMJ (test = 31.9 ± 6.6; retest = 34.2 ± 6.9 cm) and CMJAS (test = 39.4 ± 9.7 cm; retest = 39.7 ± 10.0 cm) values were observed between testing sessions in male recreationally active adults. Non-significant differences ($p > 0.05$) were observed between testing sessions for SJ (ES = trivial; CI 95% (0.4; 2.1)), CMJ (ES = small; CI 95% (1.6; 2.9)), and CMJAS (ES = trivial; CI 95% (−0.5; 1.1)) as observed in Table 2. High test-retest reliability (ICC > 0.93; TE < 5% for CMJ and CMJAS, respectively) was observed for all measures.

Table 2. Test-retest reliability and usefulness of My Jump 2 in male recreationally active adults.

	SJ	CMJ	CMJAS
Test (cm)	29.6 ± 6.0	31.9 ± 6.6	39.4 ± 9.7
Retest (cm)	30.8 ± 6.6	34.2 ± 6.9	39.7 ± 10.0
ES	0.19 (trivial)	0.34 (small)	0.03 (trivial)
Diff (95% CI)	1.2 (0.4; 2.1)	2.3 (1.6; 2.9)	0.3 (−0.5;1.1)
ICC (95% CI)	0.93 (0.86;0.96)	0.96 (0.93; 0.97)	0.97 (0.95; 0.99)
TE (95% CI)	1.8 (1.5;2.3)	1.3 (1.1;1.7)	2.0 (1.6;2.6)
CV% (95% CI)	5.8 (4.7; 7.6)	4.1 (3.4; 5.5)	5.0 (4.0; 6.6)
SWC%	1.2 (4.3%)	1.3 (4.0%)	2.0 (5.3%)
Rating	marginal	OK	OK

Abbreviations: SJ, squat jump; CMJ, countermovement jump; ES, effect size; Diff, difference; CI, confidence interval; ICC, intraclass correlation coefficient; TE, typical error; CV, coefficient of variation; SWC, smallest worthwhile change.

Table 3 shows the test retest results for SJ (test = 23.9 ± 6.0 cm; retest = 25.8 ± 6.8 cm), CMJ (test = 26.8 ± 6.3; retest = 27.3 ± 6.2 cm), and CMJAS (test = 29.3 ± 6.0 cm; retest = 30.2 ± 6.4 cm) in female recreationally active adults. There were no significant differences ($p > 0.05$) between testing sessions for SJ (ES = small; CI 95% (1.0; 2.8)), CMJ (ES = trivial; CI 95% (−0.1; 1.1)), and CMJAS (ES = trivial; CI 95% (0.2; 1.6)). High test-retest reliability (ICC > 0.94; TE < 5% for CMJ and CMJAS, respectively) was observed for all measures.

3.2. Test Usefulness

The TE for SJ for both male and female participants was greater than the presumed SWC; consequently, these measures were rated as "marginal." In contrast, TE for CMJ and CMJAS for both genders were similar or lower than SWC and was rated as "OK" and "good".

Table 3. Test-retest reliability and usefulness of My Jump 2 in female recreationally active adults.

	SJ	CMJ	CMJAS
Test (cm)	23.9 ± 6.0	26.8 ± 6.3	29.3 ± 6.0
Retest (cm)	25.8 ± 6.8	27.3 ± 6.2	30.2 ± 6.4
ES	0.30 (small)	0.08 (trivial)	0.15 (trivial)
Diff (95% CI)	1.90 (1; 2.8)	0.5 (−0.1; 1.1)	0.9 (0.2; 1.6)
ICC (95% CI)	0.94 (0.86; 0.97)	0.97 (0.93; 0.98)	0.97 (0.92; 0.98)
TE (95% CI)	1.6 (1.3; 2.3)	1.1 (0,8; 1.5)	1.2 (0.9; 1.6)
CV% (95% CI)	7.2 (5.6; 10.3)	4.3 (3.4; 6.1)	4.3 (3.3; 6.0)
SWC%	1.3 (5.1%)	1.2 (4.5%)	1.2 (4.2%)
Rating	marginal	good	OK

3.3. The Validity of the Test

There were no significant differences ($p > 0.05$) between the My Jump 2 app and Optojump for all jumps in male participants with trivial effects size (from −0.03 to −0.09) (Table 4). Very large correlations were observed between the My Jump 2 app and Optojump for SJ (r = 0.95, $p = 0.001$), CMJ (r = 0.98, $p = 0.001$), and CMJAS (r = 0.98, $p = 0.001$).

Table 4. Descriptive statistics and validity analysis in male recreationally active adults based on Pearson's r.

	My Jump 2	Optojump	Diff. (95% CI)	ES	r (95% CI)	Rating
SJ	29.6 ± 6.0	30.0 ± 6.3	0.40 (−3.26; 2.46)	−0.07	0.95 (0.91; 0.97)	Very large
CMJ	31.9 ± 6.6	32.5 ± 7.1	0.60 (−3.79; 2.59)	−0.09	0.98 (0.95;0.99)	Very large
CMJAS	39.4 ± 9.7	39.7 ± 9.5	0.30 (−4.76; 4.16)	−0.03	0.98 (0.97;0.99)	Very large

Abbreviations: r, Pearson's correlation coefficient.

Similar results were obtained for female recreationally active adults (Table 5). No significant differences ($p > 0.05$) were observed between the My Jump 2 app and Optojump for all jumps in female recreational athletes with trivial effects size (from −0.09 to −0.19). Very large correlations were observed between the My Jump 2 app and Optojump for all jumps (r > 0.94, $p = 0.001$).

Table 5. Descriptive statistics and validity analysis in female recreationally active adults based on Pearson's r.

	My Jump 2	Optojump	Diff. (95% CI)	ES	r (95% CI)	Rating
SJ	23.9 ± 6.0	24.5 ± 7.5	0.6 (−3.79; 2.59)	−0.09	0.97 (0.93; 0.99)	Very large
CMJ	26.8 ± 6.3	27.7 ± 7.8	0.9 (−4.90; 3.10)	−0.13	0.96 (0.91;0.98)	Very large
CMJAS	29.3 ± 6.0	30.7 ± 8.4	1.4 (−5.51; 2.71)	−0.19	0.94 (0.87;0.98)	Very large

Figures 2–4 show the level of agreement for all jumps. Bland and Altman's plot depicting limits of agreement for SJ height between the Optojump and My Jump 2 show that the majority of data points are within the 95% CI's (Figure 2).

Further analysis of the Bland–Altman plots in male athletes revealed very low R^2 values ($R^2 \leq 0.10$), meaning outcomes estimated from My Jump 2 had no predisposition to overestimate or underestimate jump performance. On the contrary, in female participants, the plot shows bias related to the magnitude of jump height ($R^2 = 0.74$), such that, at lower jump heights, values derived from Optojump data tended to be higher than those from My Jump 2, resulting in positive difference scores. Moreover, the mean bias between the two methods for all jumps was 0.51 cm.

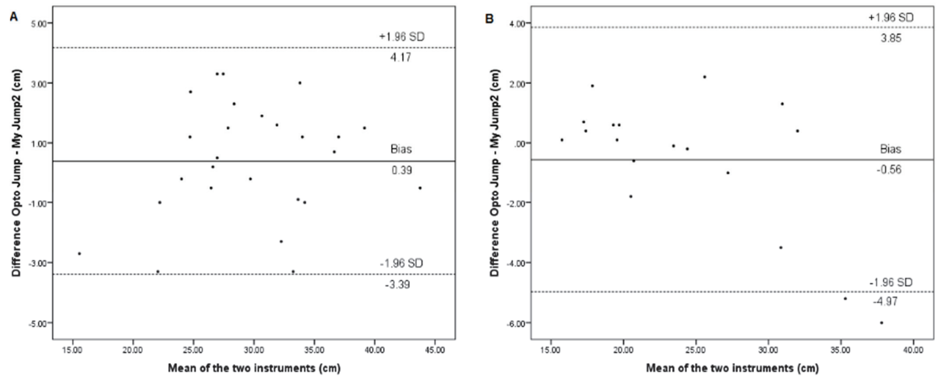

Figure 2. Level of agreement (Bland–Altman) with 95% limits of agreement (dashed lines) and the mean difference (solid line) between My Jump 2 and the Optojump for SJ in (**A**) male and (**B**) female participants.

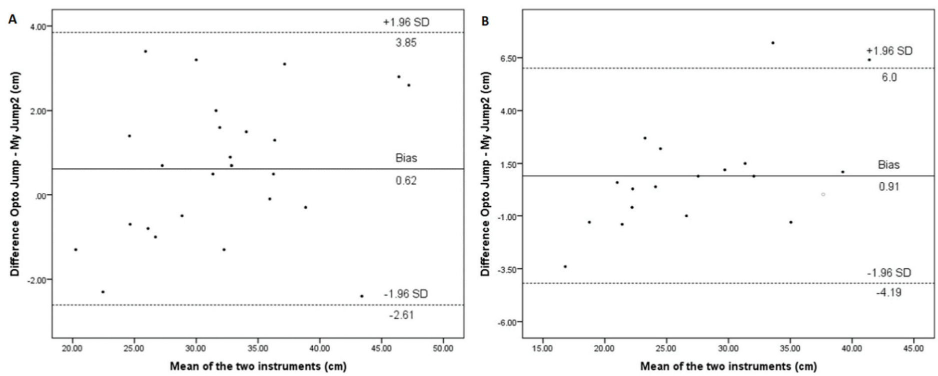

Figure 3. Level of agreement (Bland–Altman) with 95% limits of agreement (dashed lines) and the mean difference (solid line) between My Jump 2 and the Optojump for CMJ in (**A**) male and (**B**) female participants.

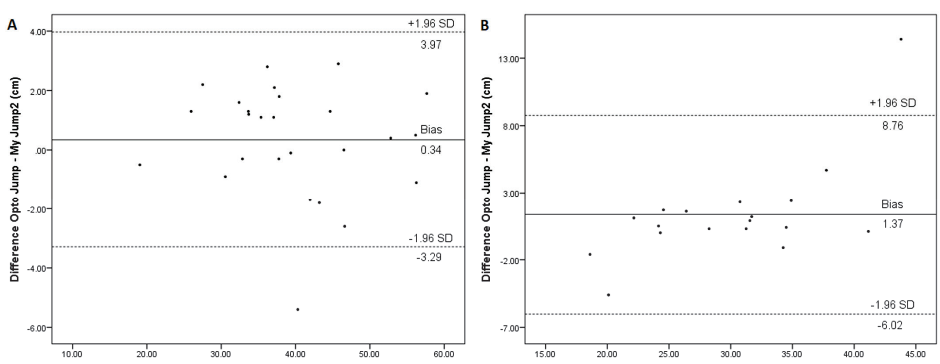

Figure 4. Level of agreement (Bland–Altman) with 95% limits of agreement (dashed lines) and the mean difference (solid line) between My Jump 2 and the Optojump for CMJAS in (**A**) male and (**B**) female participants.

4. Discussion

The CMJ and SJ tests have been strongly recommended to researchers and health practitioners. However, there is a great variety of testing methods and devices, and the majority of them are expensive and nonportable. The present study examined the concurrent validity and test-retest reliability of My Jump 2 installed on an iPhone X compared to a validated Optojump instrument for measuring jump performance during SJ, CMJ, and CMJAS in recreationally active males and females. My Jump 2 was found to be highly valid and reliable in determining the jump height of an SJ, CMJ, and CMJAS in comparison with an Optojump. Moreover, CMJ and CMJAS tests showed to be practically useful to assess and monitor vertical jump performance in recreationally active adults. Furthermore, the data presented in Bland–Altman plots (Figures 2–4), show that most of the values are close to the mean of the differences between instruments, thereby representing a high level of agreement [27]. The plot shows a systematic bias (Figures 2–4) such that, across all jump heights, values derived from the Optojump tended to be slightly higher than those from My Jump 2 app (resulting in positive difference scores). The mean bias between My Jump 2 and the Optojump for jump height was less than 0.9 cm. According to the authors knowledge, this is the first study to compare these two instruments. However, the low bias obtained in our study is in agreement with previous studies (mean bias: 0.2–1.1 cm) that compared My Jump app with force platform [15,28,29]. Higher bias (1.37 cm) was found only in females for CMJAS, which could be due to higher variability influenced by the lack of proper technique among females.

Our test-retest design in the group of recreationally active males and females revealed that SJ, CMJ, and CMJAS appear as reliable assessment outcomes (ICC > 0.90), with slightly greater variability (CV > 5%) for SJ outcomes between two sessions. The current results showed mean differences of 0.3–2.3 cm in all jumps for both males and females. This is in line with a mean difference of 0.43 cm for CMJ reported in recreationally active adults on My Jump app [22].

The concurrent validity of SJ, CMJ, and CMJAS was assessed by comparing outcome measures to the Optojump, which is already validated for estimating vertical jump. Very large correlations were observed between My Jump 2 app and Optojump in both, the male ($r = 0.95$–0.98) and female ($r = 0.94$–0.97), recreationally active adults. Most studies have compared My Jump app with force platform on several different jumps [8,15,28]. The abovementioned studies showed nearly perfect correlation ($r = 0.97$–0.99) for CMJ and SJ in trained athletes [15,28], but also for drop jumps ($r = 0.94$–0.97) in sport science students [8]. The mean differences found in previous validity studies for CMJ performance that compared portable measurement devices with force plates were between −1.06 cm and 11.7 cm [18,30,31]. Regarding the My Jump app, Gallardo-Fuentes et al. [28] found a small mean difference between devices (0.1 cm) when testing CMJ and SJ jump in both male and female athletes. In one recent study [22] on recreationally active males and females, the mean difference in CMJ between devices was 0.21 cm, which is slightly lower than the mean difference found in our study for SJ and CMJ (0.4–0.9 cm). As mentioned earlier, concurrent validity studies have compared My Jump to force plate data. However, it was also important to examine the validity of My Jump compared to a more frequently used field measurement tool. Optojump has also been found to be a valid and reliable vertical jump measurement tool [18], that is amenable to multiple testing locations and, thus, is more commonly used in different vertical jump test settings.

From a practical perspective, the use of healthy recreational adults from across the general population, iPhone X with a 240 Hz high-speed camera, the relatively large number of participants, and field-testing conditions rather than a precise laboratory space all signify strengths of the current research. However, the main limitation was that we did not use force plate, which is considered as the "gold standard" in measuring vertical jump in various populations. Nevertheless, comparing My Jump app with Optojump is more appropriate because both use the flight time to measure jump height. Additionally, different methods for determining the height of the vertical jump exist, which can also impact the validity of instruments [32]. Most of the research has compared methods that calculate jump height to methods that calculate flight distance. Struzik and Zawadzki [33] mention a method based on a force–displacement curve. The method used to calculate jump height should be determined by the equipment available and the definition of jump height used by the practitioner [34].

Furthermore, a possible limitation of our study was that some participants might not have been familiar with the SJ jump style and the usage of hands in CMJAS test, especially among female participants. Relatively high variability obtained in SJ may be due to a lack of proper technique among recreational athletes, while previous research was conducted on elite athletes [28] with greater experience performing these jumps.

Additionally, in comparison to male participants, females have a little difference in achieved jump height between CMJ and CMJAS. We can speculate that females did not swing with their arms correctly and use them to enhance their jump performance. During this jump, the arms reduce the pressure on the ground by moving downward toward the ground, which creates a negative effect, and later the arm swing creates a positive effect by moving upward and increasing the pressure on the ground [35]. Optimal jump is performed when the arms move in the jumping movement direction [36]. Additionally, female athletes show the trend for the increased differences in jump height between the two devices with increasing jumping height, which was confirmed by Attia et al. [37].

Furthermore, another limitation was that we did not check for the inter-rater reliability because some factors could contribute to differences in scores (i.e., the experience of the tester, the different variability of scores, testers' seat position, and assessment view angle) [38]. Therefore, future study should include a larger number of observers to compare results and to account for probable human error. Nevertheless, our results support the usage of smartphone apps in measuring vertical jump in recreationally active males and females. Due to its advanced technology, popularity, low cost, and portability, smartphone apps will soon be commonplace for measuring variables associated with physical fitness and health with great precision [39].

5. Conclusions

The results of present research suggest that smartphone app My Jump 2 is a valid, reliable, and useful tool for measuring jump height in recreationally active adults. Therefore, in addition to its affordable price compared with several available reference methods and given its simplicity and practicality, it can be used by practitioners, coaches, and recreationally-active adults to evaluate physical fitness with a robust and simple test as SJ, CMJ, and CMJAS.

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3

Effects of Upper-Limb, Lower-Limb, and Full-Body Compression Garments on Full Body Kinematics and Free-Throw Accuracy in Basketball Players

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Abstract: Compression garments can enhance performance and promote recovery in athletes. Different body coverage with compression garments may impose distinct effects on kinematic movement mechanics and thus basketball free-throw accuracy. The objective of this study was to examine basketball free-throw shooting accuracy, consistency and the range of motion of body joints while wearing upper-, lower- and full-body compression garments. Twenty male basketball players performed five blocks of 20 basketball free-throw shooting trials in each of the following five compression garment conditions: control-pre, top, bottom, full (top + bottom) and control-post. All conditions were randomized except pre- and post-control (the first and last conditions). Range of motion of was acquired by multiple inertial measurement units. Free-throw accuracy and the coefficient of variation were also analyzed. Players wearing upper-body or full-body compression garments had significantly improved accuracy by 4.2% and 5.9%, respectively ($p < 0.05$), but this difference was not observed with shooting consistency. Smaller range of motion of head flexion and trunk lateral bending ($p < 0.05$) was found in the upper- and full-body conditions compared to the control-pre condition. These findings suggest that an improvement in shooting accuracy could be achieved by constraining the range of motion through the use of upper-body and full-body compression garments.

Keywords: range of motion; basketball shooting; proprioception

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1. Introduction

Basketball is one of the most popular sports; at least 450 million people play basketball worldwide, ranging from registered elite players to amateurs [1]. Basketball skills can be categorized into offensive skills, including shooting, passing and dribbling and defensive skills, including blocking and stealing [2]. While shooting is the mean to score in the game, free-throws (or foul shots) are considered as one of the easiest movements, yet they can significantly influence the outcome of a game [3,4]. Movement mechanics and coordination are key to free-throwing performance [5,6] and may be regulated by wearing compression garments [7].

Compression garment can enhance performance and recovery in various sports [7,8]. Specifically, compression garments improve joint awareness, reduce muscle soreness and encourage blood circulation and thus, promote recovery [9]. Conversely, some studies have argued that upper-body compression garment may impose negative effects in hot environments and the claimed benefits may only be confined to perception of comfort [10,11]. Different movement tasks, selection of indicators, and the physical status of the athletes may also contribute to the variability and effectiveness of using compression garments during exercise, whereas garment design, such as type, coverage and tightness, may affect the functions of the garment [9]. The tightness of the compression garment has been hypothesized to change the interfacial pressure of the body [12]; however, there is a lack of studies exploring the influence of body coverage with different compression garments.

The benefits of compression garments could be attributed to the enhancement of proprioception to improve movement mechanics [13]. Hooper et al. [14] demonstrated the relationship between throwing velocity and accuracy, and improved proprioceptive signals in upper-body compression garments for baseball athletes. The compression on the cutaneous receptors or muscle spindle receptors not only enhanced the sensory information, but also filtered irrelevant mechanoreceptor information [15]. Depending on the task, the nervous system integrated these signals or information at multiple levels to mediate cutaneous and muscle afferent feedback, which is imperative for smooth coordination of movements [15–17].

There is insufficient evidence to support the use of compression garments (upper-body or lower-body) to enhance basketball performance. Atkins et al. [7] showed that wearing lower-body compression garments overnight produced negligible effects on the countermovement jump, repeated sprint and agility test performances, despite improvements in perceived fatigue and muscle soreness. Other evidence indicated that lower-leg compression garments were found to significantly reduce the range of abduction motion of the hip joint during a drop vertical jump, but produced minimal effects on the kinematics/kinetics of other lower extremity joints [13].

Furthermore, lower-body compression was shown to improve lower limb balance and stability in active females during a single-leg balance task [18]. Poor stability results in higher motion variability and may potentially weaken shooting accuracy [6,19,20]. How these findings affect other functional performances (e.g., basketball shooting) requires further investigation. Since compression garments produce mechanical restraints on body segments and joints, range of motion (ROM) has been one of the key parameters for the evaluation of kinematic effects during exercise in previous basketball studies [13,21].

Considering the relationship between compression garment coverage (upper-body, lower-body and combined) on the kinematics and shooting performance of basketball specific maneuvers is currently questionable, coaches and athletes are eager to understand what type of compression garment coverage could help them improve performance and consistency of performance. The objective of this study was to examine the effect of upper- and lower-body compression garment coverage (top, bottom and full) on the full body range of motion (ROM) and shooting accuracy of basketball free-throws. It was hypothesized that a certain compression garment condition would improve free-throw performance and consistency compared to the no-compression garment control group.

2. Materials and Methods

2.1. Participants

Twenty ($n = 20$) male basketball players were recruited from local universities. Their average age, height and body mass were 22.6 ± 1.1 years, 179.4 ± 3.4 cm and 72.7 ± 8.2 kg, respectively. All participants had at least 4 years of experiences in playing basketball and were right hand dominant single-handed shooters. The average basketball training experience and training time were 8.5 ± 2.4 years and 5.2 ± 1.6 h per week, respectively. All participants were physically fit and healthy and reported no injuries over the previous 6 months. Ethical approval (IRB-2017-BM-006) was granted from the institutional ethics committee. Written informed consent was obtained from all participants.

2.2. Experimental Conditions and Procedure

All free-throw shooting conditions were performed in our biomechanical laboratory. The free-throw distance and the height of the basketball rim were set according to the International Basketball Federation standards [19]. The participants performed single-handed free-throws under five different garment conditions, control-pre: no garment pre-control, Top: upper-body compression garment (Li Ning, Powershell, AULM043-I, Beijing, China), Bottom: lower-body compression garment bottom (Li Ning, Powershell, AUDL101-1, Beijing, China), full: both upper-body and lower-body compression garment and control-post: no garment post-control, as shown in Figure 1. Control-pre and control-post were the first and the last test conditions. The remaining three compression garment conditions (top, bottom and full) were randomly assigned as the second to the fourth conditions across participants. As the experimental protocol compared the first and last conditions, we were able to evaluate the fatigue effect [22]. For each free-throw condition, 20 free-throw shooting trials were performed. Testing of the next condition started immediately after the participant changed their garments.

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

Figure 1. Compression garment conditions: (**a**) top; (**b**) bottom; (**c**) full (top + bottom).

The control conditions (control-pre and control-post) were self-selected comfortable sportswear that were not compression garments. The experimenters measured the height, waist and chest circumference of the participants to determine the appropriate garment [23]. The appropriate compression garment size was pre-determined by the manufacturer's sizing guidelines and was based on the body height and mass of each participant. Next, we assigned participants compression garments one size smaller than the pre-determined appropriate size in order to increase the interfacial pressure, as recommended by the experimental protocol detailed by Williams and colleagues [12].

A motion capturing system with multiple inertial measurement units (MyoMOTION, Noraxon, Inc., Scottsdale, AZ, USA) was used to measure full-body kinematics during the free-throw shooting trials. The inertial measurement units (IMU) were attached and strapped to each body segment according to the instrument guidelines. During each free-throw trial, the participants performed shooting from the same position behind the free-throw line. The sampling frequency of the IMU was 200 Hz. The kinematic data during the free-throw motion were post-processed using Matlab software (MathWorks, Inc., Natick, MA, USA) using a 6 Hz cutoff 4th order Butterworth low-pass filter.

2.3. Outcome Measures

Outcome measures including performance score (accuracy) and joint ROM variables were investigated. The performance score was gauged using an ordinal six-point (0 to 5 point) scoring system. Five, four and three points denoted a clean score, that the ball hit the rim and went in, and that the ball hit the backboard and went in, respectively. Two, one and zero points denoted that the ball hit the rim and missed, hit the backboard and missed and missed complete, respectively, as illustrated in Table 1 [19,24]. The consistency of the score was also assessed by the coefficient of variation (i.e., the ratio of the standard deviation to the mean of the trials).

Table 1. The six-point basketball shooting performance score system.

Performance	Scored			Missed		
	Clean (Swish)	**Rim & In**	**Backboard & In**	**Rim & Out**	**Backboard & Out**	**Complete Miss**
Score	5	4	3	2	1	0

ROM of the head, trunk, elbow, shoulder, wrist, hip, knee and ankle joints in the sagittal, coronal and frontal planes were calculated. Data were averaged across trials for each participant in each condition which served as the targeted average profile for subsequent statistical analysis [25]. We did not view the within-participant effect (trial) of ROM as an independent observation or random factor to be analyzed.

2.4. Data Analysis

All statistical analysis was performed in SPSS 21 (IBM, New York, NY, USA). Prior to statistical analysis, the Shapiro–Wilk test was performed to check for the normality of the kinematic data, and it was satisfied. The Wilcoxon signed-rank test was performed to compare free-throw performance scores between the control-pre- and control-post-control conditions to ensure that there was no learning or fatigue effect (i.e., Control pre- and post-control were not significantly different). Furthermore, one-way repeated measures analysis of variance (ANOVA) was performed to examine any significant difference for joint ROM variables between the control-pre, top, bottom and full conditions, followed by the post hoc pairwise comparison of Least Significant Difference (LSD) if a significant main effect was found. We chose the LSD approach as our research hypothesis was more focused on planned comparisons. As such, we regarded the ANOVA as an additional constraint [26]. Similarly, the comparison for the performance score and the coefficient of variation was performed using a nonparametric test (Friedman test), with the post hoc pairwise Wilcoxon signed-rank test, as the performance score was gauged in an ordinal scale. Level of significance was set at $p = 0.05$. The indices of effect size for the ANOVA and post hoc pairwise comparison were partial η^2 and Cohen's d, respectively.

3. Results

3.1. Control-Pre and Control-Post Conditions

There was no significant difference in performance score between the control-pre (Median = 2.975) and control-post (Median = 3.075) conditions ($Z = -1.430$, $p = 0.153$). Similarly, there was no significant difference in the coefficient of variation of performance score between the control-pre and control-post conditions ($Z = -1.382$, $p = 0.167$). We assumed that there was no pronounced carry-over or fatigue effect that significantly affected performance over the course of the experiment.

3.2. Free-Throw Accuracy

There were no significant differences in free-throw performance score ($\chi^2(4) = 6.510$, $p = 0.089$) or the coefficient of variation of the performance score ($\chi^2(4) = 5.629$, $p = 0.131$) between the conditions (control-pre, top, bottom or full). However, post hoc pairwise comparison showed that the free-throw performance scores of the top (Median = 3.1, $Z = -2.357$, $p = 0.018$) and full (Median = 3.15, $Z = -2.112$, $p = 0.035$) conditions were significantly larger than that of the control-pre condition (Median = 2.975), as shown in Tables 2 and 3.

Table 2. Descriptive statistics of the averaged and coefficient of variation of the free-throw performance score.

Condition	Performance Score		Coefficient of Variation (%) Mean (Standard Deviation)
	Median	Mean (Standard Deviation)	
Control-Pre	2.975	2.975 (0.419)	38.04 (6.67)
Top	3.100 *	3.168 (0.382)	36.78 (7.05)
Bottom	3.050	3.035 (0.411)	37.06 (7.07)
Full	3.150 *	3.175 (0.385)	36.19 (7.58)
Control-Post	3.075	3.123 (0.476)	35.88 (8.61)

* significant difference ($p < 0.05$) compared to the control-pre condition by post hoc Wilcoxon signed-rank test.

Table 3. Probability values (p-value) of the average (upper right triangle) and coefficient of variation (lower left triangle) of the free-throw performance score.

Coefficient of Variation	Performance Score			
	Control-Pre	Top	Bottom	Full
Control-Pre		0.018 *	0.230	0.035 *
Top	0.296		0.152	0.888
Bottom	0.227	0.654		0.159
Full	0.107	0.794	0.344	

* significant difference ($p < 0.05$) by post hoc Wilcoxon signed-rank test.

3.3. Full-Body Joint Range of Motion (RoM)

One-way ANOVA repeated measures showed that the variation in compression garments imposed significant effects on the ROM of head flexion ($p = 0.014$, partial $\eta^2 = 0.169$), trunk lateral bending ($p = 0.024$, partial $\eta^2 = 0.152$), left shoulder flexion ($p = 0.041$, partial $\eta^2 = 0.152$), right shoulder rotation ($p = 0.048$, partial $\eta^2 = 0.128$) and left knee flexion ($p = 0.003$, partial $\eta^2 = 0.212$). Post hoc pairwise comparison showed that the top condition significantly reduced the head flexion ($p = 0.037$; d = 0.503; 1.346, 95% CI 0.376 to 2.315) and trunk lateral bending ($p = 0.042$; d = 0.487; 1.039, 95% CI 0.041 to 2.036) ROM compared with the control-pre condition (Table 4). Similarly, the full condition significantly reduced head flexion ($p = 0.009$; d = 0.650; 1.346, 95% CI 0.376 to 2.315) and trunk lateral bending ($p = 0.028$; d = 0.532; 1.446, 95% CI 0.173 to 2.718) ROM compared to the control-pre condition.

Table 4. Descriptive statistics and one-way ANOVA repeated measures outcome of the range of motion of head and trunk in different compression garment conditions.

	Range of Motion, Mean (Standard Deviation)				ANOVA Repeated Measure	
	Control-Pre	Top	Bottom	Full	Effect Size	p-Value
Head FL/EX	10.57 (3.81)	9.53 (3.1) ^{a}	9.75 (3.37)	9.22 (3.07) ^{A}	0.169	0.014 *
Head lateral bending	6.14 (2.83)	5.80 (2.69)	6.05 (2.99)	5.87 (2.62)	0.019 ^{g}	0.694
Head axial rotation	13.17 (8.04)	17.11 (12.74)	15.02 (10.49)	14.42 (8.63)	0.053	0.368
Trunk FL/EX	19.20 (6.24)	17.15 (5.96)	18.43 (5.85)	18.15 (6.42)	0.11	0.082
Trunk lateral bending	10.21 (4.24)	9.17 (4.38) ^{a}	9.88 (3.63)	8.77 (4.01) ^{a}	0.152	0.024 *
Trunk axial rotation	11.05 (4.56)	11.46 (5.04)	10.99 (4.36)	11.39 (4.37)	0.018 ^{g}	0.687

FL/EX: flexion/extension; * significant difference ($p < 0.05$) using one-way ANOVA repeated measures; ^{g} Greenhouse–Geisser correction to adjust the lack of sphericity; ^{a} and ^{A} denote $p < 0.05$ and $p < 0.0125$ than the control-pre condition.

Compared to that of the bottom condition, both the top ($p = 0.01$; d = 0.642; 3.422, 95% CI 0.929 to 5.915) and full ($p = 0.003$; d = 0.778; 3.530, 95% CI 1.405 to 5.655) conditions significantly reduced the ROM of the left shoulder flexion, while the top condition had significantly larger right shoulder rotation compared with the control-pre ($p = 0.013$; d = 0.611; 38.316, 95% CI −8.98 to 67.65) and bottom ($p = 0.041$; d = 0.491; 23.028, 95% CI 1.08 to 44.976) conditions (Table 5). The control-pre condition had significantly larger left knee flexion ROM than the bottom (p = 0.026; d = 0.539; 2.605, 95% CI 0.345 to 4.864) and full (p = 0.002; d = 0.804; 2.908, 95% CI 1.214 to 4.602) conditions. Similarly, the top condition had a significantly larger left knee flexion ROM than the bottom (p = 0.044; d = 0.482; 2.047, 95% CI 0.059 to 4.035) and full (p = 0.018; d = 0.585; 2.351, 95% CI, 0.469 to 4.232) conditions (Table 6).

Table 5. Descriptive statistics and one-way ANOVA repeated measures outcome of the range of motion of the upper limb in different compression garment conditions.

	Range of Motion, Mean (Standard Deviation)				ANOVA Repeated Measure	
	Control-Pre	Top	Bottom	Full	Effect Size	p-Value
L elbow FL/EX	49.35 (23.12)	51.38 (23.28)	50.95 (22.82)	51.24 (24.05)	0.048	0.417
R elbow FL/EX	93.30 (13.27)	89.43 (12.46)	89.14 (13.79)	89.82 (14.38)	0.098	0.116
L shoulder FL/EX	30.06 (12.77)	26.98 (11.58)	30.40 (12.66) [B]	26.87 (10.48) [C]	0.152 [g]	0.041 *
R shoulder FL/EX	44.11 (18.88)	40.48 (16.96)	44.01 (19.11)	41.47 (17.2)	0.148 [g]	0.148
L shoulder AB/AD	124.84 (113.81)	125.37 (132.49)	133.85 (120.05)	132.36 (124.17)	0.022 [g]	0.66
R shoulder AB/AD	72.93 (51)	75.01 (55.07)	76.05 (48.47)	78.02 (71.69)	0.012 [g]	0.821
L shoulder rotation	50.50 (40.81)	58.41 (51.4)	67.74 (83.18)	59.95 (61.04)	0.089 [g]	0.176
R shoulder rotation	90.38 (45.46)	128.70 (81.71) [a]	105.67 (67.28) [b]	118.07 (73.94)	0.128	0.048 *
L wrist RA/UL	35.80 (26.08)	34.93 (27)	35.12 (28.48)	38.26 (32.93)	0.036	0.552
R wrist RA/UL	70.79 (27.55)	79.36 (29.02)	71.34 (33.37)	77.37 (31.8)	0.071 [g]	0.249
L wrist FL/EX	39.21 (33.56)	39.37 (41.11)	42.20 (43.93)	42.90 (45.34)	0.011 [g]	0.885
R wrist FL/EX	105.39 (34.39)	109.35 (36.59)	110.83 (35.36)	106.85 (35.66)	0.026	0.675
L palm rotation	49.54 (47.21)	51.35 (48.84)	60.70 (72.13)	52.23 (55.49)	0.067 [g]	0.269
R palm rotation	93.97 (46)	126.74 (77.75)	113.01 (81.54)	110.93 (66.53)	0.117	0.066

FL/EX: flexion/extension; AB/AD: abduction/adduction; RA/UL: Radial/Ulnar deviation; * significant difference ($p < 0.05$) using one-way ANOVA repeated measures; [g] Greenhouse–Geisser correction to adjust the lack of sphericity; [a] denotes $p < 0.05$ than the control-pre condition; [b] and [B] denote $p < 0.05$ and $p < 0.0125$ than the top condition; [C] denotes $p < 0.0125$ than the bottom condition.

Table 6. Descriptive statistics and one-way ANOVA repeated measures outcome of the range of motion of the lower limb in different compression garment conditions.

	Range of Motion, Mean (Standard Deviation)				ANOVA Repeated Measure	
	Control-Pre	Top	Bottom	Full	Effect Size	p-Value
L hip FL/EX	23.56 (6.8)	22.58 (6.65)	22.34 (7.8)	21.81 (7.19)	0.076 [g]	0.22
R hip FL/EX	26.47 (4.49)	25.14 (6.02)	26.19 (6.26)	25.93 (5.85)	0.065	0.274
L hip AB/AD	5.69 (1.73)	6.46 (2.38)	5.57 (1.64)	6.28 (2.36)	0.069	0.251
R hip AB/AD	6.95 (2.6)	7.22 (2.72)	6.31 (2.42)	6.97 (2.62)	0.069	0.248
L hip rotation	9.99 (3.57)	10.03 (3.52)	9.31 (3.04)	9.45 (2.53)	0.043 [g]	0.445
R hip rotation	12.62 (3.89)	12.83 (4.19)	12.43 (4.17)	12.30 (4.61)	0.013	0.861
L knee FL/EX	51.86 (8.63)	51.30 (8.15)	49.25 (10.08) [a, b]	48.95 (9.37) [A, b]	0.212	0.003 *
R knee FL/EX	53.77 (7.33)	52.90 (7.88)	53.61 (6.61)	52.94 (6.67)	0.036	0.549
L knee rotation	10.64 (4.39)	11.35 (5.59)	10.26 (3.99)	10.35 (4.21)	0.051 [g]	0.37
R knee rotation	14.96 (4.52)	14.72 (6.12)	15.68 (5.86)	15.43 (5.01)	0.031	0.61
L knee AB/AD	7.15 (4.52)	7.92 (4.46)	6.81 (2.89)	7.00 (3.28)	0.029 [g]	0.587
R knee AB/AD	8.32 (3.72)	7.67 (3.84)	9.10 (4.32)	8.48 (3.58)	0.048	0.418
L ankle PL/DO	61.22 (16.82)	64.36 (8.64)	61.71 (8.01)	61.83 (9.08)	0.049 [g]	0.362
R ankle PL/DO	60.93 (11.16)	61.97 (6.1)	60.76 (7.49)	61.82 (8.21)	0.019 [g]	0.682
L ankle EV/IV	25.31 (14.66)	22.84 (13.08)	24.29 (11.37)	23.38 (12.54)	0.041	0.49
R ankle EV/IV	26.49 (13.19)	23.04 (11.28)	22.95 (8.38)	21.27 (10.34)	0.123	0.056
L ankle AB/AD	15.28 (4.08)	15.47 (4.5)	15.52 (4.76)	16.12 (5.18)	0.016 [g]	0.732
R ankle AB/AD	13.98 (3.73)	14.69 (4.46)	15.17 (4.9)	14.43 (4.14)	0.077	0.204

FL/EX: flexion/extension; AB/AD: abduction/adduction; EV/IV: eversion/inversion; PL/DO: plantarflexion/dorsiflexion; * significant difference ($p < 0.05$) using one-way ANOVA repeated measures; [g] Greenhouse–Geisser correction to adjust the lack of sphericity; [a] and [A] denote $p < 0.05$ and $p < 0.0125$ than the control-pre condition; [b] denotes $p < 0.05$ than the top condition.

4. Discussion

This study examined the effect of upper and lower-body compression garments on the body kinematics and shooting accuracy of basketball free-throws. Our study found that upper-body (top) or full-body (top + bottom) compression garments significantly improved the performance of basketball free-throws; however, there was no significant improvement in the consistency of performance. Overall, mechanically, compression garments had a significant influence on the ROM of the head flexion, trunk lateral bending, left (non-dominant side) shoulder flexion, right (dominant side) shoulder rotation and left knee flexion as indicated by the ANOVA findings. Post hoc comparisons showed that wearing either upper- or full-body garments constrained the ROM of head flexion and trunk lateral bending which could be associated with improved trunk stability and thus, improved performance [27]. The relationship between the condition of the head movement and stability and free-throw accuracy was advocated previously, but not well understood [28]. On the other hand, garment coverage of the lower body (bottom or full-body gear) significantly reduced the ROM of the left (non-dominant) side knee joint in the sagittal plane, but not the right (dominant) side, because experienced players tended to adjust the knee joint of the dominant side to greater extent for better performance [29]. Theoretically, compression of the knee joint enhanced proprioception and thus performance [30,31] notwithstanding that our study did not demonstrate an improved shooting score for lower-body (bottom) garments. In addition, the reduced head flexion and trunk lateral bending ROM could implicate successful shooting performance.

Elbow and wrist movements are determinants of free-throw performance and player skill levels [20]. Skilled players coordinate the shooting arm by constantly compromising between elbow and wrist movements to adapt to subtle changes in release parameters of the ball (e.g., release height, angle of ball projection, velocity at ball release) [20]. In addition, more highly skilled players tend to maximize the ROM of the wrist joint [20]. top compression garments help to constrain the ROM of the elbow, and thus players can focus on optimizing distal joint (wrist) motion only [20]. In our study, although there were no significant main effects on the ROM of the elbow and wrist joints, pairwise comparisons showed that upper-body (top) garments significantly reduced the ROM of the right (dominant) side elbow, but increased that of the wrist radial/ulnar deviation and palmar rotation compared to that of the control-pre condition. This was likely due to the fact that the uncovered wrist joint compensated the reduced motion of the elbow [20]. In fact, some statisticians argued that conducting and interpreting post hoc analyses could still be valid even though the main effect was not significant [32,33].

The enhanced proprioception by compression garments may also facilitate the organization of compensatory behavior between joints for better performance. This was supported by existing studies that the proprioception (joint position sense) of the elbow and wrist joints was correlated with the success rate of the free-throw tasks [34]. More highly skilled players managed to optimize their performance based on the perceptual consequence of their actions [35].

A previous study suggested that the shoulder joint plays an important role in the action of basketball free-throws. Kaya et al. [36] found that free-throw performance was significantly correlated with the peak torque of the shoulder joint muscles and the shoulder joint position sense at 160° in the dominant side. While we anticipated that compression garments would amplify the proprioception [30], enhance stability and reduce the ROM of the shooting limb (right side), our study found that the ROM of the upper-body was significantly smaller when wearing top compression garments than when wearing bottom garments. Although there were no significant differences compared to that of the control-pre condition, we believe that the increased trend of the joint ROM may indicate that wearing lower-body (bottom) garments alone had a negative effect on the shoulder joint. From the kinetic chain perspective, intervention at the lower limb level may alter energy generation which can be transferred to the upper limbs and thus considerably influences upper limb movement tasks (e.g., racket and ball speed in racket sports) [19,37]. The influence of lower limb garments on the upper limbs may also be the reason that the full-body garments did not have an effect on the elbow and wrist joints, despite upper-body garments having an effect.

There were some limitations in this study. First, although we demonstrated no carry-over effect as revealed by the fact that there was no significant difference between the performance score of the control-pre and control-post conditions, there was an improvement trend on both the performance score and consistency. We believed that the randomized order assigned on the garment condition could minimize the carry-over effect. Second, our short adaptation time for each compression garment condition may not be adequate enough, despite that there is no consensus on the duration of adaptation in the past studies. Future studies may consider tests with longer adaptation in different days or weeks or considering the variation of kinematic variables [38]. Third, we presented only joint ROM in this study. More comprehensive analysis with discrete variables (peak angle, angular velocity), joint power, muscle force, proprioception as well as stability should be considered to evaluate their influence and underlying mechanism on the free-throw shooting performance. Asymmetry sport activity (e.g., single-handed shooting) may produce unique sequential coordination of the upper and lower limb with coherent patterns of muscle activation [39]. Forth, our study confined to non-professional basketball players. Playing level and sex effects may contribute to variations in movement strategy, skeletal alignment and muscle strength and could also be investigated. Lastly, the compression garments may impose different levels of pressure on the participants depending on their body built. Future study shall consider measuring the compression level in each condition.

5. Conclusions

Players wearing upper-body or full-body compression garment significantly improved basketball free-throw accuracy by 4.2% and 5.9%, respectively, but not on the intertrial consistency. full body kinematics data suggested that the improved performance could be attributed to the reduced ROM of head flexion and lateral bending of the trunk. Future studies investigating the relationship between shooting performance in basketball, reduced ROM and enhanced proprioception or stability are required.

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4

The Characteristics of Feet Center of Pressure Trajectory during Quiet Standing

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Abstract: To investigate the level of bilateral symmetry or asymmetry between right and left foot center of pressure (COP) trajectory in the mediolateral and anteroposterior directions, this study involved 102 participants (54 females and 48 males). Ground reaction forces were measured using two Kistler force plates during two 45-s quiet standing trials. Comparisons of COP trajectory were performed by correlation and scatter plot analysis. Strong and very strong positive correlations (from 0.6 to 1.0) were observed between right and left foot anteroposterior COP displacement trajectory in 91 participants; 11 individuals presented weak or negative correlations. In the mediolateral direction, moderate and strong negative correlations (from −0.5 to −1.0) were observed in 69 participants, weak negative or weak positive correlations in 30 individuals, and three showed strong positive correlations (0.6 to 1.0). Additional investigation is warranted to compare COP trajectories between asymptotic individuals as assessed herein (to determine normative data) and those with foot or leg symptoms to better understand the causes of COP asymmetry and aid clinicians with the diagnosis of posture-related pathologies.

Keywords: symmetry; asymmetry; foot; force; balance; postural stability; standing

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1. Introduction

The force exerted by the body on the ground when standing is mirrored by a reaction force. The use of footwear during many activities of daily living not only provides a level of protection but also modifies the pressure distribution characteristics of the feet and, therefore, the forces that act on the foot. Depending on the type of footwear, these forces may be attenuated or "dampened". Additionally, the shape and construction of the sole, insole and heel may all modify the forces and load experienced by different foot regions [1–3].

From a mechanical perspective, balance preservation during upright standing is quite complex as the human body is never in a condition of perfect equilibrium and that balance must be maintained via two points of contact (both feet). The center of all external forces acting on the plantar surface of the foot is known as the center of pressure (COP). While providing a base of support, the feet can independently induce changes in COP trajectory in the coronal (anteroposterior direction) and sagittal (mediolateral direction) planes and, therefore, obfuscate right and left COP to create a condition of asymmetry. While COP displacement in the anteroposterior direction can be understood as bilaterally symmetrical when the trajectory (forward or backward) is concurrently equal between both feet, the level of bilateral symmetry in the mediolateral direction is far more problematic to measure. One possible method for assessing bilateral symmetry in mediolateral COP trajectory is by considering foot structure. Albeit an oversimplification, this paradigm allows us to treat COP displacement as symmetrical when left and right foot mediolateral COP both shift either to the inside or outside of the feet.

The literature is abound with studies investigating balance preservation during quiet standing primarily by analyzing COP-related variables. However, the vast majority is based on using a single force plate and thus measure the exerted force concurrently for both feet [4–10]. Few investigations have addressed the magnitude and distribution of force separately for the right and left foot with the use of more than one force platform. Although Soangra and Lockhart [11] and Brauer et al. [12] investigated the similarities and dissimilarities between right and left foot COP trajectories, no studies have yet addressed COP displacement in regards to the level of symmetry (or asymmetry) between the left and right foot. As COP has been identified as a measure of the neuromuscular response to maintain balance, differences in right and left foot COP trajectories can serve as a measure of sensorimotor control and function. In this way, the respective COP trajectories for either foot and the congruence between both points of application signify the ability of the central nervous system to integrate information from the sensory systems and then activate different postural muscles (exerting pressure at a specific foot region) so that upright stance is preserved.

While the foot can be assumed to hold two degrees of freedom relative to the lateral gastrocnemius (plantarflexion/dorsiflexion and inversion/eversion), COP displacement can independently shift not only in the anteroposterior and mediolateral directions but also combinations of the two, such as an anterior or posterior slant to either the medial or lateral side. If analyzed as a temporal series, this bivariate approach for each foot could help identify patterns in left and right foot COP trajectory. Knowledge of COP displacement between both feet across different population cohorts can help identify postural pathologies including foot deformities resulting from improper footwear or decreased neuromuscular control by some deficiency in central nervous system function.

Therefore, the purpose of the study was to define the incidence of right and left foot symmetry or asymmetry via displacement in COP trajectory by identifying what associations exist between left and right foot COP direction in an asymptomatic population. By knowing the value of the symmetry between left and right foot COP displacement, it could be possible to formulate criteria for evaluating postural balance during upright standing. We hypothesized that the temporal and spatial characteristics of right and left foot COP trajectories during upright stance would show little variability in a sample of healthy young adults.

2. Material and Methods

2.1. Subject

An age-homogeneous sample of 102 university students (54 females and 48 males) was recruited. Mean age was 21.08 ± 1.08 years, height 172.89 ± 9.56 cm and body mass 68.09 ± 13.12 kg. All individuals provided their written informed consent to participate in the study and ethical approval was obtained. All procedures were performed at a biomechanics research laboratory at the Opole University of Technology in Opole, Poland. The participants signed informed consent and were informed of the protocol and procedures for the experiment prior to the exercise. The study was approved by the Human Ethics Committee.

2.2. Ground Reaction Force (GRF) Measurement Procedure

The study protocol involved measuring ground reaction force during two 45-s trials of static standing. Ground reaction force (GRF) data were synchronously collected on two 600 × 400 mm piezoelectric force platforms (Kistler Type 9286B; Kistler Instruments AG, Winterthur, Switzerland) placed under each foot. Four tri-axial force sensors located in the corners of each platform quantified the ground reaction force signals at a sampling frequency of 50 Hz (measurement range was from 10 kN to 20 kN). The force platforms were calibrated before use and integrated with a base transceiver station (BTS) Smart optoelectronic system (BTS Bioengineering, USA) to register the force–time characteristics.

During signal acquisition, the participant was asked to assume a relaxed upright posture (minimizing head and trunk movements) with the upper extremities resting freely against the trunk and to fixate on a point placed at eye level 3 m from the subject. Only running shoes were allowed, and the participant stood with their feet completely parallel (no ankle rotation with a 30 cm stance width).

2.3. COP Measurement

GRF signals were recorded 10 s after trial commencement for 45 s. The two trials were executed one after the other with no change in foot position. The Bioware software automatically calculated *x*- and *y*-axis COP location (in mm) separately for right and left foot using the equations:

$$\text{Ax} = (\text{Fx*az0} - \text{My})/\text{Fz} = -\text{My}'/\text{Fz} \tag{1}$$

where:

Ax	path of the lateral direction of the force component
Fx	magnitude of the component of the ground reaction force, caused by the pressure of the foot acting on the left to the right direction
az0	constant value for a given measuring instrument (see Kistler platform) determining the distance of the piezoelectric sensor from the platform level
My	the moment of the component of the ground reaction force, caused by the pressure of the foot acting in the forward–backward direction
Fz	the magnitude of the component of the ground reaction force, caused by the pressure of the foot acting vertically

$$\text{Ay} = (\text{Fy*az0} + \text{Mx})/\text{Fz} = \text{Mx}'/\text{Fz}, \tag{2}$$

where:

Ay	way of action of force component back and forth
Fx	the magnitude of the component of the ground reaction force, caused by the pressure of the feet acting on the left to the right direction
az0	constant value for a given measuring instrument (see Kistler platform) determining the distance of the piezoelectric sensor from the platform level
Mx	the moment of the component of the ground reaction force, caused by the pressure of the foot acting in the direction to the left to the right
Fz	the magnitude of the component of the ground reaction force, caused by vertical pressure of the feet

In order to aid the quantification of COP trajectory as a function of time, the initial COP location was shifted to the mean COP location (calculated for each trial).

A mathematical coordinate system was used to present the test results. The *x*- and *y*-axes show the direction of action of ground reaction forces caused by foot pressure on the ground. This is called phase plane (ground). The *x*-axis represents the action of these forces in the left and right directions. The *y*-axis represents the action of these forces in the forward and reverse directions. In the context of these axes, anterior foot pressure means that the subject transfers weight to the toes and hind toes. There are people who transfer weight to the foot in opposite directions: outside, to the left foot to the left and to the right foot to the right; or inwards, to the left foot to the right and to the right foot to the left. Hence the term "mid-lateral".

The COP data (2250 measures per trial) were then plotted as a statokinesigram for COP spatial trajectory in the mediolateral and anteroposterior directions as well as a stabilogram showing the temporal domain of COP in both directions. An example of the COP trajectory as a stabilogram and statokinesigram is illustrated in Figures 1 and 2, respectively.

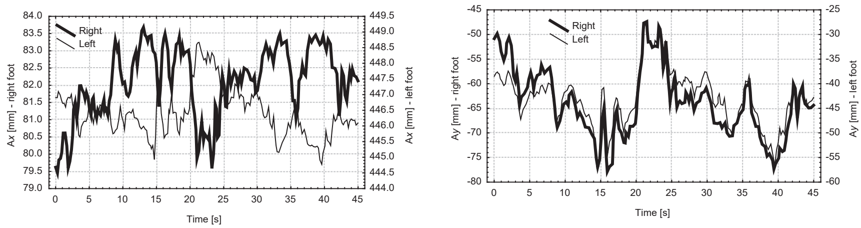

Figure 1. Exemplary stabilogram of center of pressure (COP) time series data in the mediolateral (Ax) and anteroposterior (Ay) directions.

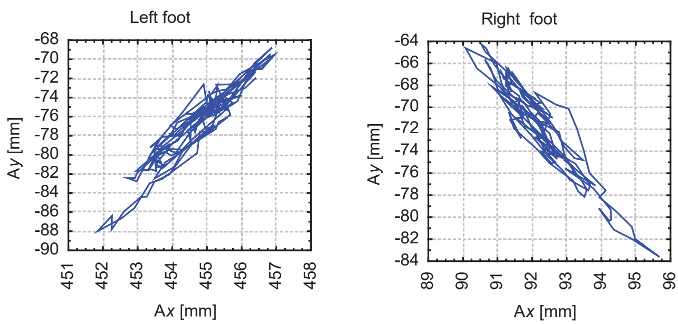

Figure 2. Exemplary statokinesigram of COP spatial data in the mediolateral (Ax) and anteroposterior (Ay) direction.

The selection of a right COP shift trajectory of the right and left foot may illustrate the motor abilities of the subject. The human may determine and choose a right COP displacement trajectory for the right or left foot. Namely, the human may decide on part of a foot which will put pressure on the surface.

2.4. Statistical Analysis

Basic descriptive statistics were calculated (means ± standard deviations). The Shapiro–Wilk test was used to determine if the data set was well-modeled by a normal distribution. Differences between the obtained values were assessed using Student's *t*-test. The level of symmetry or asymmetry in COP displacement trajectory between the right and left foot in both the mediolateral and anteroposterior directions was assessed with Pearson's correlation coefficients. Correlations between right and left foot COP trajectory were independently calculated in the mediolateral and anteroposterior directions for each participant. Additionally, the correlation coefficients of COP trajectory between the mediolateral and anteroposterior directions were also calculated independently for the right and left foot. The level of significance was set at $\alpha = 0.05$. All data processing was performed with the Statistica 10.0 software package.

3. Results

The assumption of normality was confirmed, indicating a normal distribution. As no significant between-sex differences for COP displacement trajectory were found, the data were analyzed for the entire sample ($n = 102$). Furthermore, *t*-tests revealed no significant differences between the first and second trial, hence analysis involved only data from the first trial.

The correlations between right and left foot COP trajectory in the mediolateral and anteroposterior directions are presented as a histogram (Figure 3). Strong and very strong positive correlations (0.6 to 1.0) for right and left foot COP displacement trajectory in the mediolateral direction were observed in

91 participants while 11 individuals presented weak or negative correlations. In the anteroposterior direction, moderate and strong negative correlations (−0.5 to −1.0) were observed in 69 participants while 30 were found with weak negative or positive correlations. Strong positive correlations (0.6 to 1.0) in the mediolateral direction were observed in only three individuals.

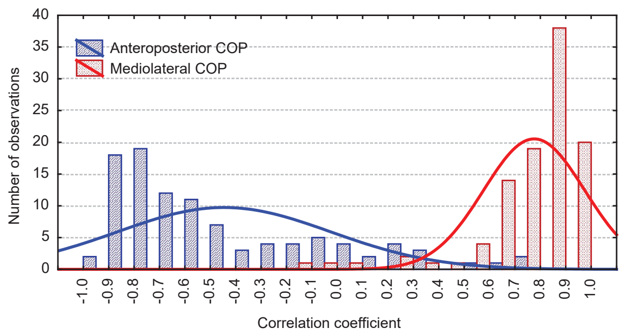

Figure 3. Histogram of correlation coefficients between right and left foot COP trajectory in the mediolateral and anteroposterior directions.

In order to better illustrate the correlations of COP trajectory between the mediolateral and anteroposterior directions, a scatter plot was generated in which the correlations were plotted for each participant (Figure 4). The *x*-axis was used to delineate the right foot correlations between COP trajectory directions whereas the *y*-axis represented the left foot. Based on this structure, four quadrants were defined.

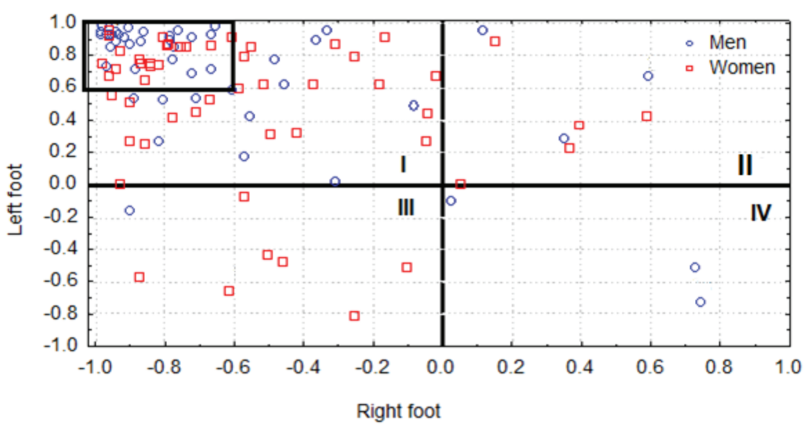

Figure 4. Scatter plot illustrating the correlations between right (*x*-axis) and left foot (*y*-axis) mediolateral and anteroposterior COP trajectories for each participant.

The first quadrant (Quadrant I) contains the participants (42 females and 41 males) presenting positive left foot and negative right foot correlations. The second quadrant (Quadrant II) represents those participants (five females and three males) with a positive correlation between the mediolateral and anteroposterior directions in both the right and left foot The third quadrant (Quadrant III), in turn, contains those participants (seven females and one male) with negative right and left foot correlations in both directions whereas the fourth quadrant (Quadrant IV) entails the small group of participants (three males) with positive right foot and negative left foot correlations.

The majority of the sample (n = 83) was grouped in the second quadrant (Quadrant II). A statokinesigram representative of this group was extracted from Participant 1 (Figure 5), who was found with a correlation coefficient of 0.94 for the left and −0.94 for the right foot. Furthermore, 57 participants (more than half the sample) in this quadrant presented strong correlations in both the left (r = 0.6 to 1.0) and right (r = −1.0 to −0.6) feet. A box was drawn in Figure 4 to highlight this congregation.

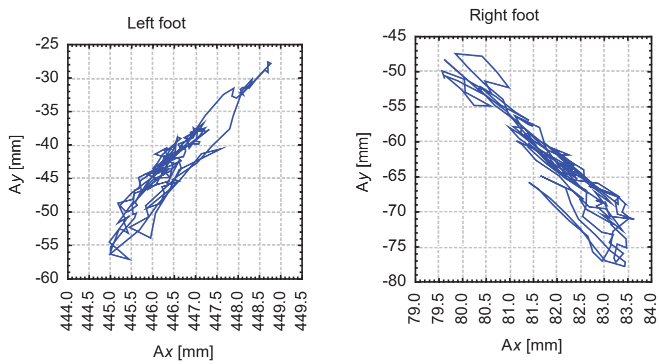

Figure 5. Stabilogram of right and left foot COP trajectories in the mediolateral (Ax) and anteroposterior (Ay) directions for Participant 1 (Quadrant II).

On the opposite spectrum, the three individuals composing the fourth quadrant (Quadrant IV) can be characterized by the statokinesigram of Participant 13 (Figure 6), who was found to present a correlation coefficient of −0.72 for the left and 0.74 for the right foot. The third quadrant (Quadrant III) can be represented by the statokinesigram of Participant 72 (Figure 7), with correlation coefficients of −0.65 for the left and 0.62 for the right foot. An exemplary statokinesigram of the participants located in the first quadrant (Quadrant I) is provided by Participant 93 (Figure 8), with correlation coefficients of 0.67 for the left and 0.59 for the right foot.

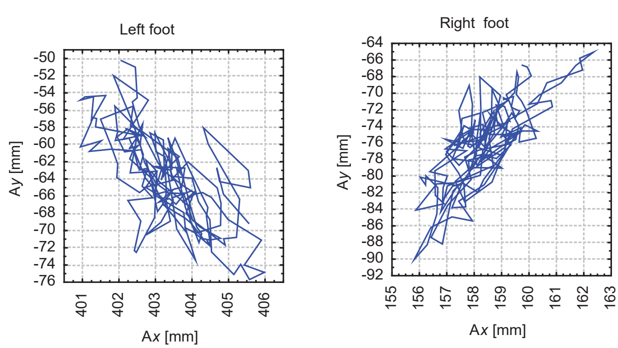

Figure 6. Stabilogram of right and left foot COP trajectories in the mediolateral (Ax) and anteroposterior (Ay) directions for Participant 13 (Quadrant IV).

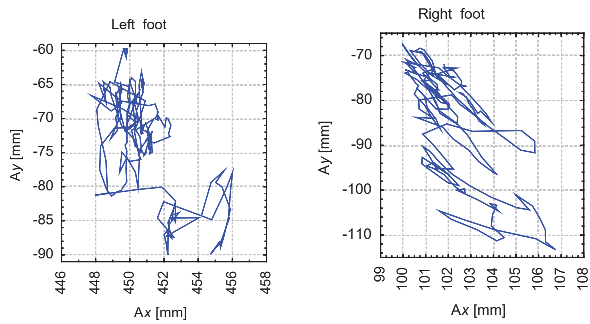

Figure 7. Stabilogram of right and left foot COP trajectories in the mediolateral (Ax) and anteroposterior (Ay) directions for Participant 72 (Quadrant III).

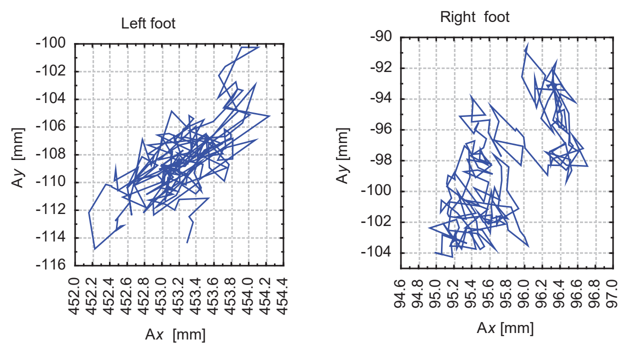

Figure 8. Stabilogram of right and left foot COP trajectories in the mediolateral (Ax) and anteroposterior (Ay) directions for Participant 93 (Quadrant I).

4. Discussion

This investigation considered how balance is maintained during an upright stance when COP is assessed over two points of application (both feet) with respect to the supporting surface. At first glance, it seems logical that the COP of each foot should be equal and, therefore, exhibit symmetry when balance is being maintained in a static position. The present study operated under the premise that a positive correlation between right and left foot COP trajectories (in the mediolateral and anteroposterior directions) indicates symmetry whereas a negative correlation indicates asymmetry. We hypothesized that the incidence of fluctuations or asymmetry between right and left foot COP can indicate foot pathology resulting from improper footwear or decreased neuromuscular control.

In our sample of healthy young adults, strong negative correlations between right and left foot COP in the mediolateral direction were found in approximately 67% of the participants. This indicates that during the quiet standing task the participants exerted pressure on the lateral boundary of the left foot while concurrently exerting pressure on the medial boundary of the right foot and, therefore, exhibit asymmetry. In turn, approximately 29% of the sample presented correlations coefficients between −0.4 and 0.3 between right and left foot mediolateral COP, which was considered to indicate more balanced symmetry albeit without any characteristic trend. More puzzling is the fact that the remaining 3% of the sample was found to show COP displacement simultaneously towards the medial and lateral sides of the right and left foot. From a biomechanical perspective, this type of balance control is difficult to understand. Interpretation of the correlation coefficients between right and left foot anteroposterior COP displacement found that approximately 88% of the sample presented considerable symmetry between both feet and that compared with mediolateral COP, anteroposterior

COP trajectory shows a greater level of symmetry between both feet. In effect, these results find that the majority of the sample presented asymmetry albeit defined as mediolateral COP trajectory traversing towards both the medial and lateral boundaries of the feet. Symmetry, in turn, was observed in the anteroposterior direction in which COP trajectory was along the toes or heel of the foot. This asymmetry and symmetry were presented by 86 participants ($n = 102$) and, therefore, suggests that this form of postural control defines healthy and active adults.

Research has suggested that the occurrence of asymmetry when maintaining balance to be the result of musculoskeletal dysfunction or lower extremity dominance. These conclusions were drawn by Ageberg [13], Lin [14] or Barone [15] who performed posturography on two force plates or by comparing COP variables between the dominant and non-dominant leg in static bilateral conditions. However, during single-leg testing, Hoffmann [16] and Greve [17] or Cuğ et al. [18] did not observe any differences in postural balance between the dominant and non-dominant leg in young adults. The aforementioned studies have mentioned that functional leg dominance may play an important role in bilateral postural stability, where, in most individuals, the left leg is the functionally dominant extremity and the right leg the functionally non-dominant extremity. Interestingly, Micarelli et al. [19] found greater COP displacement in the right rather than left leg during quiet standing in a group of children 4–13 and 4–7 years old. Research on the development of postural control by Assaiante [20] found that the trunk serves as an important reference frame in the emergence of structured postural strategies. It has been suggested that shifting the center of mass over the left increases weight-bearing load of the left leg over time.

Of consideration is the use of a scatter plot as presented herein as it can provide facile comparisons with other cohorts or normative data or illustrate more clearly the relationships of COP trajectories between the right and left foot. For example, when considering left foot COP trajectory in both directions, over half of the sample was found to present a pattern in which COP displacement traversed in the anteroposterior direction with a rightwards slant (medial direction) whereas right foot COP followed an anteroposterior displacement with a leftwards shift (medial direction). This medial shift of right and left foot COP trajectory with an anteroposterior displacement in the majority of the sample is contrasted by the marginal number of participants (three individuals) who presented an inverse pattern. In this small group, anteroposterior COP trajectory was associated with a lateral COP displacement (towards the outside of both feet) in the forward direction whereas in the backward direction COP transversed in the medial direction (towards the outside of the feet). In addition, the visualization of COP data by mapping its trajectory in the spatial domain can show patterns in COP trajectory over the base of the support, i.e., the feet. The trends that we observed in the majority of the sample (Quadrant II, Figure 5) confirm the findings of Oba et al. [21], Rival et al. [22] and Cumberworth et al. [23], who concluded that postural stability (COP displacement) is limited to an area between the heel and toe alongside the lateral edge in adults.

Our sample, as previously mentioned, was composed of young and asymptotic individuals free of lower extremity injury or disability. Although not assessed in the present study, it can be presumed that the incidence of foot or leg pathologies can modify the maximum displacement of COP trajectory and, therefore, stability. The literature contains numerous reports that utilize data on COP area and its relation to the base of support as a measure of postural stability [24–26]. Riach and Starkes [20] sound that children show a larger area of stability than adults and that the limit of stability decreases with age. Young adults aged 18–27 years appropriate on average 73% of the anteroposterior and 75% of the mediolateral base of support during upright standing. By ages 40–70, individuals use only 54% and 59%, respectively [22]. Clifford and Holder-Powell [23] indicated that the elderly show an even further reduced base of support for postural stability. Bottaroa [24] reported that this observation is associated with the fact that young asymptotic adults use up to 80% while older adults only up to 50% of foot length to maintain balance.

The present findings raise a number of questions concerning the etiology of disturbed COP displacement trajectory between the right and left foot and indicate the need for additional investigation in this area. Further research on COP symmetry in various populations (athletes, physically impaired, sedentary individuals) can provide further insight on the causes of asymmetry (e.g., decreased neuromuscular control) and aid clinicians with the diagnosis of various posture-related pathologies.

5. Conclusions

Approximately 88% of the participants exhibited left and right foot symmetry for anteroposterior COP trajectory magnitude and direction. Asymmetry was noted in 67% of participants for mediolateral COP trajectory, in which COP displacement was observed along the lateral boundary of one foot and along the medial boundary of the other. In 82% of the samples, COP displacement followed an anteroposterior trajectory with a medial slant for left foot and lateral slant for right foot COP, whereas the remaining 12% showed other variations. These findings raise a number of questions concerning the etiology of asymmetric COP displacement trajectory between the right and left foot. Additional investigation is warranted to compare COP trajectories between asymptotic individuals and various populations (e.g., sedentary individuals, patients with foot or leg pathologies or athletes engaged in sports with strong lateralization) to better understand the causes of asymmetry and aid clinicians with the diagnosis of various posture-related pathologies.

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Anthropometric Characteristics, Maximal Isokinetic Strength and Selected Handball Power Indicators Are Specific to Playing Positions in Elite Kosovan Handball Players

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Abstract: Anthropometric characteristics and physical performance are closely related to the game demands of each playing position. This study aimed to first examine the differences between playing positions in anthropometric characteristics and physical performance with special emphasis on the isokinetic strength of elite male handball players, and secondly to examine the correlations of the latter variables with ball velocity. Anthropometric characteristics, maximal isokinetic strength, sprinting and vertical jumping performance, and ball velocity in the set shot and jump shot were obtained from 93 elite handball players (age 22 ± 5 years, height 184 ± 8 cm, and weight 84 ± 14 kg) pre-season. Wing players were shorter compared to other players, and pivots were the heaviest. Wings had the fastest 20 m sprints, and, along with backcourt players, jumped higher, had better maximal knee isometric strength, and achieved the highest ball velocity compared to pivots and goalkeepers, respectively. There were no significant differences between playing positions in unilateral and bilateral maximal leg strength imbalances. Ball velocity was significantly correlated with height, weight, squat jump and maximal torque of extensors and flexors. Our study suggest that shooting success is largely determined by the player's height, weight, muscle strength and power, while it seems that anthropometric characteristics and physical performance are closely related to the game demands of each playing position.

Keywords: morphology; isokinetic; sprints; vertical jump performance; handball shooting

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1. Introduction

Handball is an Olympic team sport [1,2], split into two periods (each 30 min long) and consisting of a high degree of body contact and predominantly aerobic activities separated by anaerobic bouts of sprints, jumps, throws, changes of directions in the offense (counterattack and attack buildup) and defense [2,3]. Therefore, competition success in elite handball is not only closely related to the technical and tactical skills of each individual or team, but also to the players' anthropometric characteristics, physical performance (e.g., maximal strength and power as measured using strength, sprinting, and jumping tests), and handball shooting performance [3–5].

To date, only a few studies evaluating the physical and physiological demands in handball matches have shown significant differences between playing positions [6–8]. Results obtained during gameplay have shown that backcourt players cover larger distances and spend less time standing and walking, and, together with pivots, have higher in-game heart rates and spend longer durations at higher intensities (>80% maximal heart rate) [7]. In contrast, wing players are faster than other playing positions, and pivots endure more body impact than other players [6]. Furthermore, similar differences between playing positions were also obtained in anthropometric characteristics [9,10] and physical performance [11–16]. Wings are the shortest, have significantly lower body mass and body mass index (BMI) than other players; pivots are the heaviest, whereas other playing positions do not differ in height [9–16]. Research comparing playing positions in sprinting (e.g., sprints on 20–30 m) and jumping performance (squat jump (SJ) and/or countermovement jumps (CMJs)) is relatively scarce, and there are some discrepancies in sprinting times and jump heights among studies [11,14–16]. Nevertheless, backcourt players and wings demonstrated the fastest sprinting times [14,15], while CMJ height was the highest in wings [14], compared to pivots and goalkeepers [15].

Isokinetic dynamometry has long been the gold standard for assessing changes to an athlete's maximal muscle strength/torque during the season [17,18] and risk of injury (measured as imbalances between agonist and antagonist muscles or bilateral differences) in different types of a team sport [19]. In the past, only a few studies have evaluated isokinetic maximal knee strength and strength imbalances in male handball players [19–21], with the majority of studies having been performed on female handball players, focusing on shoulder muscle strength, and several on knee muscle strength in relation to shoulder and knee injuries [18,22–25]. Evidence from two studies evaluating isokinetic maximal strength in male handball players has suggested no differences between dominant and nondominant lower limb knee extensor and flexor strength, as well as normal hamstring-to-quadriceps ratio on both limbs at 60°/s and 180°/s [19,20]. However, no studies in handball have investigated isokinetic maximal strength of the knee joint according to playing position. Furthermore, it also remains to be elucidated the potential role of playing position on development of muscle strength imbalances between and within legs as a product of specific game demands. Such findings would serve as excellent feedback to practitioners to focus on preventing potential injuries and to improve player performance.

Ball velocity has been recognized as one of the most important determinants of game performance [14]. Maximal ball velocity is achieved through a proximal-to-distal manner; this movement allows momentum of force to transfer from the lower limb and/or pelvis through the trunk to the throwing arm, thereby enabling higher velocities of the shot [26]. During the game, the majority of shots are performed using two shooting techniques: a three-step jump shot and a standing set shot from the ground. Both techniques use two different kinetic strategies (braking the body with lead leg in the standing set shot vs. opposed leg movement during the flight phase of the three-step jump shot) [27,28], while ball velocity in both is influenced by an optimal proximal–distal principle, trunk movement, and maximal arm rotation [26,28]. It is well established that elite players are able to maintain an optimal proximal-to-distal principle and arm movement while performing different shooting techniques [27]. Lower ball velocity has been associated with lower strength in the lower and upper limbs, leading to inefficient transfer of power from the proximal (pelvis, trunk) to distal (shooting arm) parts of the body [29]. Backcourt players and wings shoot the ball faster compared to pivots or goalkeepers [14,15,30]. For overhead throws, only inconsistent correlations have been reported between ball velocity and anthropometric characteristics or physical performance, and this is likely due to the complex nature of this movement [10,11,13,31,32]. Body height and weight were the only anthropometric characteristics significantly correlated with ball velocity of the standing shot [11] and/or three-step running shot [10,13,31], whereas two additional studies reported significant correlations with lower limb strength (1-RM half back squat) [31], standing long jump, 30 m sprint, and maximal oxygen uptake (estimated from 20 m shuttle run) [31]. Other studies have failed to detect such correlations [11,13]; therefore, additional studies are warranted to transfer these findings into training settings.

Based on the identified gaps in the knowledge of the anthropometric characteristics, maximal strength and power performance, and handball shooting performance of elite handball players, our study consisted of three aims. The first aim of the study was to provide further evidence on the isokinetic maximal strength and potential limb imbalances between playing positions in elite male handball players. The second aim was to examine the differences in anthropometric characteristics, sprinting and vertical jump performance between playing positions, and the last aim was to investigate the correlations between handball-specific performance (e.g., ball velocity) and selected anthropometric characteristics and physical performance indicators.

2. Materials and Methods

2.1. Study Design

The study was designed as a cross-sectional study of a sample of elite Kosovo handball players. Measurements were conducted two weeks before the start of a competitive season, at the end of August 2019. Testing procedures were split into three days. On the first day anthropometric characteristics (height, weight, wingspan, and thigh circumference) and body composition (skeletal muscle mass and fat mass) were measured, and familiarization with all testing procedures was conducted. On the second day, following a standardized warm-up procedure, sprinting performance (20 m sprints), vertical jump performance (CMJ and SJ), and shooting performance (shooting velocity of three-step set shot and jump shot) were assessed. Finally, during the last day measurements of unilateral isokinetic knee flexor and extensor torque at 60°/s and 180°/s were conducted after the standardized warm-up. Forty-eight hours of rest were given between the second and third testing day to minimize any potential effects of fatigue. Testing procedures were performed by experienced strength and conditioning specialists.

2.2. Subjects

A total of 93 elite male handball players from Kosovo's first handball league were enrolled into the study, age (mean (SD)) 22 (1) years, height 184.0 (7.83) cm, weight 84.10 (13.74) kg, and with 8 (4) years of professional playing experience. The sample consisted of 35 (37.63%) backcourt players, 26 (27.96%) wing players, 15 (16.13%) pivots, and 17 (18.28%) goalkeepers. Playing positions were determined according to registration data obtained from the Handball Federation of Kosovo. Players' adherence was consistent throughout the study, and no injuries or other health issues were reported.

Prior to enrolment into the study, we informed all participants about the aims of our study, methods and procedures, and potential testing risks. Measurements were performed at the end of the last week of the specific preparation phase for the upcoming season, with at least 48 h of rest after the last training session, and 10–14 days prior to beginning of the season. The exclusion criteria were: fewer than two years of professional playing experience, age younger than 18 years, and any recent musculoskeletal injuries (<6 months). All participants signed a written consent prior to inclusion in the study. The study design was approved by the Ethics Committee of Universi College Prishtina (document number: 488/18), while the study was conducted according to the Declaration of Helsinki guidelines for the use of human participants.

2.3. Procedures

2.3.1. Anthropometric and Body Composition Measurement

Prior to physical performance measurements on the first testing day, anthropometric and bioimpedance measurements were obtained according to the international guidelines [33]. Body height and weight were measured while standing barefoot using a SECA 763 stadiometer with electronic scale (Seca Instruments Ltd., Hamburg, Germany) to the nearest cm and kg, respectively. The wingspan was measured using a horizontal wall-mounted scale to the nearest cm, with arms abducted at 90° from a neutral position and back facing towards the wall, and thigh circumference was measured using a tape measure while standing at 2/3 of the distance between the lateral epicondyle of the knee and the greater trochanter on the dominant leg. Skeletal muscle mass and fat mass measurements were obtained using a Biospace Inbody 720 bioimpedance device (Inbody Co., Leicester, United Kingdom). Participants were asked to place toes and heels on the anterior and posterior electrodes of the weighting platform, and to firmly grasp the hand grip with both hands. Measurements were taken early in the morning, and participants were advised to avoid any moderate to vigorous physical activity a day before the measurement [34].

2.3.2. Sprint Performance Measurement

After the general 15-min warm-up (10 min running, and 5 min of whole-body dynamic mobility exercises) and an additional three repetitions of progressive acceleration from faster to sprint running, participants performed two 20 m sprints, with 3 min of rest between each exertion. Prior to testing, four photocell gates (Polifemo Radio Light, Microgate, Bolzano, Italy) were placed at the start, at 5 m, at 10 m, and at 20 m. The dominant foot (lead-off foot) [35] was placed one meter behind the first photocell. The time recording was automatically initialized when a participant crossed the first photocell gate and stopped when the participant crossed the last photocell gate at 20 m distance. Participants were instructed to sprint at least 25 m in order to reach the highest maximum sprinting speed. The fastest of the two split times on 5 m, 10 m, and 20 m distances was used in the final analysis [4]. All measurements were performed indoors.

2.3.3. Vertical Jump Performance Measurement

Vertical jump performance, measured as jump height (cm), was evaluated from the CMJ and SJ using an OptoJump infrared timing system (Microgate, Bolzano, Italy). The participants first performed three trials of CMJs followed by three trials of SJs [15]. One minute of rest was given between two trials. Prior to performing both jumps, participants were instructed about the jumping technique and later performed at least two submaximal familiarization repetitions of CMJs and SJs to learn proper jumping techniques [4]. The CMJ was performed by flexing the knee to a squat position (approximately 90° of knee flexion) from an upright position and then immediately extending the hips and the knee into a vertical jump, whereas the SJ was performed by jumping to vertical from squat position (90° of knee flexion) [20]. When approaching the landing position, participants were advised to land with extended knees to avoid any measurement error resulting from prolonged flight time. Both jumps were performed with hands placed on hips and with legs straightened during the flight. The jump height was calculated from the recorded flight time (height = [gravitational acceleration (9.81 m/s^2) × flight time2] × 8 – 1) [4], and the highest jump was used in the final analysis.

2.3.4. Handball Shooting Performance Measurement

Handball shooting performance was evaluated by measuring the ball velocity of a three-step set shot from the ground and the ball velocity of a three-step jump shot from the 9-m line, using the Bushnell Radar (Bushnell, Overland Park, KS, USA) with a measurement error of ±1.60 km/h (www.bushnellspeedster.com). The investigator measured ball velocity while standing at the 9-m line within 1 m of the participant performing the throw. After the warm-up, each participant performed one familiarization shot and two test shots of each shot type, with one minute of rest between shots [14,15].

2.3.5. Maximal Isokinetic Strength Measurement

Isokinetic concentric torque of knee extensors and flexors was measured using an isokinetic dynamometer Biodex Pro 4 (Biodex Medical Systems, Shirley, New York, NY, USA) at 60°/s and 180°/s according to previous guidelines and studies [18,36]. Prior to testing day, the machine was calibrated according to manufacturer guidelines, using a long shoulder attached to the axis of the apparatus, generating a standard torque of 67.8 Nm.

Prior to testing, each participant completed a standardized warm-up protocol consisting of 10 min of light jogging, followed by short dynamic stretching exercises for lower limbs, and ending with a single 8-repetition set of squat and hip thrust exercises. After the general warm-up, the participants were seated upright in the dynamometer chair with restraining belts fastened across the chest, pelvis, and leg thigh to minimize body movement or any potential compensation of synergist muscles. Later, we aligned the dynamometer axis of rotation to the participant's knee joint axis of rotation using the lateral epicondyle as the anatomic mark. Additionally, gravitation torque error was measured prior to each trial, and the starting leg was randomly selected for each participant. The range of motion was set at 80°, from 90° to 10° of knee flexion.

Prior to measuring maximal effort, each participant first performed a specific warm-up on the dynamometer consisting of 10 submaximal concentric contractions of knee flexion and extension at 60°/s. The maximal test was conducted after 2 min of rest, with participants performing five maximal concentric knee extensions and flexions. Verbal encouragement was given by the investigator during the test to ensure participants performed at their maximal effort. The maximal value out of five measurements was normalized to body weight (N/kg) and used in the final statistical analysis. In addition, bilateral differences between left and right maximal isometric torque (left leg/right leg maximal isometric torque × 100%) and unilateral hamstring-to-quadriceps maximal isometric torque (hamstring/quadriceps maximal isometric torque × 100%) [37] was calculated prior to further statistical analysis.

2.4. Statistical Analysis

Categorical variables are presented as frequencies and percentages, and numeric variables are presented as means and standard deviations, unless otherwise stated. All numeric variables were firstly screened for assumptions of normality of distribution and homogeneity of variances using the Shapiro–Wilk test and the Levene's test, respectively. This was screened for the whole sample and according to each playing position. The difference between playing positions was calculated using one-way analysis of variance (ANOVA) for normally distributed variables and homogeneous variances, otherwise, the Kruskal–Wallis test was applied. When one-way ANOVA detected significant differences between playing positions, an additional post hoc analysis was performed using the Tukey's honest significance test or pairwise comparisons, depending on the dispersion of variances between playing positions. Correlations between anthropometric, physical, and handball performance were assessed using Spearman's rank correlation coefficient. All statistical analyses were performed using IBM SPSS version 21 (SPSS Inc., Armonk, New York, NY, USA), and the level of significance was set at p-value < 0.05.

3. Results

There were statistically significant differences for playing positions in all measured anthropometric characteristics and skeletal muscle mass and fat mass (all p-values < 0.01; Table 1). The wings were significantly shorter compared to backcourt players ($p < 0.001$), pivots ($p = 0.035$), and goalkeepers ($p = 0.018$). Similar significant differences were obtained in weight (pivots vs. wing, $p < 0.001$; wings vs. backcourt players, $p < 0.001$), wingspan (wings vs. pivots, $p < 0.001$; wings vs. backcourt players, $p < 0.001$; wings vs. goalkeepers, $p = 0.038$), and thigh circumference (wings vs. pivots, $p < 0.001$; wings vs. backcourt players, $p = 0.041$). Additionally, pivots were heavier than goalkeepers ($p < 0.001$) and backcourt players ($p = 0.003$) and had larger thigh circumference compared to goalkeepers ($p < 0.001$) and backcourt players ($p < 0.001$), whereas backcourt players were heavier than goalkeepers ($p = 0.015$).

Table 1. Differences between playing positions in anthropometric characteristics and body composition.

	Backcourt Player (N = 34)	Wing (N = 26)	Pivot (N = 15)	Goalkeeper (N = 17)	p
Height (cm)	187 (8)[1]	179 (7)	185 (7) [1]	185 (6) [1]	<0.001
Weight (kg)	87 (11) [1,2]	75 (9)	102 (11) [1,2]	78 (10)	<0.001
Wingspan (cm)	190 (9) [1]	180 (8)	191 (9) [1]	186 (8) [1]	<0.001
Thigh circ. (cm)	60 (5) [1]	57 (5)	68 (5) [1,2,3]	58 (5)	<0.001
Skeletal muscle mass (kg)	44 (5) [1,2]	38 (4)	47 (5) [1,2]	37 (4)	<0.001
Body fat mass (kg)	12 (5)	10 (4)	23 (8) [1,2,3]	15 (7) [1]	<0.001

Cir.–circumference, [1]—significantly different from the wing, [2]—significantly different from the goalkeeper, [3]—significantly different from backcourt players.

Body composition measurements revealed that goalkeepers had significantly less skeletal muscle mass than backcourt players ($p < 0.001$) and pivots ($p < 0.001$), while wings had less skeletal muscle mass than backcourt players ($p < 0.001$) and pivots ($p < 0.001$). Pivots, on the other hand, had significantly more fat mass than wings ($p < 0.001$), backcourt players ($p < 0.001$) and goalkeepers ($p = 0.005$). Lastly, goalkeepers also had significantly more muscle mass than wings ($p = 0.010$).

Similar to the anthropometric characteristics, the playing positions differed in maximal isokinetic concentric extensor strength at 60°/s and 180°/s ($p < 0.001$), concentric knee flexion of the left knee at 60°/s ($p = 0.007$), and borderline significance of the right knee ($p = 0.070$) (Table 2). Pivot players displayed lower isokinetic concentric torque of knee extensors and flexors compared with wings and backcourt players at 60°/s and 180°/s, respectively, whereas backcourt players were superior compared to goalkeepers at 60°/s and 180°/s of knee flexion and extension. Additionally, wings performed better than goalkeepers at 60°/s and 180°/s of knee extension and flexion. Otherwise, no differences between playing positions were obtained in hamstring-to-quadriceps ratio (HQR) and in bilateral differences in maximal concentric torque of extensors and flexors at 60°/s and 180°/s.

Table 2. Differences among playing positions in isokinetic concentric torque.

	Backcourt Player (N = 34)	Wing (N = 26)	Pivot (N = 15)	Goalkeeper (N = 17)	p
Extension left knee at 60°/s (Nm/kg)	2.80 (0.58)	2.88 (0.55)	2.23 (0.45) [1,2]	2.53 (0.32) [1,2]	<0.001
Extension right knee at 60°/s (Nm/kg)	2.86 (0.45)	2.83 (0.56)	2.30 (0.43) [1,2]	2.50 (0.35) [2]	<0.001
Extension left knee at 180°/s (Nm/kg)	1.80 (0.32)	1.85 (0.28)	1.42 (0.28)[1,2]	1.49 (0.29) [1,2]	<0.001
Extension right knee at 180°/s (Nm/kg)	1.79 (0.27)	1.73 (0.36)	1.45 (0.27) [1,2]	1.51 (0.24) [2]	<0.001
Flexion left knee at 60°/s (Nm/kg)	1.74 (0.38)	1.74 (0.38)	1,50 (0.63) [1,2]	1.52 (0.24) [1,2]	0.007
Flexion right knee at 60°/s (Nm/kg)	1.78 (0.25)	1.70 (0.39)	1.57 (0.59)	1.53 (0.39)	0.070
Flexion left knee at 180°/s (Nm/kg)	1.33 (0.29)	1.33 (0.30)	1.07 (0.34) [1,2]	1.11 (0.26) [1,2]	0.002
Flexion right knee at 180°/s (Nm/kg)	1.32 (0.25)	1.21 (0.28)	1.08 (0.34) [2]	1.12 (0.22) [2]	0.012
Bilateral ratio left–right knee extension at 60°/s (%)	98.23 (15.08)	102.36 (9.97)	95.95 (7.04)	96.95 (7.04)	0.251
Bilateral ratio left–right knee flexion at 60°/s (%)	98.50 (21.14)	103.16 (11.35)	95.17 (10.47)	102.86 (20.42)	0.059
Bilateral ratio left–right knee extension at 180°/s (%)	101.89 (19.47)	110.65 (16.02)	100.71 (13.56)	98.58 (11.86)	0.112
Bilateral ratio left–right knee flexion at 180°/s (%)	101.43 (17.55)	109.70 (19.37)	97.76 (9.74)	98.59 (13.11)	0.068
HQR left knee at 60°/s (%)	63.23 (17.97)	60.84 (11.78)	70.29 (36.66)	59.93 (6.39)	0.725
HQR right knee at 60°/s (%)	63.56 (12.52)	60.54 (11.69)	69.40 (28.23)	61.06 (12.21)	0.958
HQR left knee at 180°/s (%)	74.18 (15.22)	71.55 (11.83)	77.35 (27.68)	74.67 (13.42)	0.596
HQR right knee at 180°/s (%)	74.32 (14.73)	70.90 (12.91)	74.58 (21.80)	74.17 (10.87)	0.704

HQR–hamstring-to-quadriceps torque ratio, [1]—significantly different from the wing, [2]—significantly different from backcourt players.

Measurements of sprinting and vertical jump performance showed significant differences between playing positions (all p-values < 0.01; Table 3). Wing players were significantly faster than backcourt players (5 m, $p = 0.042$; 10 m, $p = 0.001$; and 20 m, $p = 0.027$), goalkeepers (5 m, $p = 0.013$; 10 m, $p = 0.001$; and 20 m, $p = 0.001$), and pivots (5 m, $p = 0.001$ and 20 m, $p < 0.001$). At the 10-m time gate, backcourt players were significantly faster than goalkeepers ($p < 0.001$). Moreover, backcourt players and wings jumped significantly higher compared to pivots (backcourt players vs. pivots: CMJ, $p = 0.007$, SJ, $p = 0.027$; wings vs. pivots: CMJ, $p < 0.001$, SJ, $p < 0.001$) and goalkeepers (backcourt players vs. pivots: CMJ, $p = 0.006$; wings vs. goalkeepers: CMJ, $p < 0.001$, SJ, $p = 0.004$), respectively.

Table 3. Differences between playing positions in sprinting, vertical jump performance, and handball shooting performance.

	Backcourt Player (N = 34)	Wing (N = 26)	Pivot (N = 15)	Goalkeeper (N = 17)	*p*
Sprint 5 m (s)	1.05 (0.08) [1]	1.01 (0.08)	1.11 (0.09) [1]	1.10 (0.12) [1]	0.006
Sprint 10 m (s)	1.83 (0.20) [1]	1.75 (0.24)	1.93 (0.30)	1.97 (0.13) [1,3]	0.003
Sprint 20 m (s)	3.31 (0.24) [1]	3.15 (0.19)	3.44 (0.22) [1]	3.42 (0.17) [1]	<0.001
SJ (cm)	30 (5)	32 (6)	25 (6) [1, 3]	26 (4) [1,3]	<0.001
CMJ (cm)	34 (4)	37 (6)	29 (6) [1,3]	31 (5) [1]	<0.001
Set shot (km/h)	89 (7) [1,2]	84 (9) [2]	89 (5) [2]	79 (6)	<0.001
Jump shot (km/h)	88 (7) [1,2]	83 (9) [2]	87 (9) [2]	77 (6)	<0.001

SJ–squat jump, CMJ–countermovement jump, [1]—significantly different from the wing, [2]—significantly different from the goalkeeper, [3]—significantly different from backcourt players.

There were also significant differences in ball velocity among playing positions (both *p*-values <0.001). Post hoc analysis showed that goalkeepers shoot the ball at significantly lower velocity while shooting from ground position (all *p*-values < 0.01) or while performing a three-step jump shot (all *p*-values < 0.01).

Correlations between handball shooting performance and sprinting, jumping, and maximal strength performance are shown in Table 4. With the exception of body fat mass and thigh circumference, all other anthropometric characteristics were significantly correlated with the ball velocity of a three-step set shot and jump shot. A higher SJ was significantly correlated with the ball velocity of the set shot, and borderline significant with the ball velocity of the jump shot. Maximal isokinetic torque of knee flexors and extensors was significantly correlated with ball velocity of both shot types. Lastly, HQR at 60°/s was significantly correlated with the ball velocity of the set shot.

Table 4. Correlations between playing positions in sprinting, vertical jump performance, and handball shooting performance.

	Set Shot (m/s)		Jump Shot (m/s)	
	Spearman Rho	*p*	Spearman Rho	*p*
Height (cm)	0.330	0.001	0.263	0.011
Weight (kg)	0.303	0.003	0.282	0.006
Skeletal muscle mass (kg)	0.522	<0.001	0.473	<0.001
Body fat mass (kg)	−0.116	0.267	−0.083	0.428
Wingspan (cm)	0.387	<0.001	0.349	0.001
Thigh circumference (cm)	0.183	0.080	0.166	0.112
Sprint 20 (s)	0.061	0.566	−0.018	0.862
SJ (cm)	0.210	0.043	0.185	0.076
CMJ (cm)	0.128	0.221	0.057	0.585
Knee extension torque at 60°/s (Nm)	0.219	0.035	0.340	0.001
Knee extension torque at 180°/s (Nm)	0.352	0.001	0.419	<0.001
Knee flexion torque at 60°/s (Nm)	0.493	<0.001	0.465	<0.001
Knee flexion torque at 180°/s (Nm)	0.477	<0.001	0.460	<0.001
HQR at 60°/s (%)	0.317	0.002	0.171	0.101
HQR at 180°/s (%)	0.139	0.185	0.061	0.564

SJ–squat jump, CMJ–countermovement jump, HQR–hamstring-to-quadriceps torque ratio.

4. Discussion

In the present study, we identified differences between playing positions in anthropometric characteristics, isometric maximal leg strength, sprinting and vertical jumping performance, and established new evidence on the relationship between anthropometry, physical performance, and ball velocity as an indicator of game performance. The most novel findings of this study were related to isokinetic performance, adding to the few reports of the isokinetic maximal strength and strength imbalances in elite male handball players that have been published to our knowledge [19–21]. Our results suggest that maximal knee flexor and extensor strength is related to playing position, whereas no differences between playing positions were observed in bilateral muscle imbalances or the ratios between knee joint agonists and antagonists.

In the previous studies of sports performance in elite handball, the investigators applied different methods to assess maximal leg strength. Most of those studies on elite male handball players used different variations of maximal squat tests to determine maximal leg strength [4,11,15,32], while (only) a small body of evidence used isokinetic testing [19,20], despite it being considered the gold standard for assessing quadriceps and hamstring maximal strength and muscle imbalances [17,18]. Most studies measuring maximal isokinetic knee strength were performed with females [18,24,38,39], likely due to higher rates of anterior cruciate ligament injuries, compared with males [40], while only two isokinetic studies included male handball players [19,20]. In the latter studies, male handball players were recognized as functionally balanced athletes, where maximal unilateral (50–69%) and bilateral (10–15%) muscle strength differences were in the normal range [19,20]. This was similarly demonstrated in our study, although there were no bilateral differences in maximal strength of flexors and extensors, or muscle imbalances between hamstrings and quadriceps on each leg at 60°/s and 180°/s. Our relative values of maximal flexion and extension torque at 60°/s and 180°/s (N/kg) were also similar to a report by Gonzalez-Rave et al. [20]. Additionally, our study also evaluated the difference between playing positions in relative maximal strength and muscle imbalances. Backcourt and wing players were the strongest in extension and flexion, independent of muscle mass at both angular velocities. These results may potentially be associated with game demands, as wings and backcourt players perform the most jumps and throws [7]. In contrast, all playing positions showed symmetrical strength (unilateral and bilateral ratios in the normal range) between legs and within each leg analysis.

Studies have suggested that anthropometric characteristics are related to playing position. In our study, wing players were the shortest, had the lowest body weight, shortest wingspan and smallest thigh circumference, while pivots had the highest body weight, skeletal muscle mass, and body fat mass content. These results were in line with previous studies [7,10,13–16] from elite and sub-elite male handball players. Despite the similar ages of players and the differences between playing positions in body weight and height, our subsample of backcourt players, pivots, and goalkeepers was generally shorter and lighter than players competing on elite German [10,14], Norwegian [15], and national teams playing in the World Championship [9]. Thus, these differences can be explained by the level of play and age. Players competing in lower leagues were generally shorter [10], independent of playing position, whereas pivots in higher leagues were heavier and had the highest body mass index [14]. Also, younger elite handball players were shorter, lighter, had lower free fat mass and body mass index values compared to elite adult players [16].

The fastest sprint times were recorded in wing players compared to other playing positions on each of three time gates (5 m, 10 m, and 20 m). Similar variations in sprinting performance at 20 m were observed in the other two studies consisting of handball players performing in elite European leagues [14,15]. In addition to the fastest times recorded by wings, backcourt players were also faster than goalkeepers on 10-m sprints, similar to data obtained by Haugen et al. [15] (wings, 2.78 (0.08) s; backcourt players, 2.83 (0.11) s; goalkeepers, 2.94 (0.10) s). Partly contrary to our findings, one study reported significant differences between playing positions only on longer sprint distances (30 m), postulating equal starting acceleration of all playing positions, which likely contributed to the longer competitive career and a higher level of competition [14]. Moreover, sprinting performance is closely related to the in-game demands of each playing position. Data derived from game movement analysis has shown that wing players have a higher frequency of performed sprints, with the longest duration, time, and fraction of distance covered compared to pivots and backcourt players [7].

Vertical jumping is an important movement performed during the course of the game [7]. In previous studies, the best vertical jumps were performed by the wings (39–50 cm) and backcourt players (38–47 cm), compared to pivots (35–43 cm) and goalkeepers (35–47 cm) [14,15]. Similar differences between playing positions were also obtained in our study. Nevertheless, jump performance was lower compared to data from two samples of elite European Championship players [14,15], but comparable to a similar level of play [16], thus, our results were likely influenced by the level of play and quality of the training regime.

Handball scoring efficiency is largely dependent on ball velocity. In our study, ball velocity was highest in backcourt players and lowest in wings and goalkeepers. A similar superiority was also obtained in other samples of elite players [13–15], although the maximal shooting velocities of our participants (88.94 km/h) were only comparable to one study (90.72 km/h) [14], while others performed better (94.32–96.84 km/h) [13,15]. The best shooting performance from three-step shots was reported in a sample of Tunisian national team players (99.67 km/h), although it showed no significant difference between playing positions, likely due to its small sample size (N = 21) [11].

The kinematics of overhead shoots is a highly complex whole-body movement [26], with many studies undertaken to identify the possible determinants of shooting success [10,11,13,31,32]. Our results supported previous findings that suggest body height and mass may influence shooting velocities from a standing [11] and/or jumping position [10,13,31]. In contrast, several studies investigating the correlations between physical performance and ball velocity have been inconclusive [11,13,31,32]. Others, including our study, reported significant correlations between ball velocity and sprinting time, lower limb maximal strength, and endurance [31,32], while others failed to reach such conclusions [11,13]. Furthermore, our results also highlighted the importance of lower limb muscle mass and strength as an initiator of proximal-to-distal principal during the shot [41]. When an optimal sequence of force translation from proximal muscles of legs, pelvis, and trunk to throwing arm is achieved, the highest force production in the leg muscles can substantially contribute to higher ball velocities [27,29,32], as confirmed in our study. Similarly, recent study has suggested that jump height in the CMJ is significantly correlated with jump height while performing a jump shot in a game-based performance test [42]. During the handball game, this may present an advantage over the opponent, as the three-step jump shot is the most frequently executed shot [26]. In addition to faster ball-shooting velocity, the importance of jump height may also explain the higher frequency of jumps and shots performed during the game by backcourt players compared to pivots and goalkeepers [7]. However, as a handball shot is a complex, multi-joint movement, more research is needed to determine new potential physical performance determinants of shooting success (e.g., ball velocity).

In summary, the results of this study may further clarify several important aspects of the anthropometric and physical aspects of handball performance with special reference to playing position. Firstly, we confirmed previously reported variations between playing positions in anthropometric characteristics, sprinting, jumping, and handball shooting performance. Secondly, and most importantly, our study was one of the first to establish new evidence on the isokinetic maximal strength of lower limbs. We provided novel data for maximal torque of extensors and flexors in elite handball as well as demonstrated that male handball players are symmetrical with no significant maximal strength deficits between, and within, knee extensors and flexors. Lastly, our study also established further evidence on the potential role of various physical performance aspects in handball success as measured by ball velocity. Despite presenting novel findings on one of the largest samples in male handball performance research, some limitations must be acknowledged. Most of our sample were members of Kosovo's national handball team, but none of them played abroad in higher-ranking leagues or the European Championship league. As Kosovo is a young country with very few professional handball opportunities, our results may be affected by the playing level, training process, and relative lack of experience of the players. Nevertheless, our data showed results comparable to other elite playing countries, e.g., Germany and Norway, therefore, we believe that strong professional handball foundations have been built in Kosovo.

5. Conclusions

In conclusion, our study clearly demonstrated the importance of anthropometry, jumping performance, and maximal isometric strength to handball performance. In future sports practice, more emphasis should be given to handball-specific resistance training for players to gain more muscle mass of lower (legs and pelvis) and upper limbs (shoulders and arms), which will afterwards manifest in better jumping, sprinting, and shooting performance. Special consideration must be given to resistance training of pivots and goalkeepers to improve their muscle strength and shooting performance [2], as they were outperformed by wings and backcourt players. Furthermore, we also propose the routine inclusion of isometric measurement of shoulder and knee joint maximal torques to monitor changes of maximal muscle strength during the course of the handball season and to detect potential muscle imbalances, which may contribute to higher injury incidence. In line with this, more studies on elite and sub-elite handball players should be conducted to provide new practical evidence of the importance of isokinetic testing and to further determine several important aspects of physical performance in relation to handball shooting performance.

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6

The Effect of a 7-Week Training Period on Changes in Skin NADH Fluorescence in Highly Trained Athletes

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Abstract: The study aimed to evaluate the changes of nicotinamide adenine dinucleotide (NADH) fluorescence in the reduced form in the superficial skin layer, resulting from a 7-week training period in highly trained competitive athletes ($n = 41$). The newly, non-invasive flow mediated skin fluorescence (FMSF) method was implemented to indirectly evaluate the mitochondrial activity by NADH fluorescence. The FMSF measurements were taken before and after an exercise treadmill test until exhaustion. We found that athletes showed higher post-training values in basal NADH fluorescence (pre-exercise: 41% increase; post-exercise: 49% increase). Maximum NADH fluorescence was also higher after training both pre- (42% increase) and post-exercise (47% increase). Similar changes have been revealed before and after exercise for minimal NADH fluorescence (before exercise: 39% increase; after exercise: 47% increase). In conclusion, physical training results in an increase in the skin NADH fluorescence levels at rest and after exercise in athletes.

Keywords: nicotinamide adenine dinucleotide; training; athletes; mitochondrion

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1. Introduction

Skin microcirculatory function and efficiency of blood supply to the skin can impact mitochondrial activity and the changes of nicotinamide adenine dinucleotide (NADH) fluorescence in the reduced form [1]. Mitochondrial function can be indirectly evaluated by NADH fluorescence [1] that has been measured in animals [2,3] and humans [1,4] at rest and under various conditions (e.g., ischemia and temperature changes). Bugaj et al. [5] were the first to describe the time course of NADH changes in the skin in athletes at rest and after exercise. In their study, a new method of evaluating NADH fluorescence—flow mediated skin fluorescence (FMSF)—was utilized. The FMSF method is based on the ability of NADH to autofluorescence. The fluorescence measured using the FMSF method reflects the dynamics of in vivo changes in NADH levels in most superficial layers of the skin [5–7]. Bugaj et al. [5] have shown that exercise to exhaustion induces changes in skin NADH fluorescence, in other words, the values recorded after exercise were higher than those before exercise (increase in: basal NADH fluorescence 13%, maximal NADH fluorescence 7% and minimal NADH fluorescence 12%).

Nicotinamide adenine dinucleotide (NAD) is synthesized in the cytosol, mitochondria, and nucleus. This molecule is active in the cytoplasm during glycolysis and in the mitochondria during oxidative phosphorylation when adenosine-5′-triphosphate (ATP) is produced [8]. NAD occurs in two forms:

oxidized NAD^+ and reduced NADH. NAD takes part in many biological reactions including electron transport. The reduction of NAD^+ to NADH occurs almost exclusively in the mitochondria at the final stage of cellular respiration [9,10].

In the human body, there is a pool of NAD that takes reduced (NAD^+) and oxidized (NADH) forms, transforming into each other [8]. Importantly, the NAD pool is only constant for relatively short periods [8,11]. In the long term, the NAD amount changes depending on several factors such as age, diet, physical activity, medicaments, boosters, time of the day, etc. [11]. NAD^+ metabolism is complex and includes many NAD^+-consuming pathways as well as de novo and salvage pathways [8].

Mayevsky and Barbiro-Michaely [1] have claimed that the monitoring of the NADH level in tissue provides important information about the mitochondrial metabolic state (energy production, amount of intracellular oxygen). In addition, changes in the NAD^+/NADH ratio reflect cellular respiration processes in mitochondria, thus indirectly represent their function [1,5]. Studies on changes in NADH in response to physical exercise were performed on animal and human skeletal muscle samples, but not in the skin [8,9,12]. Early reports including animals did not provide a clear answer as to how NADH levels were modified by exercise [13,14]. Subsequent human research had shown that intensive exercise, unlike light exercise, shifted the NAD^+/NADH balance toward NADH [8,15]. Only Koltai et al. [16] have examined the influence of endurance training on changes in NAD^+ level in rat muscles and showed that training resulted in an increase in NAD^+ biosynthesis.

Studies on skeletal muscle mitochondria are valuable, but usually invasive due to the use of the biopsy technique [17,18] and expensive if transmission electron microscopy is used [19]. However, it has been suggested that physical exercise brings beneficial changes not only in skeletal muscle mitochondria, but also in skin mitochondria [20]. It has been demonstrated that physical exercise results in several beneficial mitochondrial adaptations [19,21–25]. Various changes were extensively studied in skeletal muscle mitochondria [19,21,25–27], while only one study dealt with the changes in the skin [20]. However, we do not know whether training only affects muscle mitochondria, or the adaptations also take place in skin mitochondria that are easily accessible to study because they lie superficially.

To the best of our knowledge, there is a lack of studies describing the effect of physical training on changes in NADH fluorescence in the skin. The novel, noninvasive, and cheap flow mediated skin fluorescence method can be a source of valuable information about the mitochondrial activity. Therefore, the study aimed to evaluate the changes in NADH fluorescence in the superficial skin layer resulting from a 7-week training period in highly trained competitive athletes. We hypothesize that physical training results in an increase in the NADH fluorescence levels in athletes.

2. Materials and Methods

2.1. Subjects

Forty-one highly trained athletes (28 men, 13 women), ages ranging from 18 to 35 years, participated in the study. They were members of the Polish national team or athletes taking part in national and international competitions. They represented the following sport disciplines: triathlon (Olympic distance: 1.5 km swim, 40 km bike ride, 10 km run) (seven men, four women); long-distance running (5 km, 10 km, and marathon) (six men, two women); Olympic taekwondo (six men, one woman); sprint (100 m, 200 m, and 4 × 100 m relay) (six men, one woman); canoeing (three men); and fencing (five women). Before starting the study, each participant was informed about the aim and procedures, potential risks, and the possibility to withdraw at any time without giving any reason. All athletes gave their written consent to participate in the examinations and fulfilled a questionnaire on their health status and potential contraindications. All athletes had valid health certificates issued by a physician who specialized in sports medicine, thus were eligible for training and competition. Exclusion criteria were illness symptoms, injuries, and taking drugs (temporarily or chronically). Only the data of those athletes who were present at both examinations was analyzed. The study was conducted in accordance with the Declaration of Helsinki. The Ethics Committee of the Poznan University of Medical Sciences in Poland approved the study protocol (approval no. 1017/16 issued on 5 October 2016).

2.2. Training Characteristics

All participants attended training sessions at least six times a week. During the whole 7-week period under study (general preparation phase of the one-year cycle), the athletes had on average 57 training sessions of a total duration of 71.2 h. The average duration of a single session was 84 min.

2.3. Study Design

The study was conducted in the Human Movement Laboratory of the Department of Athletics, Strength and Conditioning at the Poznan University of Physical Education (Poznań, Poland). Athletes arrived at the laboratory in the morning. During all measurements, the constant temperature was maintained (20-21 °C) by an air conditioning system. On the day of the examination, the participants could only eat a light breakfast. It was also recommended for them to avoid coffee or tea for 12 h, alcohol for 24 h, and hard exercise for 48 h before each examination. After arriving, athletes changed into their lightweight sports clothing (without watches and wristbands potentially affecting blood flow) and acclimatized to the laboratory conditions for at least 30 min. During this time, they completed the required questionnaires, and height and weight measurements were performed.

Athletes underwent the examinations twice: at the beginning of the general preparation phase and after seven weeks, at the end of this phase. Each time, the same procedure was applied: (1) initial resting blood pressure measurement; (2) resting NADH fluorescence measurement; (3) blood draw, (4) incremental exercise test; (5) second blood draw; (6) post-exercise blood pressure measurement; and (7) post-exercise NADH fluorescence measurement (3 min after the end of the test).

2.4. Incremental Exercise Test

The exercise test was conducted on the H/P Cosmos treadmill (h/p/cosmos sports & medical GmbH, Nussdorf – Traunstein, Germany). All participants were familiar with the treadmill test because they regularly (2-3 times a year) participated in similar tests. The purpose of this examination was to assess maximal oxygen uptake (VO_2max) and peak heart rate (HR).

Respiratory gases were collected and analyzed using the MetaMax 3B ergospirometer (Cortex Biophysik BmbH, Leipzig, Germany) and the MetaSoft Studio 5.1.0 software (Cortex Biophysik BmbH, Leipzig, Germany). The exercise protocol started with a 4-min warm-up at the treadmill speed of 6 km/h. Then, the treadmill accelerated by 2 km/h every 3 min. The treadmill inclination was 1% throughout the whole test. The test terminated if the athlete signaled his/her voluntary exhaustion by raising one hand. Maximal oxygen uptake was considered to be reached if the oxygen uptake (VO_2) was stabilized despite the further increase in treadmill speed. All participants were highly trained, so during the test, all of them reached a plateau in VO_2 uptake. We also checked three additional conditions to confirm reached maximal oxygen uptake: (i) HR reached at least 95% of the age-adjusted HR; (ii) cutoff blood lactate concentration $\geq$ 9 mmol/L for man and $\geq$7 mmol/L for women; and (iii) respiratory exchange ratio was $\geq$1.1 [28]. Heart rate was measured using the Polar H6 Bluetooth Smart monitor (Polar Electro Oy, Kempele, Finland) attached to a chest strap.

2.5. Lactic Acid Measurements

Capillary blood samples were obtained from the fingertip at rest and 2 min after the exercise test. A total of 20 μL of whole blood was drawn to a micro test tube using a capillary. Biosen C-line (EKF Diagnostics, Cardiff, UK) was used to measure the level of lactate.

2.6. Anthropometric Measure

Anthropometric measurements were performed according to standardized procedures. Body mass (kg) and height (cm) were measured with a digital measuring station Seca 285 (SECA, Hamburg, Germany). Body mass index (BMI) was calculated as body weight divided by height squared (kg/m^2).

2.7. Nicotinamide Adenine Dinucleotide Fluorescence

NADH fluorescence was measured using the AngioExpert device (Angionica, Łódź, Poland, 2016) based on the flow mediated skin fluorescence (FMSF) method. FMSF enables recording of the changes in NADH fluorescence as a function of time in response to ischemia and reperfusion in forearm skin cells. During the measurement, AngioExpert emits light at the wavelength of 460 nm [6,7]. NADH molecules have autofluorescence capability at a wavelength of 460 nm [9]. The changes in fluorescence intensity observed during the examination are produced in the most superficial skin cells (epidermis) [6,29], which is due to very shallow skin penetration by excitation light at the wavelength of 340 nm. About 90% of the recorded signal comes from the skin depth up to 0.5 mm. The activated skin region is not directly supplied with blood, but is supplied with oxygen by deeper blood vessels [6,7,29].

During the examination, each participant sat on a chair with his/her arm resting on the measuring device. Immediately before examination, systolic (SBP) and diastolic (DBP) blood pressure was measured using the Omron 3 (Omron, Kyoto, Japan) device. At the start of the FMSF examination, basal fluorescence was registered for 2 min. Then, an occlusion cuff was inflated up to the pressure of 50 mmHg above the SBP for 200 s. After this time, blood flow in the forearm was restored (cuff deflated) and the changes in NAD fluorescence were recorded for a further 3 min [7].

The following parameters related to NAD fluorescence were measured or calculated (Figure 1):

- B_{mean}—Basal fluorescence at the wavelength of 460 nm, recorded at rest at the beginning of the measurement;
- FI_{max}—The maximal increase in fluorescence above the baseline observed during forearm ischemia;
- FR_{min}—The maximal drop in fluorescence below the baseline observed during reperfusion;
- I_{max}—The relative increase in fluorescence = the difference between I_{max} and B_{mean};
- R_{min}—The relative drop in fluorescence = the difference between B_{mean} and FR_{mean};
- IR_{ampl}—The maximal range of changes in fluorescence = the sum of R_{min} and I_{max}; and
- CI_{max}—The relative (percentage) contribution of I_{max} to IR_{ampl} [7].

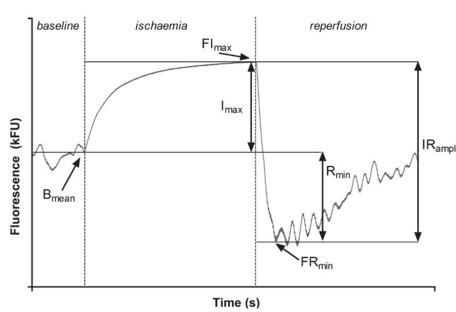

Figure 1. Parameters describing the Flow Mediated Skin Fluorescence. B_{mean}—Mean value of the basal fluorescence; FI_{max}—Maximal fluorescence during ischemia; FR_{min}—The first minimal fluorescence value during reperfusion; I_{max}—The net increase in fluorescence over the baseline during ischemia; IR_{ampl}—The amplitude of fluorescence change during ischemia and reperfusion; R_{min}—The net reduction in fluorescence below the baseline. Reprinted from Bugaj et al. [5].

The second measurement was made according to the same methodology, 3 min after the end of the treadmill test. A sample measurement of the NADH fluorescence from a 23-year-old male sprinter before and after training was shown in Figure 2.

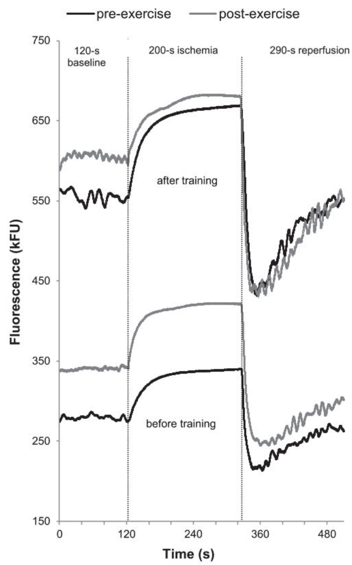

Figure 2. A sample Flow Mediated Skin Fluorescence measurement in a 23-year-old male sprinter. Changes in nicotinamide adenine dinucleotide fluorescence are shown before and after 7-weeks of training, at rest, and after cardiopulmonary exercise test until exhaustion. The first 2 min serve to determine the baseline fluorescence level. This was followed by a 200-s ischemia (increase in fluorescence) and a 290-s reperfusion (decrease in fluorescence).

3. Results

3.1. Basic Characteristics

The resting DBP, SBP, and BMI were within normal ranges. Other descriptive characteristics are presented in Table 1.

Table 1. Basic characteristics of the studied athletes.

Parameter	Before Training	After Training
Age (years)	22.4 ± 4	–
Training experience (years)	8 ± 2.3	–
Height (cm)	178.1 ± 7.3	178.1 ± 7.3
Weight (kg)	69.1 ± 10.3	69 ± 10.3
BMI (kg/m^2)	21.6 ± 2.3	21.6 ± 2.3
SBP_{rest} (mmHg)	127.6 ± 14.3	119.3 ± 10.8 ***
DBP_{rest} (mmHg)	69.9 ± 7.3	72.9 ± 9.3 *
SBP_{exerc} (mmHg)	148 ± 18.3	139.2 ± 16.3 **
DBP_{exerc} (mmHg)	74.5 ± 8.1	78.2 ± 8.1 *
VO_2max (mL/min/kg)	58.8 ± 8.6	59.5 ± 8.6
HR_{peak} (beats/min)	191.7 ± 8	191.9 ± 8.9
LA_{rest} (mmol/L)	1.2 ± 0.5	1.0 ± 0.3 **
LA_{max} (mmol/L)	9.9 ± 1.5	10.2 ± 1.9

Averaged data are presented as mean ± standard deviation (SD), and results of the *t*-test for dependent samples, * $p < 0.05$, ** $p < 0.01$, *** $p < 0.001$ significantly different pre-training. BMI−body mass index; SBP−systolic blood pressure; DBP−diastolic blood pressure; rest−before cardiopulmonary exercise test; exerc−after cardiopulmonary exercise test; VO_2max (mL/min/kg)−maximal oxygen uptake; HR_{peak}−peak heart rate; LA_{rest}−resting lactate concentration; LA_{max}−maximal lactate concentration.

3.2. Measured Parameters

The values of the measured parameters are shown in Figure 3. At the first examination (before the training period), only B_{mean} significantly increased between the pre- (410.8) and post-exercise (449.3) measurements. At the second examination (after the training period), the values of all measured parameters significantly increased between resting and post-exercise condition. B_{mean} increased from 579.5 to 671.9, 16%; FI_{max} increased from 685.8 to 742.4, 8% and FR_{min} from 459.1 to 520, 13%. All measured parameters (both resting and post-exercise) significantly increased between the first and second examination.

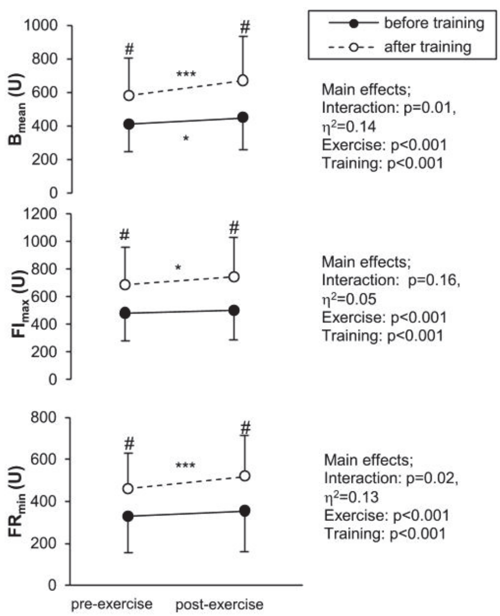

Figure 3. Measured parameters. Flow Mediated Skin Fluorescence parameters in athletes (N = 41) in two examinations, before and after the cardiopulmonary exercise test until exhaustion. B_{mean}—Changes in the mean value of the basal fluorescence; FI_{max}—Changes in maximal fluorescence during ischemia; FR_{min}—Changes in the first minimal fluorescence value during reperfusion. Values are means (SD). A two-way analysis of variance (relation between exercise and training), post-hoc Scheffe test, significant differences between pre- and post-exercise: *** $p < 0.001$, ** $p < 0.01$, * $p < 0.05$; significant differences between before and after training $^{\#}$ $p < 0.001$, ‡ $p < 0.01$, § $p < 0.05$.

3.3. Calculated Parameters

The values of the calculated parameters are presented in Figure 4. I_{max} significantly decreased after exercise in both pre- (from 72.6 to 53.9, 26% decrease) and post-training (from 106.3 to 70.6, 34% decrease) examinations. I_{max} values were higher after than before training pre- (from 72.6 to 106.3, 46% increase) and post-exercise (from 53.9 to 70.6, 31% increase).

R_{min} significantly increased after exercise compared to resting conditions in both examinations before (from 80.3 to 94.7, 18% increase) and after training (from 120.4 to 151.9, 26% increase). The pre- and post-exercise values of R_{min} were higher after than before training (pre-exercise 50% and post-exercise 60%).

The IR_{ampl} parameter did not significantly differ between resting and post-exercise conditions in both examinations. Its pre- and post-exercise values were significantly higher after than before the training period (pre-exercise from 152.9 to 226.7, 48% increase; post-exercise from 148.6 to 222.4, 50% increase).

The values of CI_{max} were significantly lower after than before exercise in both examinations (before training decreased from 0.5 to 0.4; after training decreased from 0.5 to 0.3). There were no differences observed before when compared to after training.

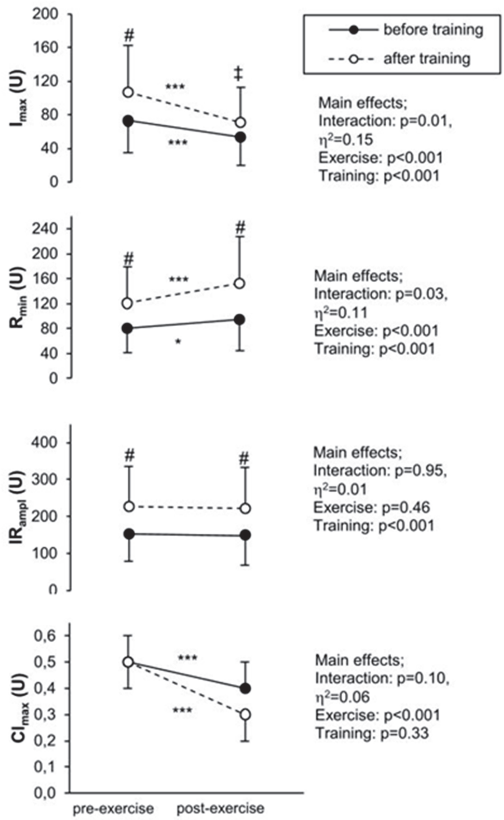

Figure 4. Calculated parameters. Flow Mediated Skin Fluorescence parameters in athletes (N = 41) in two examinations, before and after the cardiopulmonary exercise test. I_{max}—Changes in the net increase in fluorescence over the baseline during ischemia; IR_{ampl}—Changes in the amplitude of fluorescence change during ischemia and reperfusion; R_{min}—Changes in the net reduction in fluorescence below the baseline; CI_{max} – Changes in I_{max}/IR_{ampl} ratio. Values are means (SD). A two-way analysis of variance (relation between exercise and training), post-hoc Scheffe test, significant differences between pre- and post-exercise: *** $p < 0.001$, ** $p < 0.01$, * $p < 0.05$; significant differences between before and after training $^{\#}$ $p < 0.001$, ‡ $p < 0.01$, § $p < 0.05$.

4. Discussion

In this study, for the first time, the changes in NADH fluorescence in epidermal cells have been investigated in highly trained athletes before and after a training period. The main and novel finding is a significant increase in NADH fluorescence after training.

4.1. The Effect of Training

In our study, an increase in NADH fluorescence after a 7-week training period in highly trained athletes was observed. It is widely known that physical training induces several adaptations including mitochondrial adaptations [22]. The measurement of NADH fluorescence may be used to indirectly evaluate the mitochondrial function and information about its metabolic status [1,5]. However, the changes in NADH fluorescence alone do not allow us to answer the question of what particular metabolic changes took place. It is known that NAD^+ and NADH are in balance (i.e., the more NAD^+, the less NADH and vice versa [8]). Therefore, the higher post-training NADH fluorescence shown in our study may indicate increased NAD turnover.

Our participants represented different sport disciplines, but the study was only conducted in the general preparation period of the annual training cycle. The main goal of this period, regardless of sports discipline, was the development of endurance capacity. VO_2max did not change after training in our athletes, which is in line with other reports [30,31] that also did not observe such changes in highly trained athletes in an annual training cycle. However, we assume that the changes occurred at the cellular respiration level. The endurance-dominant training in all athletes significantly affected the increase in the NADH fluorescence, which can be reflected by the changes in mitochondrial functions as shown in measured NADH parameters (B_{mean}, FI_{max}, FR_{min}). The post-training increase in B_{mean}, FI_{max}, and FR_{min} suggests a training-induced increase in the total NAD pool. However, there is a lack of research on training-induced changes in skin mitochondria. We can only compare our findings with those obtained from muscle mitochondria. To the best of our knowledge, the only research on training-related changes in NAD levels was performed on rat muscles. It has been found that NAD levels increased in response to endurance training [16]. There is a lack of studies on NAD changes in trained humans. The training-related changes in mitochondria have been widely described in human muscles [19,21,22,25,32]. The training-related changes in the mitochondria are probably connected with the improvement in mitochondrial biogenesis and the removal of dysfunctional mitochondria [21,22,25,32]. After training, an increase was observed in the levels of proteins related to mitochondrial biogenesis [21,25] and an improvement in mitochondrial respiratory function [19]. It is suggested that the profile of the mitochondrial changes depends on training intensity and volume. Training volume seems to affect mitochondrial content, whereas training intensity is correlated with the improvement in mitochondrial respiration [19]. It must be remembered that exercise does not necessarily imply exactly the same metabolic changes in muscle and skin mitochondria. However, intense physical activity affects mitochondrial activity and induces an increase in NADH fluorescence, which we have shown in our previous study [5]. Therefore, the observed increase in NADH fluorescence after 7-weeks of training may indirectly indicate adaptive changes in skin mitochondria.

4.2. Exercise Response

In our recent paper [5], we showed that a single bout of exercise until exhaustion induced a significant increase in skin NADH fluorescence. The results of this study are in line with our previous observations. We found that the I_{max} parameter, related to fluorescence intensity, decreased after exercise and that the R_{min} parameter increased after exercise. The likely explanation is that with limited aerobic metabolism, NADH is accumulated and the NAD^+ amount decreases because anaerobic metabolism does not allow for restoring NAD^+ from NADH to a sufficient extent [33–35].

However, some authors [36] suggest that the decrease in NADH fluorescence intensity during reperfusion not only shows the change in mitochondrial function, but also in microcirculatory and endothelial functions related to the efficiency of blood supply to the skin. Both the skin blood vessels' thermoregulatory [37–39] and endothelial [40] functions improved after training. Our study supports this view and suggests improvements in exercise tolerance based on NADH fluorescence measurement.

4.3. Practical Application

The FMSF method might be useful to evaluate metabolic adaptations related to mitochondrial function and/or microcirculatory function as the effect of training (training efficiency). This might also be used to observe the recovery after exercise when returning to the resting NADH values.

5. Conclusions

Athletes showed significant changes in NADH fluorescence in skin cells after a 7-week training period. We found that they achieved higher post-training values in basal NADH fluorescence (B_{mean}) (pre-exercise 41% increase and post-exercise 49% increase). Additionally, the maximal increase in fluorescence during occlusion (FI_{max}) and the maximal drop in fluorescence after reperfusion (FR_{min}) were higher at rest and post-exercise after training (FI_{max} 42% at rest, and 47% post-exercise, FR_{min} (39% at rest, and 47% post-exercise). In conclusion, physical training results in an increase in the skin NADH fluorescence levels at rest and after exercise in highly trained athletes. We suggest that the measurements can reflect the training-induced changes in the metabolic status of the skin mitochondria.

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7

Analysis of Serve and Serve-Return Strategies in Elite Male and Female Padel

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Abstract: This aim of this study was to analyze serve and return statistics in elite padel players regarding courtside and gender. The sample contained 668 serves and 600 returns of serves from 14 matches (7 male and 7 female) of the 2019 Masters Finals World Padel Tour. Variables pertaining to serve (number, direction, court side and effectiveness), return of serve (direction, height, stroke type and effectiveness) and point outcome were registered through systematic observation. The main results showed that the serving pair had an advantage in rallies, under 8 shots in women and under 12 shots in men. Statistical differences according to gender and court side were found. Female players execute more backhand and cross-court returns and use more lobs than men. On the right court, serves are more frequently aimed at the "T" and more down the line returns are executed when compared to the left side. Such knowledge could be useful to develop appropriate game strategies and to design specific training exercises based on actual competition context.

Keywords: racket sports; performance analysis; game-actions; strokes

1. Introduction

Padel is a racket sport played in pairs (2 vs. 2). The court is characterized by its completely closed girth, as a small-sized grass court (20 × 10 m) surrounded by glass and metallic mesh areas on which the ball can bounce [1]. It has become a mass phenomenon in some countries, such as Spain, and is practiced in more than 35 countries around the world [2]. A professional padel circuit has been created (World Padel Tour), with tournaments in several countries. This development can be attributed to a high interaction between players and a low intensity of actions in a low level of competition [3,4]. Accordingly, the enjoyment and motivation of the players increases, inducing a greater adherence to practice [5–7].

Investigations in padel have increased in the past few years [8]. Research on padel has been mainly focused on describing the match activity and detecting effective performance indicators [9–11]. These investigations have provided primary information such as the rally length (10 to 15 s), the most common actions in offence (volley and smash) and defense (lob), and have highlighted the advantage of the net game. In addition, padel player performance has been characterized by the ratio between winning shots and errors [12]. Furthermore, previous researches have shown gender-related differences during competition [13,14]. Higher values have been observed in play time and total time in women over men players, as well as in the number and types of strokes [10]. Therefore, the players constantly try to play in offensive positions; for which they use different behaviors and technical-tactical actions, which define different styles of play [12]. The distribution of the different types of strokes, their trajectories and their efficacy stand out among these behaviours [3–5]. The results of the studies have shown that these variables may also differ depending on the gender or the side of the court on which the padel player plays [10,11]. Hence, different performance profiles of padel could exist related to gender and game-side on court [15].

However, there is an alarming lack of investigations examining players' serve and return statistics [16]. One of the most important performance indicators in racket sports is the serve [17]. In tennis, a serve directly wins the point through an ace or indirectly because of the advantage coming from the opponent's imbalance after a great serve [18]. Thus, tennis players win about 70% of points with the first serve, this percentage being higher in men's singles than in women's [17,19]. Previous studies found that serve was more determinant in tennis doubles, likely due to the presence of the server's partner covering the net [20]. Furthermore, the serve situation could influence point or game outcome in padel, since it allows players to adopt an offensive position. In this way, winners scored about 34% more points from the net than losers [9]. However, the serve in padel is different to other racket sports, because of the rules of game. In padel, the ball cannot be beaten as hard as in tennis, and the serve must be an underhand shot from a bouncing ball hit from below waist level. In addition, the effect and the side wall can affect the serve-return shot [16]. The receiver must play an accurate shot to avoid the serving pair hitting the ball into a tactically advantageous area, so they should vary the direction and height of their return of serve [15,16].

A better knowledge of players' behavior when serving or receiving is extremely useful for developing appropriate game strategies and to design specific training exercises [21]. However, at present little is known about the relationship between the serve and return of serve in padel, other than the fact that the server tends to maintain the tactical advantage until around shot five, when the advantage has dissipated [16,22]. In addition, Zhang and Zhou [23] differentiated specific serve tactics in table tennis that were associated with higher scoring rates. Furthermore, some studies highlighted the difference between first and second serves in terms of how aggressive the return could be [24]. Given that serve and return of serve are two of the most important shots in padel, the purpose of this study was to analyze serve and return statistics in elite padel players regarding court side and gender. This information could help coaches to better understand player strategies and their efficiency in padel games.

2. Materials and Methods

2.1. Sample and Variables

The study was reviewed and approved by the Ethical Committee of the University of Murcia. The sample contained 668 serves and 600 returns of serves from 14 professional matches (7 male and 7 female) of the 2019 Finals World Padel Tour. These tournaments gathered the best pairs in the world, ensuring the highest competitive level in all the matches. A total of 32 players, 16 men (mean (SD) age: 31.18 (7.27) years; height: 181.3 (4.1) cm) and 16 women (mean (SD) age: 28.66 (6.7) years; height: 168.4 (3.7) cm) performed the matches. Variables pertaining to serve and return statistics and point outcome were included in the analysis, following the methodology adopted in other similar studies [25]:

- Serve shots were analyzed regarding the serve number (1st and 2nd serve), court side (right and left), serve direction (side wall, middle and T) and effectiveness indicator (winner, error and continuity) (Figure 1).

- Return statistics were analyzed regarding the court side (right and left), stroke direction (down the line, middle and cross court), stroke height (straight and lob), stroke type (forehand and backhand) and effectiveness indicator (winner, error and continuity) (Figure 1).
- Point outcome: strokes were classified according to the winning or losing pair of the point (serving players win and returning players win).
- Total shots: number of total shots in the points were counted.

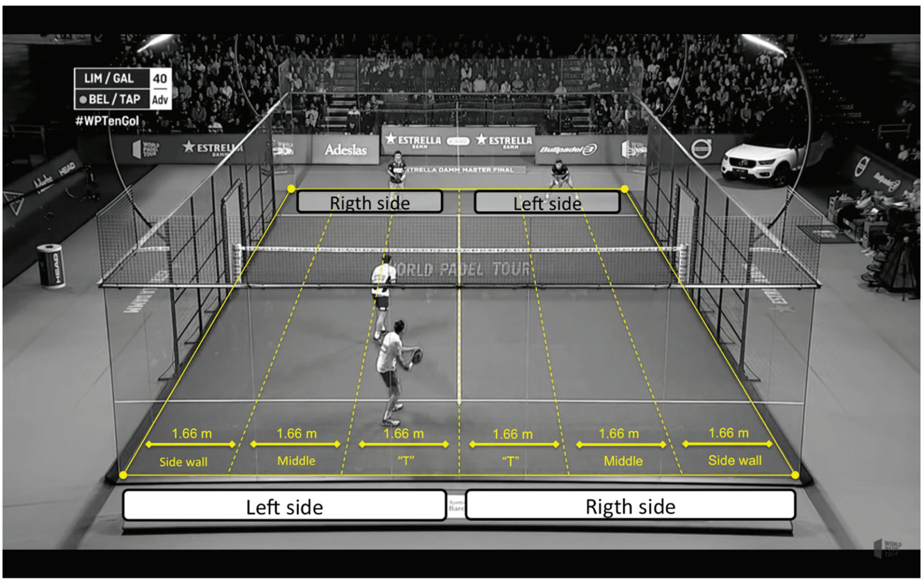

Figure 1. Illustration of court side and serve and return directions.

2.2. Procedure

The matches were downloaded from the official channel of the World Padel Tour [26]. Lince video analysis software was used to collect and register the data [27]. The Kinovea software [28] was used to place a visual grid over the video image for the serve and return directions data recording (Figure 1) and to register the feet of the serving player when impacting the ball, the place where the ball bounced after a successful serve and the direction of the ball after the returner hit it. Four observers specializing in padel (over 5 years of experience) were specifically trained to perform the recording. Observers were specifically trained in the use of the observational instrument for two weeks. The training focused on the clear identification of the variables (serve and return statistics, point outcome and total shots) and the use of the observational instrument software (Lince and Kinovea). Having completed the training process, each observer registered a training set not included in the final sample (n = 72 serves; n = 68 returns), to calculate inter-rater reliability. Consistency of records was analyzed using the free-marginal multirater Kappa [29] and the weighted Kappa [30]. The minimum score obtained was k = 0.87. Finally, intra-observer evaluation was done at the end of the observation process by Cohen's Kappa calculation, yielding a very good strength of agreement with scores over 0.92 [31].

2.3. Data Analysis

Descriptive analysis included means, standard deviations and frequencies. Assumptions of normality and homogeneity of variances were verified using the Kolmogorov–Smirnov test and Levene's test. Due to data not following a normal distribution, non-parametric tests were implemented [32].

Chi square analysis was performed to identify differences in serve and return statistics and point outcome between gender and court side. Column proportions were compared using Z tests on serve and return statistics according to the gender of the players and court side. A significance level of $p < 0.05$ was established which was adjusted according to Bonferroni in the Z tests. The associations among the categories of the variables were performed with corrected standardized residuals (CSR). The effect size was calculated using Cramer's V [33]. Rho Spearmen was used to know the relationship between serving point won and the number of strokes per point. IBM SPSS 25.0 Statistics for Macintosh (Armonk, NY: IBM Corp.) was used to process the data.

3. Results

3.1. Serve and Return Performances of Professional Padel Players Regarding Gender

Table 1 shows differences in serve and return statistics in relation to players' gender. With regards to serve performance, the players' gender determined the percentage of first and second serves ($\chi^2 = 5.05$; gL = 1; CRS = 2.2; $p < 0.05$). Thus, men obtained a significantly higher percentage of successful first serves than women. Furthermore, both men and women aimed more than 60% of their serves towards the side wall. Regarding return statistics, significant differences between men and women were found with regards to direction ($\chi^2 = 9.647$; gL = 2; CRS = 3.2; $p < 0.01$), height ($\chi^2 = 9.354$; gL = 2; CRS = 2.9; $p < 0.01$) and stroke type ($\chi^2 = 4.230$; gL = 1; CRS = 2.1; $p < 0.05$). Thus, women played a significantly higher proportion of backhand or cross-court returns and used the lob more when returning than men did. Finally, the point-result variable showed how men won a significantly higher percentage of points in a serve situation than women ($\chi^2 = 11.435$; gL = 1; CRS = 3.4; $p < 0.01$).

Figure 2 shows the relationship between the percentage of points won by the couple with the serve and the number of strokes per point, in relation to the players' gender. The correlation test results showed a significant relationship for male ($p < 0.001$; $r = 0.62$) and female ($p < 0.001$; $r = 0.54$) players between the percentage of points won in the serve and the number of strokes per point. Thus, the percentage of points won by the player with the serve went down as the number of strokes went up. Furthermore, with regards to gender, serving advantage was lost after the 12th stroke for men, while for women it was after the seventh stroke.

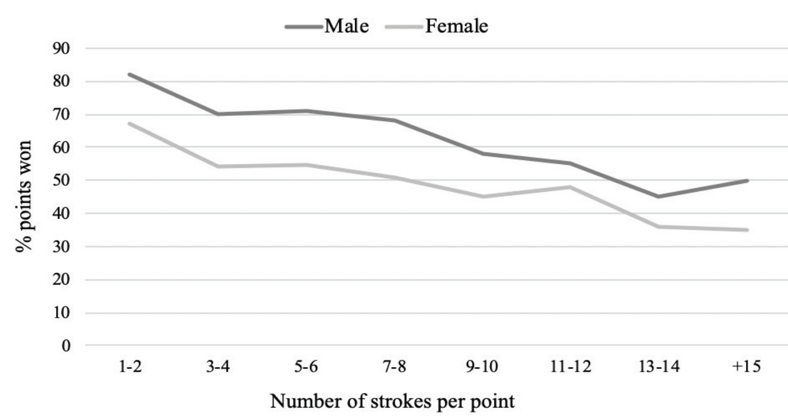

Figure 2. Percentage of points won by the serving couple with relation to the number of strokes per points: gender differences.

Table 1. Percentages for the serve and serve–return performances of the professional male and female padel players.

		Male *n* (%)	Female *n* (%)	Sig.
Serve statistics				
Serve Number				
	1st serve	378 (92.9)a	229 (87.7)b	0.025 *
	2nd serve	29 (7.1)a	32 (12.3)b	
Court side				
	Right	211 (51.8)	133 (51.0)	0.823
	Left	196 (48.2)	128 (49.0)	
Serve direction				
	Side wall	263 (64.6)	163 (62.5)	
	Middle	45 (11.1)	19 (7.3)	0.101
	T	99 (24.3)	79 (30.3)	
Effectiveness				
	Winner	0 (0.0)	0 (0.00)	
	Error	36 (8.8)	34 (13.0)	0.085
	Continuity	371 (91.2)	227 (87.0)	
Return statistics				
Stroke direction				
	Down the line	213 (57.4)a	103 (45.0)b	
	Middle	95 (25.6)	69 (30.1)	0.008 **
	Cross court	63 (17.0)a	57 (24.9)b	
Stroke height				
	Straight	242 (65.2)a	122 (53.3)b	0.009 **
	Lob	129 (34.7)a	107 (46.7)b	
Stroke type				
	Forehand	156 (42.0)a	77 (33.6)b	0.040 *
	Backhand	215 (58.0)a	152 (64.4)b	
Effectiveness				
	Winner	0 (0.0)	1 (0.4)	
	Error	20 (5.4)	9 (3.9)	0.323
	Continuity	351 (94.96)	219 (95.6)	
Point outcome				
	Serve pair win	232 (62.5)a	111 (48.5)b	0.001 **
	Returner pair win	139 (37.5)a	118 (51.5)b	

Note: n = Number; % = Percentage; * = $p < 0.05$; a, b = significant differences indicated in the Z tests for comparison of column proportions from $p < 0.05$, adjusted according to Bonferroni.

3.2. Serve and Return Performances of Professional Padel Players Regarding Court Side

Figure 3 shows serve statistics with regard to court side where said serve was played. As may be observed in the court shown above, court side significantly determined serve direction ($\chi^2 = 18.202$; gL = 2; CRS = 3.3; $p < 0.01$). It may be observed that most serves went towards the side wall, followed by the "T" and, in smaller proportion, the middle of the court. Furthermore, on the left side (ad court) players executed 12% more serves towards the side wall, whereas on the right side (deuce court) players served 14% more towards the "T." On the other hand, no significant differences were found with regards to court side for effectiveness ($\chi^2 = 1.047$; gL = 1; $p > 0.05$) and serve number ($\chi^2 = 2.972$; gL = 1; $p > 0.05$).

Figure 4 shows return statistics with regards to the player's side of the court. As may be observed, players executed around 60% of straight returns, with no significant differences regarding court side ($\chi^2 = 2.048$; gL = 2; $p > 0.05$). Furthermore, players obtained a high percentage of return effectiveness, with more than 90% of successful returns, and no difference between the right and left sides ($\chi^2 = 4.444$; gL = 2; CRS = 3.0; $p > 0.05$). Playing side significantly determined the return's direction ($\chi^2 = 28.711$; gL = 2; CRS = 7.6; $p < 0.01$). Thus, players returning from the left side executed almost 15% more down the line returns than players on the right side. Furthermore, the kind of returning stroke also showed significant differences regarding court side. Left side players executed more than 75% of their returns backhand, whereas right side players registered more balanced values, although they did execute more forehand returns. Finally, there were no significant differences between return statistics and stroke type, height or direction ($p < 0.005$).

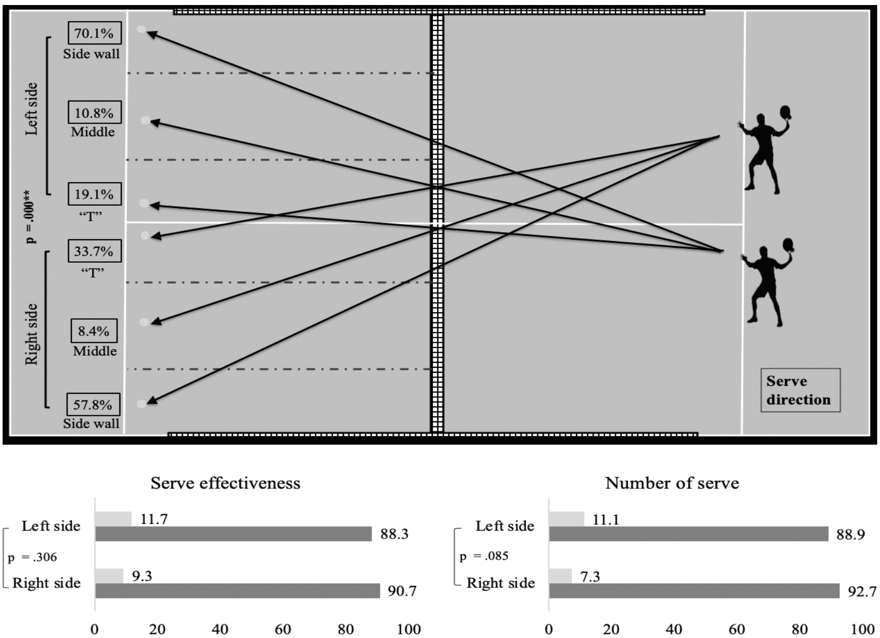

Figure 3. Serve statistics regarding court side.

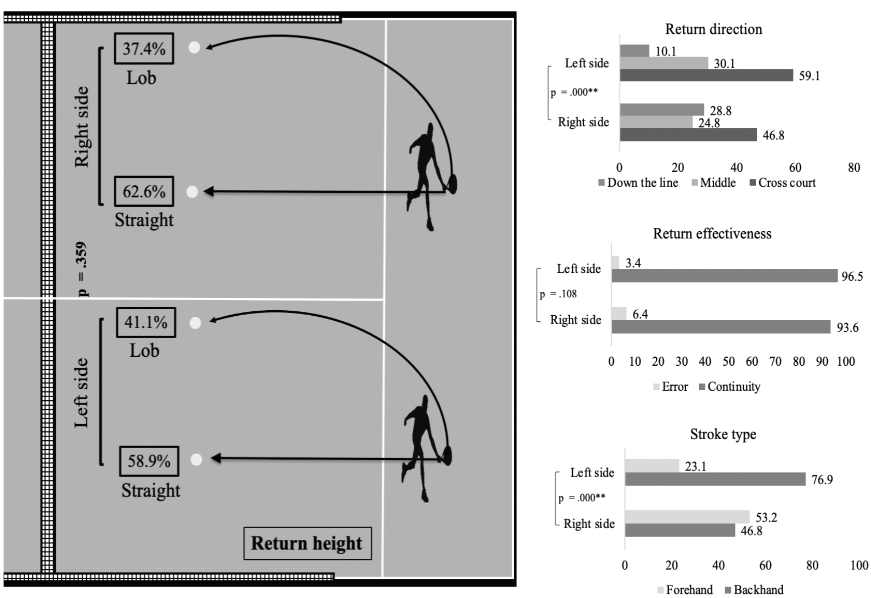

Figure 4. Serve return statistics regarding court side.

4. Discussion

This aim of this study was to evaluate the serve and return statistics in elite padel players regarding courtside and gender. The notational analysis of this research is one of the most important contributions because of the lack of previous research regarding this aspect in padel. The main results showed that the serving pair had a significant advantage in rallies, which lasted until shot 7 in women and shot 12 in men (Figure 2). Considering that serve advantage is lost after the fourth shot in tennis, this finding probably reflects the nature of padel in that it is much harder to play a winning shot, due to the court dimensions and structure, meaning that when a pair is dominating the rally, as at the start of the point when serving, it often takes more shots to finish the rally compared to tennis [16]. Regarding gender, men won a significantly higher percentage of points when serving than women (62.5% vs. 48.5%), as happens in other sports like tennis, where men obtain 14% more points with their serve than their female counterparts [34,35]. These gender differences may be due to the fact that male players are taller and can jump higher, which would enable them to sustain for a longer period of time an offensive position at the net [36,37]. However, deeper analysis and complementary variable collection is encouraged for a more relevant advance in this knowledge. In tennis, previous studies reported that men's serves had an impact on rally outcome in rallies that lasted four shots [22]. This impact of the serve was corroborated by other research, which found that the serving pair won the most rallies containing 1 to 4 shots [38]. These different results in padel suggest that the influence of service extends some way into the rally in padel because the predominance of the "serve and volley" strategy allows the serving pair to move close to the net first and adopt an offensive position [9,39]. Despite serve advantage being lost when the return players use technical actions that facilitate a change of position, such as lobs, this transition from offensive to defensive position only appears in 37% of points [40]. On the other hand, the results showed first serve effectiveness as being close to 90%. Even though there were no gender significant differences, the higher occurrence of first service faults reported by women could be due to a higher predisposition to obtain a direct point by forcing the first service [15].

With regards to serve direction, the results of this study showed how players served primarily towards the side wall, increasing that percentage on the left side (advantage), which is usually the place where more decisive points are fought. The greater distance covered by the player serving during the point [39] could explain this crossed direction of the serve towards the glass, since it would allow him to buy more time to occupy a better position at the net. Furthermore, the bounce on the side wall may complicate the return stroke, which could cause a greater number of errors, as other authors have claimed [15]. Other studies reported that the high percentage of serves to the side wall on the left side could be explained by the hand-dominance of the players [12]. Thus, since most padel players playing on the left side of the court are right-handed, servers would seek to serve towards the side wall to seek both the backhand of the return as well as the uncertainty of the wall bounce. No differences between genders were found in serve direction. Similar studies in other racket sports, such as tennis, showed that the serve aimed at the "T" was the most effective in winning the point [41]. However, on the left side of the court, better results are obtained when players serve cross-court or open [42]. It is important to highlight that some of these results regarding serve direction could be related more to players' hand dominance than court side [12], so further research is warranted.

The results of this study confirmed that the beginning of each point in professional padel seems very important and decisive for increasing the chances of winning the point. Then, serve effectiveness is directly related to the opponent's serve-return skills [17]. The results showed that a very high percentages of serve returns stayed in (around 90%), but no differences regarding gender or courtside were found. This effectiveness percentage is higher than in other racket sports such as tennis, where serve power is higher [43]. Furthermore, return height and direction in padel may allow couples in the defensive position to execute a stroke that allows them to send the attacking couple to the back of the court [11]. Previous authors showed how sending deep lobs to the corners and close to the walls will keep the rivals far from the net. However, the results of our study showed that players hit about 60% straight and 40% lob shots. These data are confirmed by a previous analysis in a national padel competition, and reflect the importance for the player serving to run toward the net and be able to approach the net to make the straight return more advantageous [16]. Furthermore, the return's height and direction showed significant differences with regards to players' gender and the side of the court. Thus, women executed more lobs and cross-court returns than men. This higher use of lobs by female players has been confirmed in previous studies that suggest a more defensive playing style among women [10,44,45]. Thus, because over-head strokes (smash and tray) are the most successful shots during a match, with a significantly higher percentage in the male category [10,46], players returning the serve have to use the lob only in comfortable positions to overcome their opponents at the net, since a poorly executed lobbed return could have as a response a winning smash from the serving players. At the national level, the lob return of serve achieved long rallies 48.8% of the time for good depth (around 5.5 m beyond the net) off first serves and 61.8% of the time off second serves, and 79.8% of the time for excellent depth (around 8.5 m beyond the net) irrespective of serve [16].

With regards to playing side, players returning the serve on the left side stroke more with their backhand than forehand, and play more straight and down the line strokes than players on the right side, corroborating the findings of Torres-Luque et al. [10], who found that about 75% of serves were directed to the backhand of players. This fact could be due to a higher game aggression of players when resting in the left side, not allowing serving players to take the lead in the point [9,47]. Furthermore, these differences in return with regards to playing side are especially important considering that 75% of the decisive points are played on the left side of the court [35].

Although this is the first study addressing serve and return statistics in professional padel, the study presents some limitations. First, contextual variables such as match status were not registered. Given the influence of situational variables on game performance [48], it would be very interesting to include such information in future research on padel. On the other hand, other variables that may affect serve and return statistics, such as serve speed or spin and players' hand dominance [12,22,43], have not been taken into account. Finally, the sample was limited, so future studies should analyze a greater number and tournaments and padel players.

5. Conclusions

The current investigation has described the advantage of serving in padel by comparing points won by servers and receiving players after a different number of shots within rallies. Given the server has a significant advantage, the aim of the return is to avoid the serving pair winning the rally quickly. This could be best achieved by good depth on lobs, regardless of the direction, and pace on straight shots, predominately aimed toward the server [16]. Statistical differences according to gender and court side were found. Female players execute more backhand and cross-court returns and use more lobs than men. On the right court, serves are more frequently aimed at the "T" and more down the line returns are executed when compared to the left side. Such knowledge may have implications for accuracy and the quality of training drills based on specific technical–tactical demands.

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8

Biomechanics of Table Tennis: A Systematic Scoping Review of Playing Levels and Maneuvers

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Abstract: This present study aims to review the available evidence on the biomechanics of table-tennis strokes. Specifically, it summarized current trends, categorized research foci, and biomechanical outcomes regarding various movement maneuvers and playing levels. Databases included were Web of Science, Cochrane Library, Scopus, and PubMed. Twenty-nine articles were identified meeting the inclusion criteria. Most of these articles revealed how executing different maneuvers changed the parameters related to body postures and lines of movement, which included racket face angle, trunk rotation, knee, and elbow joints. It was found that there was a lack of studies that investigated backspin maneuvers, longline maneuvers, strikes against sidespin, and pen-hold players. Meanwhile, higher-level players were found to be able to better utilize the joint power of the shoulder and wrist joints through the full-body kinetic chain. They also increased plantar pressure excursion in the medial-lateral direction, but reduced in anterior-posterior direction to compromise between agility and dynamic stability. This review identified that most published articles investigating the biomechanics of table tennis reported findings comparing the differences among various playing levels and movement tasks (handwork or footwork), using ball/racket speed, joint kinematics/kinetics, electromyography, and plantar pressure distribution. Systematically summarizing these findings can help to improve training regimes in order to attain better table tennis performance.

Keywords: kinematics; kinetics; table tennis; racket

1. Introduction

Table tennis is a competitive sport which requires technical preparation, tactics, as well as mental and motor training [1]. Players with higher technical capability demonstrate good coordinated movement with controlled strike power, which yield adequate speed and spin on the ball in limited decision time [2,3]. To master the stroke, professional players have to rotate the trunk efficiently and place excellent foot drive in response to various ball conditions [2]. Whole-body coordination plays an important role in table tennis, as the biomechanics of lower extremities is closely related to the upper limb performance [4]. An incorrect technique would alter movement mechanics and thus joint loadings that are related to risk potential of injury. A retrospective study found that about one-fifth of table tennis players suffered from shoulder injuries [5]. Although numerous studies had investigated the biomechanics of table tennis maneuvers, their methods and protocols were generally inconsistent. Therefore, direct comparison across studies is not feasible. Furthermore, players of different skill levels may perform different table tennis maneuvers with unique techniques and patterns. To identify the common characteristics of higher-level players, an investigation has to be conducted properly mapping playing levels with different maneuvers. Such information can help in designing sport-specific training programs in table tennis.

Biomechanical reviews of various sports, such as football [6,7], tennis [8,9], and swimming [10–12] have identified strategies to improve sports performance and prevent injuries. While previous review articles summarized physiological demands of table tennis players [13,14], conducted match analysis [15–17], and reviewed contemporary robot table tennis [18,19], there have been no sufficient reviews on the biomechanics of table tennis. There was an article reviewing the science (including biomechanics) of major racket sports [20], however its focus was not on limb movements and the joint loading of different skill levels.

A systematic scoping review accounts for the published evidence over a broad topic by summarizing, mapping, and categorizing key concepts that underpin a particular research area using a systematic protocol [21]. Such a review looks into the literature which has demonstrated high complexity and heterogeneity. The objective of this systematic scoping review was to identify recent advances in testing protocols, variables, and biomechanical outcomes regarding table tennis maneuvers and performance. The scope of sports biomechanics in table tennis is board, which has not been comprehensively reviewed. The objectives of this review were guided by the following research questions:

1. How was the biomechanics of table tennis movements analyzed?
2. What were the biomechanical differences between higher- and lower-skilled players?
3. What were the biomechanical differences among various table tennis maneuvers?

The principle focus or concept of this review pertained the categorization of biomechanical variables while the primary context was to summarize the playing skill levels and maneuvers. This study can contribute to the field of sports science by identifying key ideas for performance improvement and identify research gaps in table tennis.

2. Materials and Methods

The searches of the scoping review were designed and conducted by the first author. The first author and the third author conducted the abstract and full-text screening, and data extraction. Any disagreements were resolved by seeking consensus with the second author, and all authors conducted a final check of the review. Electronic literature searches of electronic databases, including ISI Web of Science (excluding patents, from 1970), Scopus (from 1960), and PubMed (from 1975), were performed on 13 July 2020.

The searches were conducted using the keywords “table tennis” AND the terms “biomechan*” or “kinematics” or “kinetics” in the topic field, but NOT “catalyst”, “catalysis”, “enzyme”, “biochemistry”, “oxidase”, “acid”, “biochemistry”, “colorimetric”, or “nanocomposite” to rule out a similar topic in biochemistry. The titles, abstracts, and then full-text of the papers were screened based on the following inclusion criteria: (1) published in English; (2) research article in peer-reviewed journals; (3) biomechanical studies on table tennis with experiments involving adult players; (4) original research articles either case-control or longitudinal studies investigating playing levels or differences in maneuvers. Studies were excluded if the articles (1) did not consider any table tennis moves, (2) considered participants with disability, musculoskeletal problems, or rehabilitation, (3) only considered physical, psychological attributes or tactics, (4) were not original peer-reviewed articles, (5) studied table tennis robots, or (6) used simulations or theoretical models. The searching selection process is summarized in Figure 1. There was no disagreement among authors in the selection of studies eligible for the review. The following information was extracted: bibliographic details, sample size, characteristics of participants, inclusion and exclusion criteria of studies, and experimental settings.

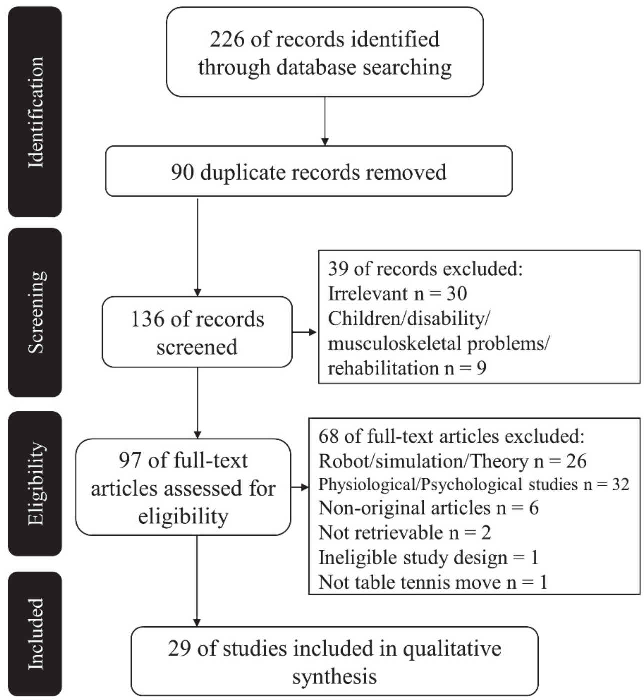

Figure 1. Flowchart of the systematic search and selection process.

3. Results

3.1. Search Results

An initial search identified 226 studies. After pooling the results and removing duplicates, 136 articles were screened for titles and abstracts. Finally, there were 29 studies successfully meeting the eligibility criteria (Figure 1). The studies were excluded because they were irrelevant (n = 30); they involved players with disabilities, musculoskeletal problems, or children (n = 9); they used robotic players, simulations or theoretical calculations (n = 26); they ocused on psychological issues, tactics, decision-making, coaching, cardiopulmonary or metabolic assessments (n = 32); they were survey, conference paper, review, and expert comment papers (n = 6). One study did not fall into the inclusion criteria of study design whilst another study did not examine any table tennis move. The full-text of one article could not be retrieved because it was too old and the journal was closed down [22]. One study was not retrievable with the given digital object identifier (DOI) [23].

The participant characteristics and study designs of the 29 included articles are summarized in Tables 1 and 2, respectively. In brief, participant characteristics, test protocols, and outcome variables of each article were summarized according to playing levels (n = 12), movement tasks (handwork, n = 6; footwork, n = 4; ball/serve against, n = 8) and other factors (n = 4) to identify performance determinants. Six included studies considered multiple factors on different servings with handwork [1,2] or playing level [24,25], racket mass with ball frequency [26], and footwork with footwear [27]. Furthermore, the categorization of dependent and independent variables are mapped in Figure 2. Key findings of the included studies related to playing levels and maneuvers are provided in Tables 3 and 4.

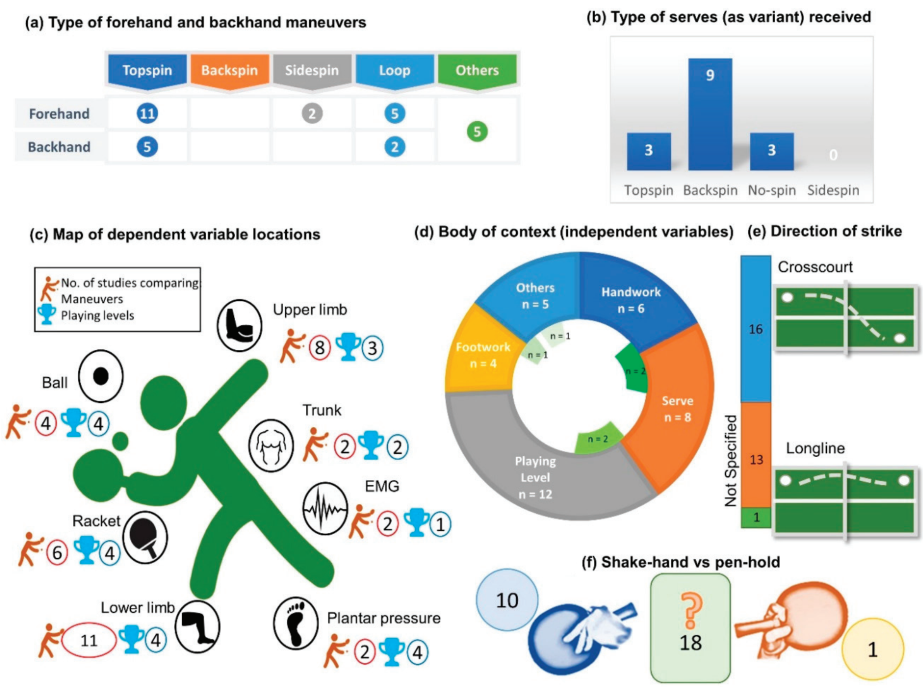

Figure 2. A scoping review map summarizing: (**a**) types of forehand and backhand maneuvers; (**b**) types of serves (as variant) to hit back; (**c**) map of dependent variables comparing the number of studies between topics on maneuvers and playing levels; (**d**) body of context (independent variables), the n-values in the interior circle denote number of studies with multiple independent variables between or within the factors stated on the exterior circle; (**e**) direction of strike; and (**f**) shake-hand vs pen-hold.

3.2. Classification of Movement Stage/Phase

While some included studies adopted the maximum or average values of performance outcome of strokes, the majority of the studies divided stroke into movement sub-phases or targeted to selected instants for subsequent analysis. Typically, the stroke was classified into backswing and forward-swing phases, targeted on the specific time points at the termination, backward-end and forward-end [1,3,28–34]. A few included studies [2,24,26,33,35,36] focused on the instant at ball impact which was used to determine the velocity of the racket and ball, while some other included studies investigated the biomechanics at pre-impact and post-impact stages [24,36–38], and over a longer period of time before and after the instant of ball contact [1,2,38,39]. Some included studies endeavored that pelvic and hip rotations were correlated with the racket velocity at impact and thus focused on the starting time of the pelvic forward rotation [36,37]. To sum up, the included studies often investigated the biomechanical parameters at the instant of ball or racket impact as well as the maximum or average value during the time before and after the ball impact.

3.3. Ball and Racket Performance

Eight included studies examined the effects of ball and racket mechanics as well as serve techniques on table-tennis performance [1,2,24–26,33,35,37], and some of these studies also compared the influences of different handworks [1,2] and playing levels [24,25]. Common variants included the type of ball spin [1,2,33,35,37,39] and the spin rate [24,25]. Moreover, seven included studies investigated ball, racket, and serve as outcome measures instead of variants [31,38,40–45].

Ball speed, accuracy, and repeatability were suggested to be the key indicators of playing level. Ball speed and accuracy were significantly correlated with player ranking in a competition [43]. Higher-level players produced higher ball speed and accuracy, which could be due to significantly shorter duration and variability of duration in the forward swing phase [31,32,38,41]. However, Iino and Kojima [24] found that racket speed at impact was not significantly different between playing levels (advanced vs. intermediate), although players with higher-level can rotate the trunk effectively to produce a greater racket acceleration at ball impact. Yet, Iino and Kojima [24] imposed a stringent significance level using a Bonferroni correction. Similarly, Belli et al. [40] found that while there was only a slight difference in ball speed comparing higher and lower-level players, players with higher-level demonstrated higher accuracy of ball target placement and made fewer errors in training and competition. On the other hand, inexperienced players showed higher inconsistency in ball speed and accuracy during within- and between-day trials [43]. Compared to the intermediate players, advanced players showed smaller variance of joint angle that affected the racket vertical angle during forehand topspin stroke [41]. Furthermore, a lower variability in the racket orientation and movement direction could be the reason for more successful returns and higher accuracy of the ball bounce location [38]. An uncontrolled manifold analysis suggested that higher-level players exploited higher degree of redundancy to maintain a similar racket angle at ball impact [41]. In brief, higher-level players exhibited higher accuracy and reproducibility on ball and racket mechanics but may not necessarily produce higher ball speed than lower-level players.

Compared to the topspin serves, returning backspin serves demonstrated significantly higher resultant and vertical racket velocities at ball impact [35,37], which could be contributed greatly by the wrist extension [35]. A possible explanation for this is that backspin serves tend to be treated back-low owing to the spin, resulting in a greater upward velocity of the shoulder joint center [37]. Moreover, peak shoulder torques in all directions, as well as elbow valgus torques, were significantly larger against backspin, in addition to the peaks of upper trunk right axial rotation and extension velocities [37]. Returning a spinning ball also alters the moving distance and velocity of the racket in the upward-downward direction, as compared to an ordinary stroke or a stroke with higher power. Hitting back a backspin serve could be more demanding than a topspin serve.

In addition, biomechanical differences between returning light and heavy backspin serves were assessed by two included articles from the same research group [24,25]. They produced different rates of ball backspin (11.4 vs. 36.8 revolutions/s) for light and heavy spin conditions. The heavy spin would direct the racket face to be more open [24]. Furthermore, their results found higher maximum loading at elbow and shoulder joints which might result in higher work done at the racket arm [25]. However, higher-level players showed a higher amount of energy transfer of the elbow for a light spin compared to intermediate players, but the opposite was true for the heavy spin [25], implying significant interaction effect between ball spin and playing level. The influence of racket mass and ball frequency were investigated by Iino and Kojima [26], who suggested that a heavier racket could impose higher demand on wrist extension torque, but did not influence trunk and racket arm kinematics and kinetics. A frequent ball serve could result in a lower racket speed at impact possibly since the pelvis and upper trunk rotations were not responsive enough. Table tennis players managed to identify the differences in ball spin, frequency, and mass, and accommodated by tilting the racket face angle and adjusting the power output of upper extremity.

Table 1. Participant characteristics of reviewed studies.

Author (Year)	Participants Information Sample Size; Age (years); Height (cm); Weight (kg)	Group/Level *	Inclusion Criteria (IC)/Exclusion Criteria (EC)
Bankosz and Winiarski (2017) [1]	n = 12F; 20.0 (5.5);167.2 (6.9); 55.3 (6.2)	Players in high-level sports training and performance	IC: 1st 16 in their category of age; EC: NS
Bankosz and Winiarski (2018) [2]	n = 10F; 16.0 (2.5); 165 (6); 54.4 (3.2)	Junior elite players	IC: Top 16 junior players EC: NS
Bankosz and Winiarski (2018) [39]	Junior, n = 4F; 18.0 (0.5); 167.7 (5.7); 52.0 (3.6) Senior, n = 6F; 24.8 (3.2); 168.3 (6.3); 64.5 (2.4)	Junior and senior high sport skill players	IC: Top 16 TT players in Poland. EC: NS
Bankosz and Winiarski (2020) [33]	n = 7M; 23 (2); 178 (3); 76.5(8)	Top-ranked international players	IC: Top 10 TT players in Poland. EC: NS
Belli et al. (2019) [40]	Local, n = 9M; 24.3 (2.6); 174.6 (3.3); 68.1 (5.7); Regional, n = 10M; 23.9 (1.8); 176.9 (2.1); 79.8 (3.1)	Local group: 2.2 (0.3) yExp, 3.2 (0.5) hrWTR egional group: 7.5 (0.9) yExp, 10.0 (0.9) hrWT	IC: Local: low experience, w/o participation in tournaments; Regional: <5 years training, completed regional and national tournament
Fu et al. (2016) [3]	Intermediate, n = 13M; 21.2 (1.6); 175.2 (2.4); 69.1 (4.1); Superior, n = 13M; 20.1 (0.9); 174.8 (2.5); 66.9 (5.1)	National level Intermediate: (Div. II) 10.2 (1.9) yExp Superior: (Div. II) 13.4 (1.2) yExp	IC: NS EC: Previous lower extremity and foot disease or deformity, injury in the last 6 months
Ibrahim et al. (2020) [44]	n = 16M; 21.5 (1.27); 168 (56); 61.59 (8.60)	Collegiate players, min 3 yExp	IC: right-handed and shake-hand grip EC: NS
Iino et al. (2008) [35]	n = 11M; 21.1 (4.4); 171 (7); 66.3 (8.1)	International and collegiate players	IC: Shakehand grip attacking players EC: NS
Iino and Kojima (2009) [24]	Intermediate, n = 8M 20.6 (1.5); 170 (8); 59 (5.7) Advanced, n = 9M 20.6 (1.2); 171 (6); 66.2 (9.5)	Intermediate 7.4 (1.8) yExp Advanced 11.2 (0.8) yExp	IC: Intermediate: not qualified for national tournaments, Division III collegiate Advanced: qualified for national tournaments, Division I collegiate EC: NS
Iino and Kojima (2011) [25]	Intermediate, n = 8M 20.6 (1.5); 170 (8); 59 (5.7) Advanced, n = 9M 20.6 (1.2); 171 (6); 66.2 (9.5)	Intermediate7.4 (1.8) yExpAdvanced11.2 (0.8) yExp	IC: Intermediate: Div. III collegiate Advanced: Div. I collegiate EC: NS
Iino and Kojima (2016) [26]	n = 8M 20.6 (1.3); 170 (4); 63.1 (5.7)	Advanced players 13.0 (1.7) yExp	IC: Div. I collegiate team in Kanto Collegiate TT League in Japan; Offensive players; use shake hands grip rackets; EC: NS
Iino and Kojima (2016) [37]	n = 10M 20.6 (1.3); 171 (5); 61.6 (5.7)	Advanced skill players 12.8 (2.4) yExp	IC: Qualified for national level TT competitions in high school or college; EC: NS
Iino et al. (2017) [41]	Intermediate, n = 8M 20.9 (0.9); 173 (7); 62.5 (6.3); Advanced, n = 7M 20.4 (1.3); 172 (7); 65.3 (5.4)	Intermediate (Div. III) 7.8 (1.0) yExp Advanced (Div. I) 11.3 (2.2) yExp	IC: Intermediate: not qualified for national tournaments Advanced: qualified for national tournaments EC: NS
Iino (2018) [36]	n = 18M; 20.7 (1.1); 171 (5); 64.0 (7.6)	Advanced players 12.2 (2.2) yExp	IC: Div. I or II collegiate players EC: NS
Lam et al. (2019) [4]	n = 15M; 23.6 (2.2); 180 (4); 72.3 (6.2)	Div. I players	IC: NS; EC: lower extremity injury in the last 6 months
LeMansec et al. (2016) [43]	Inexperience, n = 18M 19.5 (0.9); 176.9 (5.9); 69 (6.4); Advanced, n = 14M; 30.7 (11.3); 178.3 (6.2); 74 (12.3); Expert, n = 20M; 28.4 (6.7); 178.9 (6.2); 74.5 (9.7)	Inexperience Advanced: 13.4 (5.6) yExp 4.1 (2.3) hrWT Expert: 19.8 (6.8) yExp 10.4 (7.9) hrWT	IC: Inexperience: students w/o experience in TT; not ranked in the Federation of TT; Advanced: participated in regional championship; Expert: participated in National or international competition; EC: NS
LeMansec et al. (2018) [46]	n = 14M; 27.1 (4.9);177.5 (5.3); 73.5 (8.4)	National level players 4.7 (1.9) hrWT	IC: Official competition players in the national championshipEC: Lower limb pain in last 2 years

Table 1. *Cont.*

Author (Year)	Participants Information Sample Size; Age (years); Height (cm); Weight (kg)	Group/Level *	Inclusion Criteria (IC)/Exclusion Criteria (EC)
Malagoli Lanzoni et al. (2018) [45]	n = 7M; 22.2 (3.2); 177.4 (4.2); 72.9 (11)	Competitive player: 10.2 (2.5) yExp	IC: 1st and 2nd national league players and ranked among 1st 200; EC: Consume caffeine last 4 h
Meghdadi et al. (2019) [47]	Healthy, n = 30M; 24 (2.59); 176 (7.81); 74 (5.82); Syndromic, n = 30M; 25 (2.29); 174 (7.06); 75 (5.50)	National-level players: Healthy: 5 (2.11) yExp; Syndromic: 6 (1.97) yExp	IC: top 100 list of Federation and active in League; right-handed. Syndromic: impingement on dominant side; EC: History of shoulder dislocation, surgery, occult/overt instability, symptoms on cervical spine, rotator cuff tendinitis, documented injuries/pathology to shoulder
Qian et al. (2016) [28]	Intermediate, n = 13M 21.2(1.6); 175.2(2.4); 69.1 (4.1); Superior, n = 13M 20.1 (0.9); 174.8 (2.5); 66.9 (5.1)	Intermediate (Div. III) 10.2 (1.9) yExp Superior (Div. I) 13.4 (1.2) yExp	IC: NS EC: Lower extremity and foot disease or deformity, injury for the last 6 months
Shao et al. (2020) [34]	Amateur, n = 11M; 20.8 (0.6); 174.2 (1.4); 62.4 (3.5) Prof., n = 11M; 21.6 (0.4); 173.5 (1.7); 63.7 (4.2)	Amateur: university students: 0.4 (0.2) yExp; Prof.: Div. I players: 14.2 (1.4)	IC: right-handed, Prof.: Div. I players; EC: any previous lower limb injuries and surgery or foot disease for at least 6 months
Sheppard and Li (2007) [38]	Novice, n = 12(NS); 22.2 (5.6); NS; NS; Expert, n = 12(NS); 21.7 (2.9); NS; NS	Novice: university population; Expert: table tennis club and sports center players	IC: right-handed, normal or corrected vision; Expert: at least years of experience and play at least 2 h per week; EC: no physical impairment
Wang et al. (2018) [29]	Amateur, n = 10M Elite, n = 10M NS; NS; NS	NS	IC: NS; EC: lower extremity, foot diseases/deformity; Injury in the past 6 months
Yan et al. (2017) [27]	n = 8M; 21.9 (1.1); 173.1 (4.2); 62.8 (2.7)	Collegiate players	IC: right-handed, second grade EC: no history of serious injury to lower limb; did not engage in vigorous exercise 24 h before experiment
Yu et al. (2018) [30]	n = 10F 21.6 (0.3); 164 (3); 54.2 (2.8)	Advanced 15.8 (1.7) yExp	IC: Div. I players EC: NS
Yu et al. (2019) [48]	n = 12M; 20.64 (1.42); 174 (3); 67.73 (3.31)	Elite national level players	EC: No previous lower limb injuries and surgeries or foot diseases
Yu et al. (2019) [32]	Beginners, n = 9M; 22.7 (1.62); 175 (4.6); 73.7 (3.1); Prof., n = 9M; 25.5 (1.24); 175 (5.3); 74.6 (2.5)	University TT team Beginners: 0.45 (0.42) yExp; Prof.: 14.8 (1.57) yExp	EC: free from any previous lower limb injuries, surgeries or foot diseases in the past 6 months.
Zhang et al. (2016) [31]	Novice, n = 10M 23.1 (4.1); NS; NSExpert, n = 10M 24.1 (1.6); NS; NS	Novice: university population Expert: prof. from TT teams and clubs	IC: NS EC: Novice: w/o formal training
Zhou (2014) [42]	n = 18M 22.3 (1.8); 172.7 (5.1); 64.6 (5.8)	Physical education major	IC: Played table tennis for more than 5 years EC: NS

* The names of the level or group are adopted from the included studies. Numbers in brackets denote standard deviation. M: male; F: female; Number in bracket denotes standard deviation. NS: not specified; yExp: year of experience; Div: division; h: hours; hrWT: hours per week training; TT: table tennis; Prof.: professionals; w/: with; w/o: without.

3.4. Upper Limb Biomechanics

There were eight included studies targeting handwork as the variant, while two of them co-variated with different serves (Table 2). Higher racket speed and faster ball rotation were the key attributes of attacking shots and this could be determined by the kinematics/kinetics of upper extremity as well as the efficiency of energy transfer through the upper arm [25,49]. Higher-level players showed significantly larger maximum shoulder internal rotation, elbow varus, and wrist radial deviation torques, in addition to the maximum joint torque power at shoulder joint in both internal and external rotation directions [25]. Higher angular velocity of the wrist joint contributed to a higher ball and racket speed during drop shot services, while that also produced higher racket speed during long shot services [44].

Moreover, higher-level players rotated the lower trunk efficiently contributing to higher racket speed at ball impact [24]. Meanwhile, the racket horizontal velocity at ball impact was related to the hip axial rotation torque at the playing side (i.e., racket side), while the racket vertical velocity was correlated with backward tilt torques and upward hip joint forces [36]. In contrast, players with shoulder impingement syndrome had sub-optimal coordination and movement patterns of the shoulder girdle [47]. These players significantly reduced muscle activity of the serratus anterior and supraspinatus, which was compensated by increasing overall muscle activity and early activation of upper trapezius [47]. Whole-body coordination and movement would play an important role in driving a speedy ball impact.

Comparing forehand and backhand strokes, racket speed during ball impact was similar but presented differences in the upward and forward velocity components [1]. Forehand stroke lasts slightly longer duration for whole movement cycle and individual phases, and noticeably longer total traveling distance of the racket. This could be because forehand had greater body involvement while the arm and trunk range of motion (RoM) in backhand stroke is limited. Forehand stroke may produce more energy, whilst a longer backswing phase in the high-force condition may generate higher force and longer contact time with the balls [1]. The racket velocity produced by forehand and backhand strokes could be different. During forehand stroke, racket velocity was correlated with the angular velocities of internal arm rotation and shoulder adduction, whereas the racket velocity was correlated with the angular velocities of arm abduction and shoulder rotation during a backhand stroke [2].

A longline forehand topspin produced larger ball rotation, compared to the crosscourt topspin shot. At the instant of the maximum velocity of racket in a forehand topspin stroke, players put their racket more inclined whilst maintaining a more flexed knee and elbow posture, in addition to a more pronounced trunk rotation [45]. Other maneuvers including loop, flick, fast break, and curling ball were also studied [35,42,46]. Compared to curving balls, Zhou et al. [42] suggested that fast breaking significantly reduced racket speed during ball impact. While the flick maneuver was specified as an attack when the ball is closed to the net, there were no detailed explanations on the moves of the fast break and curling ball in which we believed that they could be the flick/drop shot and topspin/sidespin loop maneuvers, respectively. On the other hand, Le Mansec et al. [46] demonstrated that aggressive strokes required greater muscle activities. During smash, biceps femoris, gluteus maximus, gastrocnemius, and soleus muscles were highly activated. Forehand topspin with more power or spin produced significantly higher muscle activation of biceps femoris and gluteus maximus muscles compared to other maneuvers, including backhand top, forehand smash, and flick.

3.5. Lower Limb Biomechanics

Four included studies investigated different footwork targeting side versus cross-step [4], long versus short chasse step [48], stepping directions and friction [27], and squatting [30], as shown in Tables 1 and 4, while one study compared players of different levels performing a cross-step [34]. Lam et al. [4] identified that both side-step and cross-step footwork produced significantly higher ground reaction force, knee flexion angle, knee moment, ankle inversion and moment compared with one-step footwork, in addition to a significant higher peak pressure on the total foot, toe, first, second and fifth metatarsal regions. On the other hand, long and short chasse steps during a forehand topspin stroke were compared [48]. Long chasse steps produced an earlier muscle activation for vastus medialis, quicker angular velocity, and larger ankle and hip transverse RoM, whereas larger ankle coronal RoM and hip sagittal RoM compared with the short chasse steps [48]. A stable lower limb support base is another important attribute to tackle serve. Yu et al. [30] compared a squat serve with stand serve and found that squat serve produced larger angles and velocities of hip flexion, adduction, knee flexion, and external rotation and ankle dorsiflexion, whereas standing serve produced a higher force-time integral in the rearfoot region. Different stepping angle and footwear friction could also influence the center of mass and kinematics of knee joint, respectively [27]. Different footwork imposed different lower limb kinematics requirements for table tennis players.

Table 2. Study characteristics of reviewed studies.

Author (Year)	Variant (s)	Maneuvers/Conditions	Type of Parameters
Bankosz and Winiarski (2017) [1]	Handwork (2) × power/serve (3)	Handwork: 1. Forehand crosscourt topspin 2. Backhand crosscourt topspin Handwork power and serve: a. Strength, speed and rotation of 75% max, against no-spin serve; b. Strength, speed and rotation of 75% max, against backspin serve; c. Strength and speed close to max, against no-spin serve;	Racket kinematics
Bankosz and Winiarski (2018) [2]	Handwork (2) × power/serve (3)	Handwork: 1. Forehand crosscourt topspin 2. Backhand crosscourt topspin Handwork power and serve: a. Force, velocity and rotation of 75%, against no-spin serve; b. Force, velocity and rotation of 75%, against backspin serve; c. Force, velocity close to max, against no-spin serve;	Racket kinematics, upper and lower limb kinematics
Bankosz and Winiarski (2018) [39]	Power/serve (3)	Forehand crosscourt topspin a. Force, velocity and rotation of 75%, against no-spin serve; b. Force, velocity and rotation of 75%, against backspin serve; c. Force, velocity close to max, against no-spin serve;	Racket kinematics, lower limb kinematics
Bankosz and Winiarski (2020) [33]	Serve (2)	Forehand crosscourt topspin 1. against a topspin ball 2. against a backspin ball	Upper limb, lower limb and trunk kinematics
Belli et al. (2019) [40]	Level (2)	Forehand or backhand offensive stroke chosen by players against backspin ball 100–120 cm from net and 30 cm away from either left or right side at 25 km/h with frequency of 54 balls per min	Ball speed, accuracy, performance index
Fu et al. (2016) [3]	Level (2)	Forehand crosscourt loop	PP
Ibrahim et al. (2020) [44]	Handwork (2)	1. Forehand drop shot 2. Long shot	Ball and racket kinematics, upper limb kinematics
Iino et al. (2008) [35]	Serve (2)	Backhand crosscourt loop 1. Against topspin serve 2. Against backspin serve	Ball kinematics, Upper limb kinematics
Iino and Kojima (2009) [24]	Level (2) × Serve (2)	Forehand crosscourt topspin as hard as possible 1. Against light backspin ball 2. Against heavy backspin ball	Ball and racket kinematics, trunk and upper limb kinematics
Iino and Kojima (2011) [25]	Level (2) × serve (2)	Forehand crosscourt topspin at max effort 1. Against light backspin ball 2. Against heavy backspin ball	Kinetics of upper limb
Iino and Kojima (2016) [26]	Racket mass (3) × ball frequency (2)	Backhand topspin at max effort Racket mass (153.5 g, 176 g, 201.5 g) Ball projection frequency (75 and 35 ball per minutes)	Racket kinematics, Upper limb and trunk kinematics and kinetics
Iino and Kojima (2016) [37]	Serve (2)	Backhand crosscourt topspin at max effort 1. Against topspin serve 2. Against backspin serve	Upper limb kinetics
Iino et al. (2017) [41]	Level (2)	Forehand crosscourt topspin 1. Intermediate players 2. Advanced players	Kinematics and variability of trunk, upper limb and racket kinematics
Iino (2018) [36]	Correlation study	Forehand crosscourt topspin at max effort	Racket kinematics/kinetics and pelvis kinetics

Table 2. *Cont.*

Author (Year)	Variant (s)	Maneuvers/Conditions	Type of Parameters
Lam et al. (2019) [4]	Footwork (3)	Forehand crosscourt topspin 1. One-step; 2. Side-step; 3. Cross-step	GRF, knee and ankle kinematics and kinetics, PP
LeMansec et al. (2016) [43]	Level (3)	Forehand crosscourt topspin 1. Inexperience players 2. Advanced players 3. Expert players	Ball speed and accuracy
LeMansec et al. (2018) [46]	Handwork (5)	1. Backhand top; 2. Flick (a close to net attack); 3. Forehand spin (topspin with more spin less power); 4. Forehand top (topspin with more power less spin); 5. Smash	Lower limb muscle EMG
Malagoli Lanzoni et al. (2018) [45]	Handwork (2)	1 Forehand longline topspin 2. Forehand crosscourt topspin	Racket, upper and lower limb kinematics
Meghdadi et al. (2019) [47]	Healthy vs. syndromic (2)	Forehand topspin loop	EMG, muscle onset and offset time
Qian et al. (2016) [28]	Level (2)	Forehand topspin loop	Lower limb kinematics and kinetics, PP
Shao et al. (2020) [34]	Level (2)	Forehand loop using a cross-step with maximal power against topspin	Lower limb kinematics, PP
Sheppardand Li (2007) [38]	Level (2)	1. Forehand return aimed for speed 2. Forehand returns aimed for speed with accuracy 3. Forehand returns aimed for accuracyNote: the three conditions were not independent factors of the study	Ball speed and accuracy, racket kinematics
Wang et al. (2018) [29]	Level (2)	Backhand crosscourt loop	Lower limb kinematics and kinetics, EMG
Yan et al. (2017) [27]	Footwork (2) × Footwear (3)	Footwork: 1. 180° step 2. 45° stepSole-ground friction: a. Low; b. Medium; c. High	CoM, Lower limb kinematics
Yu et al. (2018) [30]	Footwork (2)	Stroke NS 1. Stand serve 2. Squat serve	Lower limb kinematics and kinetics, PP
Yu et al. (2019) [48]	Footwork (2)	Forehand loop 1. Short chasse step 2. Long chasse step	Lower limb kinematics, EMG
Yu et al. (2019) [32]	Level (2)	Chasse step movement and forehand loop with maximal power against topspin	Foot kinematics, PP
Zhang et al. (2016) [31]	Level (2)	Forehand crosscourt stroke 1. Novice players 2. Expert players	Accuracy, Racket kinematics
Zhou (2014) [42]	Handwork (2)	1. Fast break 2. Curling ball	Racket speed

NS: not specified; CoM: centre of Mass; w/: with; w/o: without; PP: plantar pressure distribution; EMG: electromyography.

Comparing the lower limb biomechanics among players with various playing levels, Qian et al. [28] and Wang et al. [29] reported distinct findings for respective forehand and backhand crosscourt loops. When executing forehand topspin loop, higher-level players increased knee external rotation, hip flexion and decreased ankle dorsiflexion during backward end phase, and increased hip extension and internal rotation, decreased ankle and knee internal rotation during forward end phase. There was an overall increase in the ankle sagittal RoM as well as hip sagittal and coronal RoM [28]. When performing backhand crosscourt loop against backspin ball, higher-level players increased ankle dorsiflexion, eversion and external rotation, increased knee flexion and abduction and increased hip flexion, adduction, and external rotation at the beginning of backswing, as well as increased ankle dorsiflexion, knee flexion, reduced hip flexion but increased abduction at the end of swing [29]. During cross-step footwork, higher-level players executed superior foot motor control, as indicated by a smaller RoM of foot joints and higher relative load on the plantar toes, lateral forefoot and rearfoot regions [34]. They also demonstrated smaller forefoot plantarflexion and abduction during cross-step end phase but larger forefoot dorsiflexion and adduction during forward end phase [34]. Effective coordination of lower limb facilitates better upper body rotation in higher-level players [39].

Bańkosz and Winiarski [33] compared inter- and intra-individual variabilities of kinematic parameters. They reported that both variabilities could be quite high, but players attempted to minimize variability at critical moments, such as the instant of ball impact. Higher inter-individual variability could also imply that the technique of coordination movement is rather individual. Adopting or imitating a particular training regime has to pay more attention.

Plantar pressure was also used to evaluate foot loading among different playing levels. When performing forehand loop during backward end phase, higher-level players displayed larger plantar pressure excursion in the medial-lateral direction but smaller in the anterior-posterior direction, accompanied by increased contact areas at midfoot and rearfoot regions while decreased contact area at lesser toe region [3,28]. During forward end phase, higher-level and intermediate players decreased similarly the plantar pressure excursion in the anterior-posterior direction. The contact areas were increased at midfoot, rearfoot, and forefoot regions while decreased at the hallux region [3,28]. The change of plantar pressure excursion and contact area could reflect the strategy compromising dynamic stability and agility in different directions.

Table 3. Key findings of included studies comparing playing levels.

Author (Year)	Outcome Measures	Key Findings of Higher–Level Compared to Lower–Level Players
Belli et al. (2019) [40]	Ball speed; accuracy score, performance index (average speed × accuracy/100); percentage error for ball toward target zone	↑ Accuracy score; ↑ Performance index; ↓ Percentage error.
Fu et al. (2016) [3]	ML and AP excursion; Contact area for big toe, lesser toes, medial forefoot, lateral forefoot, midfoot and rearfoot	During backward end: ↑ ML excursion; ↓ AP excursion; ↑ Contact area for midfoot and rearfoot; ↓ Contact area for lesser toes; During forward end:↓ AP excursion; ↑ Contact area for midfoot, rearfoot, medial forefoot and lateral forefoot; ↓ Contact area for big toe
Iino and Kojima (2009) [24]	Ball speed before and after impact; Racket speed, face angle, path inclination and height at ball impact; Time required to reach 25% of racket speed at impact and max racket acceleration; Contributions to racket speed by: Max lower trunk axial rotation; mid hip linear; lower trunk lateral bending, flexion/extension, axial rotation; upper trunk axial rotation relative to lower trunk; shoulder linear relative to upper trunk; shoulder abduction, flexion, internal rotation; elbow flexion/extension; forearm supination/pronation; wrist palmar/dorsi flexion, radial/ulnar deviation	↑ Max racket acceleration; ↑ Contribution of lower trunk axial rotation to racket speed

Table 3. *Cont.*

Author (Year)	Outcome Measures	Key Findings of Higher–Level Compared to Lower–Level Players
Iino and Kojima (2011) [25]	Max joint torques of: shoulder adduction, flexion, internal rotation; elbow varus, flexion; wrist dorsiflexion and radial deviation; Max joint torque power of shoulder adduction, flexion, positive and negative internal rotation, elbow flexion, wrist dorsiflexion, and radial deviation; Net work done by shoulder adduction and internal rotation; Positive and negative work done by shoulder internal rotation torque; Max rate of energy transfer by: shoulder addiction and internal rotation; elbow varus and flexion; wrist radial deviation Amount of energy transfer by: shoulder adduction, flexion, internal rotation; elbow varus and flexion; wrist radial deviation; Max rate of energy transfer and amount of energy transfer through shoulder, elbow and wrist joints; Increase in mechanical energy of racket arm;Mechanical energy transferred to racket arm; Energy transfer ratio of racket arm.	↑ Normalized max joint torques of shoulder internal rotation, elbow varus, and wrist radial deviation; ↑ Max joint torque power of shoulder internal rotation in both positive and negative directions; ↑ Negative work done by shoulder internal rotation torque; ↑ Max rate of energy transfer for shoulder internal rotation, elbow varus and wrist radial deviation.
Iino et al. (2017) [41]	Racket speed at ball impact; Standard deviation of racket face angle in vertical and horizontal directions; Total, controlled and uncontrolled variable variance for racket race angle in vertical and horizontal directions; Ratio of uncontrolled to controlled variance	↑ Racket speed at ball impact; ↓ Controlled variance for horizontal angle of racket surface.
LeMansec et al. (2016) [43]	Ball speed; accuracy; performance index (average speed × accuracy/100)	Elite ↑ ball speed, accuracy and performance index than advanced players Advanced ↑ Ball speed, accuracy and performance index than inexperienced players.
Qian et al. (2016) [28]	Joint angle of ankle, knee and hip in all planes at backward-end (BE) and forward-end (FE); RoM of ankle, knee and hip joint in all planes.ACR of ankle, knee and hip in all planes during forward–swing phase; Contact area in big toe, other toes, medial and lateral forefoot, midfoot and rearfoot regions during BE and FE.	↑ Ankle RoM in sagittal plane; ↑ Hip RoM in sagittal and transverse planes; ↓ Knee RoM in sagittal plane. ↑ ACR of ankle and hip in all planes;During BE, ↑ Hip angle in sagittal plane; ↑ Knee angle in transverse plane; ↓ Contact area in other toes; ↑ Contact area in midfoot and rearfoot;During FE, ↑ Hip angle in sagittal (–) and transverse (–) planes; ↓ Knee angle in transverse (–) plane. ↓ Contact area in big toe; ↑ Contact area in medial and lateral forefoot, midfoot and rearfoot.

Table 3. *Cont.*

Author (Year)	Outcome Measures	Key Findings of Higher–Level Compared to Lower–Level Players
Shao et al. (2020) [34]	Duration for backswing phase, forward-swing phase and whole cycle;HTA, FTA in all planes and XFA in sagittal plane at BE and FE; RoM and ACR of HTA, FTA in all planes and XFA in sagittal plane at backswing phase;PP at backswing and forward-swing phases and relative load during entire motion of hallux, other toes, medial, central and lateral forefoot, medial and lateral midfoot, medial and lateral rearfoot regions	↓ Backswing phase but ↑ forward swing phase and total duration; ↓ FTA in sagittal (–) and transverse planes at BE; ↑ XFA in sagittal plane at BE;↓ HTA in frontal plane at FE; ↑ FTA in sagittal and transverse (–) planes but ↓ in frontal plane at FE; ↓ XFA in sagittal plane at FE;↓ RoM of HTA and FTA but ↑ XFA in sagittal plane at backswing phase; ↓ RoM of HTA in sagittal and frontal but ↑ in transverse plane at forward–swing; ↑ RoM of XFA in transverse plane at forward–swing;↑ ACR in all joints and planes at backswing phase; ↑ ACR in all joints and planes at forward–swing phase except HTA in frontal plane;↑ PP of lateral forefoot and medial rearfoot but ↓ lateral forefoot, central forefoot, medial forefoot, other toes, hallux at backswing phase;↑ PP if lateral rearfoot, lateral forefoot, other toes but ↓ central forefoot, hallux at forward swing phase; ↑ relative load of other toes, lateral forefoot, medial rearfoot, lateral rearfoot but ↓ hallux, medial forefoot
Sheppard and Li (2007) [38]	Frequency of successful returns, ball speed, ball bounce location accuracy; Racket speed, position, direction of motion, orientation; and Variability of racket speed, acceleration, horizontal and vertical direction of motions, orientation; at the −200, −150, −100, −50, 0, +50 ms relative to the moment of ball contact	↑ successful returns, ball speed, ball bounce location accuracy; Significant interaction between playing level and time on the overall ball kinematics variables (MANOVA) ↑ racket speed, rightward direction, downward oriented; ↓ variability on racket horizontal direction of motion and orientation
Wang et al. (2018) [29]	Hip, knee and ankle joint angles and ACRs in all planes at beginning of backswing and end of swing phases. Standardized average, mean power frequency and median frequency for EMG of rectus femoris and tibialis anterior for both limbs.	↑ Rate of angular change for knee and hip in all planes; ↑ Rate of angular change for ankle in sagittal but ↓ in horizontal; ↑ MPF mean power frequency for all muscles; At beginning of backswing ↑ Ankle dorsiflexion; eversion; external rotation; ↑ Knee flexion; abduction; ↑ Hip flexion, adduction and external rotation;At end of swing ↑ Ankle dorsiflexion; knee flexion; ↓ Hip flexion, ↑ abduction.

Table 3. *Cont.*

Author (Year)	Outcome Measures	Key Findings of Higher–Level Compared to Lower–Level Players
Yu et al. (2019) [32]	Duration for backswing phase, forward-swing phase and whole cycle; HTA, FTA in all planes and XFA in sagittal plane at backswing and forward-swing phases; RoM and ACR of HTA, FTA in all planes and XFA in sagittal plane at backswing and forward-swing phases; PP and relative load of hallux, other toes, medial, central and lateral forefoot, medial and lateral midfoot, medial and lateral rearfoot at backswing and forward-swing phases.	↓ Backswing phase but ↑ forward swing phase and total duration; ↓RoM of HTA, HFA in all planes ↑RoM of XFA in sagittal plane ↑ Relative load for other toes, lateral forefoot; ↓ Relative load for medial forefoot and medial rearfoot. During backswing phase, ↑ HTA in sagittal and transverse (–); ↓ HTA in frontal; ↓ FTA in all planes (−) ↑ XFA in sagittal (–); ↑ ACR of HTA in sagittal and frontal;↑ ACR of RTA in frontal;↓ ACR of XFA in sagittal ↑ Lateral forefoot, medial and lateral rearfoot; ↓ PP for hallux, medial and central forefoot; During forward-swing phase, ↑ HTA in sagittal and transverse (–); ↑ HFA in frontal and transverse; ↑ XFA in sagittal (–) ↑ ACR of FTA in y direction; ↑ PP for other toes, central and lateral forefoot; ↓ PP for hallux.
Zhang et al. (2016) [31]	Accuracy; Duration and variability of duration for each phase (preparatory, backswing, forward–swing, follow through)	↑ Accuracy; ↓ Variability of duration for forward-swing and follow through phases;

ACR: angular changing rate; AP: anteroposterior; BE: backward-end; EMG: electromyography; FE: forward-end; FTA: right forefoot to hindfoot angle; HTA: right hindfoot to tibia angle; ML: mediolateral; PP: peak pressure; RoM: range of motion; XFA: right hallux to forefoot angle. (–) in negative direction/value. The increase/decrease of (–) refer to the absolute magnitude; ↑: significantly higher/larger/increase; ↓: significantly lower/smaller/decrease.

4. Discussion

There was evidence suggesting that higher-level table tennis players produced higher ball accuracy, performance index, and trial-to-trial repeatability in both training and competition. Meanwhile, it was generally perceived that ball and racket velocities were deterministic to playing level since high velocities make the opponent difficult to return the ball. In particular, the maximum racket speed at the moment of impact was regarded as the most important playing technique [1]. However, the current evidence did not come into a consensus that higher-level players necessarily produce higher ball or racket speed. Shoulder joint seems to play an important role to coordinate an effective stroke, as indicated by the effective use of elbow flexion torque, while the power of wrist joint is important during drop shot or long shot services. On the other hand, lower extremities facilitated momentum generation for increased racket velocity. In fact, leg–hip–trunk kinetics accounted for more than half of the energy and muscle force generation in racket sports [28]. Apart from a shorter period of swinging time, the increase in hip flexion and knee external rotations for higher-level players would potentially facilitate a more efficient muscle output to maximize racket velocity through the kinetic chain [28,29], in addition to larger hip and ankle angular velocities [28] which could be correlated with an increased ball speed after ball impact [50]. It should be noted that body coordination movement varies across individuals and trials but players attempted to reproduce movement during critical instants [33]. This was known as functional variability such that players could adapt to the conditions and requirements of the tasks and compensated for the changes with other movement parameters [51]. An optimal training model of body movement could be different among athletes.

Techniques in footwork could play an important role in compromising between dynamic stability and agility to recover back to the ready position for next moves or strokes. Less experienced players tended to have a larger peak ankle dorsiflexion and anterior center of pressure but lesser contact area, which indicated a poorer support base and stability [3,28]. Additionally, a shorter center of pressure in the anterior–posterior direction in higher level players facilitates quicker responses to resume to a neutral position for the next move [3,28]. However, it should be noted that higher level players exhibited larger ankle RoM during the match which may inherit the risk of ankle sprain [28,29].

Regarding the methodological quality, more than half of the included studies did not reveal clearly the source of population and sampling method. There was also a lack of blinding. Although blinding the maneuver conditions seemed to be impossible since the participants needed to be acknowledged for the tasks they performed, it could be accomplished by counting successive returns from delivering random serves by the coaches or serving robots [30,40]. Furthermore, the implementation of a randomized cross-over design across various interventions and maneuvers is necessary to avoid carry-over effects. Future studies can investigate how technologies can improve training outcomes. For instance, augmented reality (AR) technology with different filmed footages of different balls and gaze information can be modulated with artificial intelligence program to simulate the virtual opponent with the matched playing levels. Such simulations would provide a steppingstone towards individualized training solutions. On the other hand, several studies investigated a large number of outcome variables which was not well justified. While a full biomechanical profile with a large number of outcome variables were endeavored, statistical analyses were performed without corrections for multiple or multivariate comparisons. This may fall into the trap of data dredging or p-hacking [52] and those research may confine to exploratory studies [53].

There are some limitations when interpreting our findings. A systematic scoping review covered a vast volume of literature over a topic and thus offered an overview picture within the discipline [21]. However, due to the heterogeneity and breadth of the included studies, the established data framework did not attempt to answer a single research question which shall be put forward by a systematic review. It is also not possible to conduct meta-analysis to estimate overall determinants on playing levels, movement tasks, and equipment because of the diversity of objectives and designs across the included studies. In fact, the amount of literature required for a subset study was insufficient to formulate a focused research question for a traditional systematic review. For example, only two included studies were comparing upper limb kinematics of forehand topspin among different player levels in our review. In other words, it is pragmatically demanding to call for more research to establish the map over biomechanical variables, maneuvers, and playing levels, and reinforce key ideas on the determinants of performance using a unified study design and protocol.

Additionally, there was potential publication or language bias since some relevant articles were excluded for being published in Chinese, despite the fact that China is one of the dominating countries in the table tennis sports [16]. Summarizing information from the Chinese literature can enhance the impact of table tennis research but may require considerable effort in screening, translation, and interpretation. Furthermore, we found that there was a lack of literature on backspin maneuvers, longline maneuvers, strikes against sidespin ball, and pen-hold players that warrant further investigations.

Table 4. Key findings of included studies comparing different movement tasks.

Author (Year)	Outcome Measures	Key Findings
Bankosz and Winiarski (2017) # [1]	Time parameters (total time, duration of forward, hit-to-forward end, backward phases, time to reach max velocity (resultant and direction components) of racket; Distance parameters (resultant and direction components of distance travelled by racket during whole cycle, forward, hit-to-forward end, backward phases); Velocity parameters of racket (resultant and direction components of mean, max and at impact).	Forehand stroke ↓ total duration than against a spin serve and more power ball. Backhand stroke ↓ total duration than against a spin Strokes with more power produced ↑ velocity and distance parameters in AP direction; strokes against spin produced ↑ velocity and distance parameters in vertical direction; Forehand stroke (against spin and more power), produced ↑ velocity and distance parameters than backhand stroke.
Bankosz and Winiarski (2018) # [2]	Max racket velocity, racket velocity at ball impact, time to reach max racket velocity, time to reach racket velocity at ball impact; Angular velocity (max, min, at impact) for wrist, elbow, shoulder, pelvis, hip, knee and ankle; Multiple regression on racket velocity and angular velocity parameters of body segments.	For maximum-effort forehand topspin, racket velocity was correlated with hip flexion velocity on playing side, hip extension velocity on opposite site, and ankle flexion velocity on playing side; For maximum-effort backhand stroke, racket velocity was correlated with shoulder joint angular velocities on playing side, flexion velocity of ankle and adduction velocity of hip on opposite side.
Bankosz and Winiarski (2018) # [39]	Racket speed at impact; RoM of ankle, knee, hip, wrist, elbow and elbow joints in all planes during forward, hit-to-forward end, backward phases.	Diff forehand topspin types produced different RoM; ↑shot power accompanied by ↑rotation of upper body, pelvis and shoulders, flexion and rotation in shoulder, elbows and knees.
Bankosz and Winiarski (2020) # [33]	Lumbar, chest, hips, knees, shoulders, elbows and wrists angles and inter-individual coefficient of variation at ready, backswing, contact and forward instants; Above data for exemplary players and intra-individual coefficient of variation; Acceleration of hand at contact instant.	↑ intra-individual variability and high range of inter-individual variability; ↑ variability was observed in abduction/adduction of hip joints, wrist joints, thoracic and lumbar spines; Slightly ↑ variability when hit against a backspin compared to that against a topspin; ↑ variability at ready instants than other instants;
Ibrahm et al. (2020) [44]	Horizontal velocity of ball and racket head; Mean angular velocity of shoulder, elbow and wrist joints; Correlation between horizontal velocity of ball and racket head, and body segmental angular velocity at impact.	In forehand drop shot, Ball horizontal velocity correlated with racket head horizontal velocity positively; Wrist radial deviation velocity positively correlated with horizontal ball and racket head velocity; In long shot, Wrist radial deviation velocity and palmar flexion angular velocity positively correlated with horizontal racket head velocity.
Iino et al. (2008) [35]	Ball speed before and after impact; Magnitude, direction components of upper arm flexion, abduction, external rotation, elbow extension, forearm supination, wrist ulnar deviation and dorsiflexion at impact; Contributions to forward and upward racket velocities.	Against topspin, compared to against backspin: ↑ Upward component of elbow extension (–); ↓ Upward component of wrist dorsiflexion; ↑ Contribution to racket upward velocity by elbow extension (–), ↓ wrist dorsiflexion and racket tip linear.
Iino and Kojima (2016) [26]	Racket speed, face angle and path inclination at ball impact; Racket trajectory length; Ball impact location; Max pelvis axial rotation velocity; Upper trunk axial rotation velocity relative to pelvis, shoulder flexion velocity, external rotation velocity, elbow extension velocity, wrist dorsiflexion velocity at impact; Peak torque for shoulder, elbow and wrist; Shoulder, elbow and wrist angular velocities at instants of their matching peak joint torque.	No significant interaction between racket mass and ball frequency on all variables.Higher ball frequency, compared to lower ball frequency: ↓ Racket speed at impact; significantly more forward impact location; ↓ Max pelvis axial rotation velocity, upper trunk axial rotation velocity relative to pelvis at impact; Large racket mass, compared to small racket mass: ↑ Peak wrist dorsiflexion torque.

Table 4. *Cont.*

Author (Year)	Outcome Measures	Key Findings
Iino and Kojima (2016) [37]	Racket resultant, horizontal and vertical velocity at impact; Max shoulder joint center velocity in rightward and upward; Max angular velocity of upper trunk in extension and axial rotation; Peak joint torque for shoulder, elbow and wrist; Torque work by shoulder and elbow; Amount of energy transfer by joint torque and force components; Energy transfer ratio of racket arm.	Against backspin, compared to against topspin ↑ Resultant and vertical; but ↓ horizontal racket velocity; ↑ Max shoulder center velocity in upward direction; ↑ Max angular velocity of upper trunk in both extension and axial rotation; ↑ Peak shoulder flexion, external rotation torque and elbow valgus torque; ↑ Torque work by shoulder flexion/extension; but ↓ shoulder internal rotation and elbow extension;↑ Energy transfer through shoulder joint in rightward, upward, flexion/extension torque, abduction torque; ↑ Sum of energy transferred through shoulder; ↑ Mechanical energy of racket arm; ↑ Energy transfer ratio of racket arm;
Iino (2018) [36]	Correlation coefficients with horizontal (hV) and vertical velocities (vV) of racket at impact on: peak pelvis angular velocities in axial rotation and backward tilt; lateral flexion, axial rotation and backward tilt of playing side hip and forward tilt of non-playing side hip;Torque and force of both hips; Posterior tilt torques and vertical forces at both hips;Axial rotation torques at both hips; Total work done on pelvis.	Peak pelvis angular velocity in axial rotation direction was significantly correlated with hV and vV (–); Forward tilt of non-playing side hip was significantly correlated with hV and vV (–); Axial rotation torque of playing side hip was significantly correlated with hV; Axial rotation torque of non-playing side hip was significantly correlated with hV and vV(–); Posterior tilt torques and vertical forces at both hips was significantly correlated vV; Axial rotation torques at both hips was significantly correlated with hV.
Lam et al. (2019) [4]	Max vGRF and hGRF; Max knee flexion angle and moment; Max ankle inversion angle, angular velocity and moment; PP; Pressure time integral for plantar regions: total foot, toe, 1st MT, 2nd MT, 3rd–4th MT, 5th MT, medial and lateral midfoot and heel.	One-step, compared to both side-step and cross step: ↓ Max hGRF and vGRF; ↓ Max knee flexion and moment; ↓ Max ankle inversion, angular velocity and moment. and max ankle inversion angular velocity; ↓ PP for total foot, toe, 1st MT, 2nd MT, 5th MT. Side-step, compared to cross-step only: ↓ Max hGRF and vGRF; ↓ Max knee flexion and max ankle inversion angular velocity; ↓ PP for total foot and 1st MT. One-step, compared to cross-step only: ↓ PP for medial midfoot, medial heel and lateral heel
LeMansec et al. (2018) [46]	EMG muscle activity level of vastus lateralis, vastus medialis, rectus femoris, soleus, gastrocnemius lateralis, gastrocnemius medialis, biceps femoris, gluteus maximus Global level (average level) of EMG for all muscles	Comparing 5 maneuvers: Backhand top (BT), flick (FL), forehand spin (FS), forehand top (FT), smash (SM): Global level of EMG BT ↑ all others; FL ↑ FS, FT, SM; FS↑ SM; For EMG of vastus lateralis and vastus medialis FS ↑ B, FL, SM; FT ↑ SM For rectus femorisFS and FT ↑ BT, FL, SM; For soleus and gastrocnemius lateralis BT ↓ FL, FT; SM ↑ all others For gastrocnemius medialis SM ↑ all others; FL ↑ BT, FS; FT ↑ BT, FS; For gluteus maximusFS, FT, SM ↑ BT, FL For biceps femoris FS, FT, SM ↑ BT, FL; FL ↑BT

Table 4. *Cont.*

Author (Year)	Outcome Measures	Key Findings
Malagoli Lanzoni et al. (2018) [45]	Angle of racket in all planes; Average feet-table angle; Max, min angle and moment of max velocity of racket (MMV) for angulation of: shoulders-table, shoulder-racket, pelvis-table, elbow and left/right knees	Cross-court, compared to long-line was: ↓ Racket angle in axial direction (z); ↓ Average feet-table angle; ↓ Max and min shoulder-table; ↓ Max but ↑ MMV of shoulder-racket angles; ↓ Max, min and MMV of pelvis-table angles; ↑ Elbow MMV; ↓ Right knee MMV
Meghdadi et al. (2019) [47]	Muscle activity; muscle onset; offset time instant for: supraspinatus, upper trapezius, lower trapezius, serratus anterior, biceps brachii, anterior deltoid	Shoulder impingement syndrome group, compared to healthy group ↓ muscle activity of supraspinatus and serratus anterior; ↑ muscle activity of upper trapezius; Significantly later muscle onset time for serratus anterior but significantly earlier muscle onset time for upper trapezius
Yan et al. (2017) [27]	Buffer time; CoM in AP and ML directions; Right knee angle at peak GRF	180° step compared to 45° step ↑ CoM in AP direction (A or P direction not specify); Higher sole-ground friction ↓ right knee angle at peak GRF.
Yu et al. (2018) [30]	Duration from initiation to backward-end, from backward-end (BE) to forward-end (FE), from forward-end to initial ready position (RP)Hip, knee and ankle angle at RP, BE and FE in three planes. Force-time integral in big toe, other toes, medial forefoot, lateral forefoot, midfoot and rearfoot	Squat serve, compared to standing serve: In sagittal plane ↑ hip angle at RP, BE and FE; ↑ knee angle at BE and FE; ↓ ankle angle at RP but ↑ at BE and FE; In frontal plane ↑ hip angle (–) at BE and FE; ↓ knee angle at BE and FE;In transverse plane ↑ hip angle at FE; ↑ knee angle at BE and FE; ↓ force-time integral in rearfoot
Yu et al. (2019) [48]	RoM of hip, knee and ankle joint in three planes. Hip, knee and ankle joint in three planes at take-off (T1) and backward-end (BE) instants. ACR of hip, knee ankle in three planes.	Long chasse steps, as compared to short chasse steps: ↑ RoM of hip in sagittal and transverse planes; ↑ RoM of knee in coronal plane; ↑ RoM of ankle in coronal and transverse planes; ↑ ACR of hip in sagittal plane; ↓ ACR of knee but ↑ that of ankle joint in coronal plane; ↑ ACR of hip and ankle in transverse plane; During T1, long chasse steps, compared to short chasse steps: ↓ hip angle in sagittal and transverse planes; ↓ knee angle in transverse plane; ↑ ankle angle in sagittal plane but ↓ in coronal and transverse planes (–); During BE, long chasse steps, compared to short chasse steps: ↓ hip angle in sagittal plane; ↑ knee angle in coronal plane but ↓ in transverse; ↑ ankle angle in sagittal plane
Zhou et al. (2014) [42]	Racket speed at ball contact, during backswing and follow through; percentage duration of backswing, attack and follow through phases	Curving ball, compared to fast break: ↑ racket speed at ball contact

ACR: angular changing rate; AP: anterior-posterior; BE: backward-end; CoM: center of mass; EMG: electromyography; FE: forward-end; hGRF: horizontal ground reaction force; hV: horizontal velocity; MMV: maximum velocity of the racket; MT: metatarsal; PP: peak pressure; RoM: Range of Motion; RP: ready position; T1: take-off; vGRF: vertical ground reaction force; vV: vertical velocity. (–) in negative direction/value. ↑: significantly higher/larger/increase; ↓: significantly lower/smaller/decrease of the absolute magnitude. # Only highlighted key findings were summarized in the table since these studies included too many outcome variables and/or pairwise comparison results to be listed.

5. Conclusions

A systematic scoping review of published studies specific to the biomechanics of table tennis maneuvers was conducted to categorize biomechanical variables within the domain of playing levels and maneuvers. This review could serve as the first scoping review to provide a clear overview about table tennis research in the past decades. Recent research on table tennis maneuvers targeted the differences between playing levels and between maneuvers using parameters which included ball and racket speed, joint kinematics and kinetics, electromyography, and plantar pressure distribution.

Different maneuvers underlined changes on body posture and lines of movement which were accommodated particularly by racket face angle, trunk rotation, knee and elbow joints, and different contributions of muscles. Key findings regarding determinants of playing levels were summarized to offer practical implications as follows:

- Higher-level players produced ball striking at higher accuracy and repeatability but not necessarily of higher speed.
- Strengthening shoulder and wrist muscles could enhance the speed of strike.
- Whole-body coordination and footwork were important to compromise between agility and stability for strike quality.
- Personalized training shall be considered since motor coordination and adaptation vary among individuals.

Moreover, this scoping review found that while most investigations focused on the upper and lower limb biomechanics of table tennis players performing different maneuvers, fewer studies looked into trunk kinematics and EMG. Furthermore, our study identified research gaps in backspin maneuvers and longline maneuvers, strikes against sidespin, and pen-hold players that warrant future investigations.

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Effects of Backpack Loads on Leg Muscle Activation during Slope Walking

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Abstract: Hikers and soldiers usually walk up and down slopes with a load carriage, causing injuries of the musculoskeletal system, especially during a prolonged load journey. The slope walking has been reported to lead to higher leg extensor muscle activities and joint moments. However, most of the studies investigated muscle activities or joint moments during slope walking without load carriage or only investigated the joint moment changes and muscle activities with load carriages during level walking. Whether the muscle activation such as the signal amplitude is influenced by the mixed factor of loads and grades and whether the influence of the degrees of loads and grades on different muscles are equal have not yet been investigated. To explore the effects of backpack loads on leg muscle activation during slope walking, ten young male participants walked at 1.11 m/s on a treadmill with different backpack loads (load masses: 0, 10, 20, and 30 kg) during slope walking (grade: 0, 3, 5, and 10°). Leg muscles, including the gluteus maximus (GM), rectus femoris (RF), hamstrings (HA), anterior tibialis (AT), and medial gastrocnemius (GA), were recorded during walking. The hip, knee, and ankle extensor muscle activations increased during the slope walking, and the hip muscles increased most among hip, knee, and ankle muscles (GM and HA increased by 46% to 207% and 110% to 226%, respectively, during walking steeper than 10° across all load masses (GM: $p = 1.32 \times 10^{-8}$ and HA: $p = 2.33 \times 10^{-16}$)). Muscle activation increased pronouncedly with loads, and the knee extensor muscles increased greater than the hip and ankle muscles (RF increased by 104% to 172% with a load mass greater than 30 kg across all grades (RF: $p = 8.86 \times 10^{-7}$)). The results in our study imply that the hip and knee muscles play an important role during slope walking with loads. The hip and knee extension movements during slope walking should be considerably assisted to lower the muscle activations, which will be useful for designing assistant devices, such as exoskeleton robots, to enhance hikers' and soldiers' walking abilities.

Keywords: muscle activations; electromyography; slope walking; backpack loads

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Highlights

- Hip extensor muscle activations increase most during slope walking;
- Muscles increased pronouncedly during slope walking with backpack loads;
- Knee extensor muscle activations increased most with increasing backpack loads.

1. Introduction

Hiking is a popular exercise providing benefits, including the acceleration of calorie consumption and the burning of fat tissue [1,2]. Hiking on slopes and with backpack loads can cause pain and injuries of the musculoskeletal system, especially during a prolonged load journey [3–5].

Walking with load carriage leads to high energetic consumptions and joint moments. Many studies have explored the energetics and kinematics with different backpack loads during level walking [6–10]. Karen et al. [6] analyzed the energy cost of walking with and without a backpack load and pointed out that the load increased the oxygen uptake at a constant rate. Raymond et al. [7] further analyzed the effects of load masses added to the legs on energetics and biomechanics. They summarized that the metabolic rate increased with the load mass and the kinematics and muscle moments increased rapidly with loads at the feet. Additionally, the increased metabolic rate with the load carriage may be caused by increased ankle positive work during push-off [8]. Morrison et al. [9] also analyzed the motion entropy changes to the load carriage at a joint level and pointed out that the entropy of spine slide flexion increased while hip flexion entropy decreased. Kari et al. [11] pointed out the sex effect on the kinematics with loads: females used more hip and knee moments with loads compared to males during walking. Furthermore, Krajewski [12] studied the effect of load carriage magnitudes and different locomotion patterns (fast run and force marching) on knee moments.

Most studies investigating load carriages suggest a positive correlation exists between load mass and joint moments, as well as energetics. However, joint moments or joint work in one gait calculated by joint moments and angular displacement do not account for muscle activations during load carriage walking [13]. Researchers have made explorations into the muscle activation patterns during level walking with load carriages [4,5,14]. Karina et al. [4] found that the muscle activity changed differently with increased load masses to adjust to maintain balance and attenuate the loads placed on lower limbs. The muscle activations of the soleus, medial gastrocnemius (GA), lateral hamstrings (HA), and rectus femoris (RF) increased with load, and the muscle activation patterns were similar between men and women [5]. Kenneth et al. [14] studied the musculoskeletal stiffness during load carriages at different walking speeds and found the musculoskeletal stiffness increased as a function of both speed and load. Walsh et al. [15] investigated the effect of stable and unstable load carriages on muscles of older adults. They pointed out that unstable load carriages increased the activity of the RF and soleus, while stable load carriages increased the RF activity.

Subjects such as soldiers and hikers who habitually walk uphill with load carriages usually feel tired because of muscle fatigue. It is important to design assistive devices for soldiers and hikers according to the load carriage effects. A backpack load, for example, increases the dynamic forces on the human body. Huang et al. [16] and Yang et al. [17] designed suspended-load backpacks as an assistive device to reduce the dynamic forces.

Slope walking has been reported to lead to greater leg extensor muscle activities and joint moments. However, most of the studies investigated muscle activities [18–20] or joint moments [21–24] during slope walking without load carriages or only investigated the joint moment changes [9,10,12] and muscle activities [15,16,25] with load carriages during level walking. Whether the muscle activations such as the signal amplitude are influenced by the mixed factor of load mass and grade and whether the influence of the degrees of load mass and grade on different muscles are equal have not yet been investigated. It is important to know how the muscles are activated during one gait when slope walking with backpack loads for designing assistive devices, such as exoskeleton robots, to enhance people's movement abilities with backpack loads.

This study aimed at investigating the effects of load carriages on muscle activities during slope walking to provide suggestions for the design of assistive devices. We hypothesized that (1) hip, knee, and ankle extensor muscle signal amplitudes during one gait would increase during the slope walking compared to level walking, especially the hip extensor muscles, such as the gluteus maximus (GM) and HA; (2) the muscle signal amplitudes would increase pronouncedly with loads during slope walking; (3) the muscle signal amplitudes would increase at different degrees. The knee extensor muscles, such as the RF, would be activated more compared to the hip and ankle extensor muscles, such as GM, HA, and GA.

2. Methods

2.1. Subjects

Ten young male adults volunteered for this study (mean ± standard deviation; age: 24.10 ± 1.79 years, height: 175.30 ± 5.12 cm, and mass: 69.40 ± 8.15 kg). All subjects were familiar with treadmill walking and had no neuromuscular, cardiovascular, or orthopedic diseases. All subjects were provided with the informed consent form for the experiment, and the experiment was approved by the Ethics Committee of Beijing Sport University (No. 2019007H (2019.01–2021.01)).

2.2. Experimental Protocol

Subjects walked on a force treadmill (Bertec, Columbus, OH, USA), on which the walking grade and walking speed could be adjusted accordingly. Each subject was familiarized with walking on the treadmill for about 5 min with the walking speed set to 1.11 m/s and the walking grade set to 0°. The speed of 1.11 m/s (4 km/h) was selected according to the American College of Sports Medicine [22] and the literature on uphill and downhill walking [23]. Each subject was required to walk for 2 min on the treadmill with three mass backpack loads (10, 20, and 30 kg) during four grades (level (0°), 3, 5, and 10° grades). The level and slope walking on the treadmill without loads (0 kg) were also completed by each subject as a reference. Subjects were asked to perform all the load mass tests at one certain grade in order to shorten the experiment time. The load masses were in random order during one grade test, and the grade testes were also in random order. Subjects were asked to walk at one grade for 3 min with one load during one trial. Subjects rested for 3 min between different load mass trials and 30 min between different grade tests. The experimenters adjusted the load mass and the walking grade while the participants rested. The speed of the treadmill was set to 1.11 m/s in every trial. The temporal stride kinematics and surface electromyographic (EMG) during the final 30 s of each trial were recorded.

The temporal stride kinematics were calculated according to the ground force data recorded by the force platform in the treadmill. Two force platforms were embedded in the treadmill and recorded the ground interactive forces and moments between foot and ground. The software in the optical motion-capture system (Motion, Columbus, OH, USA) [26] calculated the center of force operation based on the forces and moments.

The EMG signals were collected by the wireless EMG system (Delsys Trigno^{TM} Wireless EMG System (Natick, MA, USA)). The muscles selected included the GM, RF, HA, anterior tibialis (AT), and GA, which are the main activated muscles during human lower extremity movements [27]. The pre-amplified single differential electrodes (Trigno, Delsys, Natick, MA, USA) were placed on the muscle bellies after preparing the skin with alcohol (the schematic diagram of human lower limb muscles in this study is shown in Figure 1). The surface EMG (sEMG) signal sensors were fixed with double-sided tapes and bandages to prevent displacement between the sensors and muscles during walking.

This optical motion capture system (Motion, Columbus, OH, USA) can integrate the Delsys device and the Bertec force platforms by a data acquisition card (DAQ, National Instruments, Austin, TX, USA); thus, the sEMG signal and kinetic data can be collected simultaneously. The sampling frequency was selected as 1200 Hz.

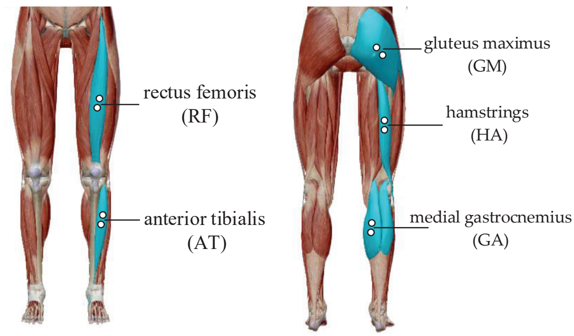

Figure 1. The schematic diagram of the human lower limb muscles in this study. The position of the two white points on one muscle represented the position of the electromyographic (EMG) electrodes attached.

2.3. Data Analysis

2.3.1. The Temporal Stride Parameters

The kinetic data were obtained by the output of the software provided by the optical motion-capture system (Cortex-64, Santa Rosa, CA, USA), which was linked with the force treadmill system (Bertec, Columbus, OH, USA). Raw kinetic data were smoothed using a 4th-order Butterworth filter with a cutoff frequency of 10 Hz [28]. We then used vertical ground reaction force data and a threshold of 20 N (on the basis of the standard deviation of the vertical ground reaction force signal during leg swing) [29] to determine the heel strike and toe-off for each leg and computed the temporal characteristics of each trial using custom software (MathWorks Inc., Natick, MA, USA). The step length was determined as the distance between the point where the left heel strike occurred and the point where the right heel strike occurred in the walking direction, and the step width was determined as the distance between the two heel strike points in the walking transverse distance [30].

2.3.2. The EMG Analysis

The raw EMG signals were filtered by a bidirectional Butterworth band-pass filter with cutoff values of 20 and 500 Hz [31,32] in a custom script written in MATLAB (MathWorks Inc., Natick, MA, USA). The signals were full-wave rectified and filtered by a low-pass filter at 10 Hz [33,34]. The filtered EMG signals were then used to calculate the root mean square (RMS EMG) with a window of 10 ms to describe the muscle activation during movement [33]. In addition, the filtered EMG signals were used to calculate the mean of the EMG (MEMG) to describe the work of one muscle during movement. Since the gait cycle was usually normalized to 0~100%, the sEMG data acquired synchronously should be also interpolated into 0~100% to describe how the muscles were activated during the gait [34,35]. Therefore, all the RMS EMGs were interpolated into 101 points corresponding to the gait cycle.

2.4. Statistical Analysis

We calculated the mean values of step length, step width, and RMS EMG, as well as MEMG, over ten consecutive strides. We then normalized the temporal stride parameters and determined the EMG parameters by calculating the ratio of the values during slope walking with loads to the values during level walking without loads. We used a two-factor (grade × load mass) analysis of variance for repeated measures to test the significant effects of the grade and load masses. When there was a significant effect ($p < 0.05$), Bonferroni corrected post hoc comparisons (adjusted $p < 0.0072$ (0.05/7, two dependent stride variables and five sEMG variables) [18,22–24,36] were carried out to evaluate the differences between the grades and load masses. All statistical analyses were conducted using SPSS software (IBM SPSS Statistics 22).

3. Results

3.1. Temporal Stride Kinematics

As the load mass and grade increased, subjects took a longer step length than level walking (Table 1). The changes in the normalized step widths were different between the load mass and grade groups. Both of the temporal stride kinematics were not significantly different from level walking without backpack loads (all $p > 0.05$).

Table 1. Normalized step lengths and step widths during different grades of walking with different load masses.

	Normalized Step Length				Normalized Step Width			
	0°	**3°**	**5°**	**10°**	**0°**	**3°**	**5°**	**10°**
0 kg	1.00	1.07 (0.06)	1.03 (0.06)	1.08 (0.07)	1.00	1.03 (0.05)	1.04 (0.07)	1.02 (0.05)
10 kg	1.06 (0.06)	1.10 (0.09)	1.06 (0.10)	1.05 (0.09)	0.98 (0.08)	0.98 (0.06)	1.04 (0.06)	1.02 (0.06)
20 kg	1.09 (0.05)	1.13 (0.09)	1.10 (0.08)	1.10 (0.08)	1.04 (0.06)	0.98 (0.09)	0.95 (0.06)	1.07 (0.01)
30 kg	1.10 (0.08)	1.08 (0.06)	1.06 (0.08)	1.02 (0.06)	1.09 (0.09)	1.00 (0.08)	0.97 (0.07)	1.03 (0.05)

All the step lengths and step widths were divided by the values during level walking without backpack loads. Therefore, the normalized step lengths and step widths were "1.00", and all the normalized step lengths and step widths were dimensionless in Table 1.

3.2. Muscle Activities

As expected, the mean muscle activities of the hip, knee, and ankle extensors generally increased with the increase of the load mass and grade (Figures 2–5 and Table S1). Both the grade and load mass had a significant effect on all muscles ($p < 0.05$), and none of the muscles showed significant grade-load-mass interactions ($p > 0.05$).

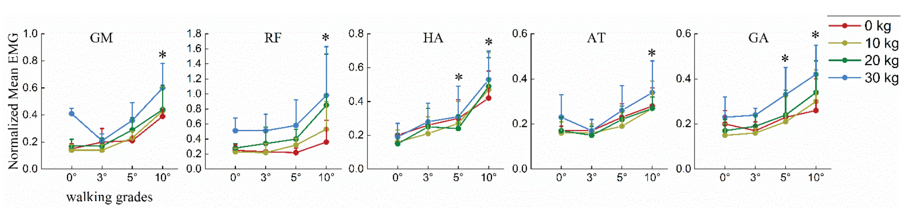

Figure 2. The mean EMG signals for muscles during different slope walking across all load masses in one gait from right heel strike to the next right heel strike, normalized to the mean activity during level walking without backpack loads. * Representing the mean EMG was significantly different from level walking across all the backpack loads, according to post hoc comparisons with a Bonferroni adjusted level of significance ($p < 0.0072$). The red line represented the normalized mean EMG of each muscle at different grades without backpack loads. The yellow one represented the normalized mean EMG of each muscle at different grades with 10-kg backpack loads. The green one represented the normalized mean EMG of each muscle at different grades with 20-kg backpack loads. The blue one represented the normalized mean EMG of each muscle at different grades with 30-kg backpack loads.

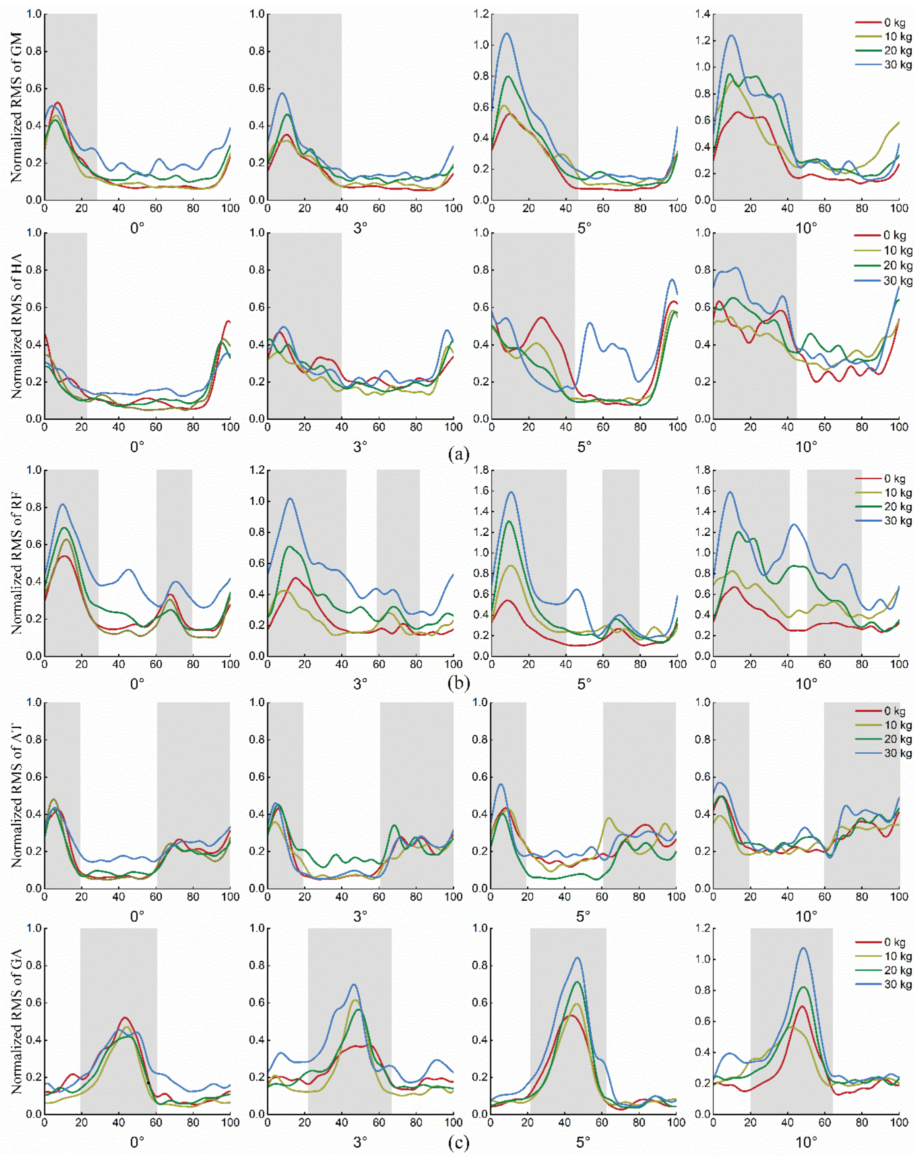

Figure 3. The muscle activity of leg muscles in one gait during different slope walking across all backpack loads, normalized to the mean activity during level walking without backpack loads: (**a**) The muscle activity of the hip extensor muscles. (**b**) The muscle activity of the knee extensor muscles. (**c**) The muscle activity of the ankle muscles. The different colors of the curves had the same representation as those colors in Figure 2. The gray area represented the higher muscle activation duration in one gait. RMS: root mean square.

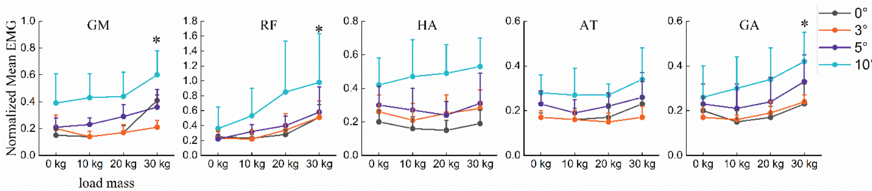

Figure 4. Mean EMG signals for muscles while walking across all grades with different backpack loads during one gait from right heel strike to the next right heel strike, normalized to the mean activity during level walking without backpack loads. * Representing the mean EMG was significantly different from walking without backpack loads across all the grades, according to post hoc comparisons with a Bonferroni adjusted level of significance ($p < 0.0072$). The line in dark gray represented the normalized mean EMG of each muscle with different load masses during level walking. The line in orange represented the normalized mean EMG of each muscle with different load masses during slope walking at grade 3°. The line in purple represented the normalized mean EMG of each muscle with different load masses during slope walking at grade 5°. The line in blue represented the normalized mean EMG of each muscle with different load masses during slope walking at grade 10°.

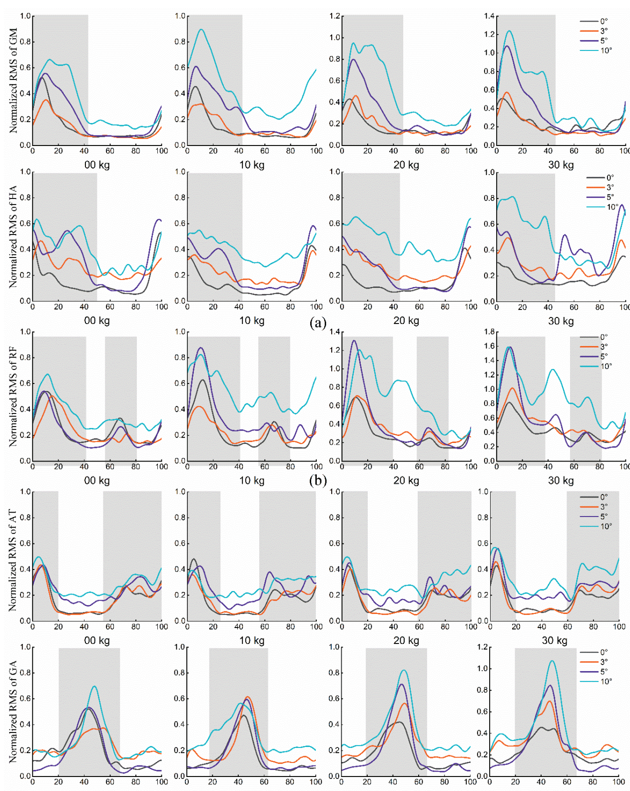

Figure 5. The muscle activity of leg muscles in one gait during walking with different backpack load masses across all grades, normalized to the mean activity during level walking without backpack loads: (**a**) The muscle activity of the hip extensor muscles. (**b**) The muscle activity of the knee extensor muscle. (**c**) The muscle activity of the ankle muscles. The different colors of the curves had the same representation as those colors in Figure 4. The gray area represented the higher muscle activation duration in one gait.

3.2.1. Grade Effects

The mean EMG signals of the hip, knee, and ankle extensor muscles increased generally during most of the slope walking, especially during the 10° slope walking (GM, $p = 1.32 \times 10^{-8}$; HA, $p = 2.33 \times 10^{-16}$; RF, $p = 0.2 \times 10^{-5}$; and GA, $p = 6.74 \times 10^{-9}$). Compared to the level walking, these increases were statistically significant for GM, RF, and AT at the 10° grade and for HA and GA at the 5° and 10° grades (Figure 2). Compared to the level walking with the same loads, the mean EMG of GM increased greatly during walking at the 10° grade by 46% to 207%, HA increased by 110% to 226%, RF increased by 44% to 203%, AT increased by 48% to 68%, and GA increased by 30% to 100% (shown in Table S2).

As the grade increased, the muscle activity of the hip extensor muscles (GM and HA, especially GM) increased both in the activation value and duration (Figure 3a). The maximum activation value of the GM changed from 0.52 to 1.24, and the activation duration increased from about 28% to 48% of one gait as the grade increased. The maximum activation value of the HA changed from 0.52 to 0.81, and the activation duration changed from 20% to about 45% of one gait as the grade increased.

The muscle activity of the knee extensor muscle (RF) increased most in the activation value, shown in Figure 3b. The maximum activation value of the RF increased from 0.82 to 1.60 during the early stance stage and from 0.4 to about 0.9 during the early swing stage.

As expected, the muscle activity of the ankle extensor (GA) increased considerably in the muscle activation value (Figure 3c). The maximum activation of the GA during the median and the late stances increased highly, from 0.52 to 1.07. Compared to the ankle extensor, the muscle activity of the ankle dorsiflexion (AT) increased slightly, from 0.48 to 0.57 during the early stance stage and swing stage.

3.2.2. Load Mass Effects

The mean EMG signals of the hip, knee, and ankle extensor muscles increased generally as the backpack load masses increased. Compared to walking without backpack loads, the increases were statistically significant for the GM and GA during walking with 30-kg loads (GM, $p = 0.000091$ and GA, $p = 0.00091$) and for the RF during walking with 30 kg ($p = 8.86 \times 10^{-7}$) (Figure 4). The increases in the HA and AT were not statistically significant ($p > 0.0072$). Compared to walking without loads at the same grade, the mean EMG of the GM increased by 5% to 173% with a 30-kg backpack load, the HA increased by −5% to 26%, the RF increased by 104% to 172%, the AT increased by 0% to 35%, and the GA increased by 15% to 61% (the data are shown in Table S3).

The muscle activity of the hip extensors (GM and HA) increased as the backpack load mass increased, as shown in Figure 5a. The maximum muscle activity of the GM increased greatly from 0.66 to 1.24 as the backpack load mass increased from 0 to 30 kg. The maximum muscle activity of the HA increased slightly from 0.63 to 0.81 as the load mass increased.

The knee extensor muscle RF increased greatly in muscle activation as the backpack load mass increased (Figure 5b). The maximum muscle activation of the RF increased greatly from 0.67 to 1.60 during the early stance stage as the load mass increased from 0 to 30 kg across all the slope walking.

The ankle extensor muscle GA increased greatly in muscle activation from 0.70 to 1.07 as the backpack load mass increased from 0 to 30 kg, and the dorsiflexion muscle AT increased slightly, from 0.50 to 0.58 (Figure 5c). The muscle activity demonstrated that the ankle extensor muscles were activated more than the flexor muscles during slope walking with backpack loads.

4. Discussion

This study quantified the hip, knee, and ankle muscle activations during level and slope walking with different backpack loads. Compared to level walking, the hip, knee, and ankle muscle activations increased generally during slope walking, especially the hip extensor muscle activations. The increased mean EMG of the leg muscles in this study were consistent with published findings [4,5,18–20]. Moreover, the increase became more pronounced with backpack loads, especially the hip and knee muscle activations. The hip extensor muscles increased the most with grades changing, and the knee extensor muscles increased the most with loads changing, which expanded our knowledge of muscle activation strategies during slope walking with backpack loads.

The results of this study supported our first hypothesis that the hip, knee, and ankle extensor muscle activations would increase during the slope walking, especially the hip extensor muscle activations, compared to level walking. In this study, all the leg extensor muscle activations increased during slope walking to raise the body's center of mass, which were consistent with prior studies [18–20]. In addition, the hip extensor muscle activations (GM and HA) increased remarkably more (the GM increased by 46% to 207% and the HA increased by 110% to 226%) than the ankle extensor muscle (GA increased by 30% to 100%) at steeper grades. The activation value and duration of the hip extensor muscles increased remarkably at the early stance stage during slope walking (shown in Figure 2). This demonstrates the pronounced role of hip extensor muscles during slope walking [18,23], which was also described by the greatest increase of the hip extension moment (increased from 1.01 to 1.37 when the treadmill gradient increased from 0% to 20% [22]) and the greatest increased power of the hip extensor muscles (increased by 85% at push-off and by 75% during mid-stance while walking on uneven terrain [21]). The hip extensors provided greater acceleration of the COM and generated more power for the trunk and ipsilateral leg during slope walking [19], which may also be the reason that the hip extensor muscles were more pronouncedly activated on slopes in our study.

The results of this study also supported our second hypothesis that muscle activations would increase pronouncedly with backloads during slope walking. The increases of the GM, RF, and GA became significant statistically when slope walking with a big backpack load (30 kg). Consistent with previous investigations [4,37,38], the mean amplitude of the RF and GA increased with loads in this study. The RF activation increased to provide more force and energy to extend the knee to attenuate the impact forces with heavy load carriage [38] and to maintain lower limb stability as the load mass increased [4]. The increase of the GA activations provided more power for walking by increasing the plantar flexing [38], which was thought to overcome the inertia associated with increasing backpack loads [39]. However, the mean EMG of the AT increased in this study, while the average amplitude of the AT remained unaffected [37,38]. This difference may be caused by different experimental designs in our study and theirs. The walking conditions in our study were slope walking with backpack loads, while their studies' conditions involved level walking. People elicited larger AT activity during slope walking to provide greater ankle dorsiflexion than level walking [19,22,28]. The slope grades may enlarge the influence of backpack loads on muscles.

The results of this study supported our third hypothesis that the muscle activations would increase at different degrees, and the knee extensor muscles would be activated more compared to the hip and ankle extensor muscles. Compared to walking without loads at the same grade across all slope walking, the mean EMG of the knee extensor muscle (RF) increased significantly by 104% to 172% with 30-kg backpack loads. The increase of the knee extensor muscle was much greater than that of the ankle extensor muscle (GA increased by 15% to 61% with 30-kg backpack loads relative to without loads across all grades). The knee extensor muscle increased most to provide greater force for body support during the early stance stage, which was consistent with other investigations [18,38,40]. Except for the knee extensor muscle activations, the hip extensor muscle GM activations also increased more pronouncedly than the ankle extensor muscle GA (GM increased by 5% to 173%). With the loads increasing, the energy and power for walking increased greatly [7,10]. The hip extensor muscle GM played an important role in the acceleration of the trunk [19]. Thus, the GM activations increased pronouncedly to provide more power for the acceleration of the trunk as the backpack loads increased. The results implied that the leg extensor muscles may have different contributions during walking with backpack loads. The knee extensor and hip extensor muscles may play a greater role during walking with heavy loads, which was also speculated by Harman [38].

In our present study, the EMG of the GM, HA, RF, AT, and GA were analyzed to investigate the muscle strategy during slope walking with backpack loads. However, one limitation of our study is that we did not acquire the kinematic data in the experiment that may give force to our work.

In addition, another limitation of our study is that the muscles analyzed were relatively few and most of them were focused on the leg extensor muscles. The muscles around the trunk, such as the external oblique muscles, were not analyzed in this study, which influenced pronouncedly during the inclined walking with backpack loads. The vastus medialis and vastus lateralis at the knee joint were influenced a lot during slope walking to provide more forces for lower limb stability [18,19], which were also not analyzed in this study. Thus, we should acquire the kinematics data and analyze more muscles by experiment or simulation [19,41] in the future to expand the insights into the muscle strategy during slope walking with backpack loads. Finally, considering the load intensity, only male participants were recruited in this study. Males and females may have different muscle-activation strategies during slope walking with backpack loads. Future studies may be needed to understand how muscle activations are influenced by the grade and loads using female subjects.

5. Conclusions

In this study, we explored the effects of backpack loads on leg muscle activations during slope walking. It was concluded that the hip, knee, and ankle extensor muscle activations increased during slope walking, and the hip muscle increased the most among the hip, knee, and ankle muscles. Moreover, muscle activations increased pronouncedly with loads during slope walking, and the knee extensor muscle activations increased more than the hip and ankle muscles. The results in our study imply that the hip and knee muscles play an important role during slope walking with loads. Our results are important for the design of assistant devices, such as exoskeleton robots, to enhance people's walking ability, especially for hikers and soldiers. The hip and knee extension movements during slope walking should be considerably assisted to lower the muscle activations. Future studies could explore the effects of loads and grades on more muscles and involve more participants, including female subjects, to expand the insights into the muscle strategy for providing more suggestions for the design of assistant devices.

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