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Single-Leg Landings Following a Volleyball Spike May Increase the Risk of Anterior Cruciate Ligament Injury More Than Landing on Both-Legs

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Abstract: Volleyball players often land on a single leg following a spike shot due to a shift in the center of gravity and loss of balance. Landing on a single leg following a spike may increase the probability of non-contact anterior cruciate ligament (ACL) injuries. The purpose of this study was to compare and analyze the kinematics and kinetics differences during the landing phase of volleyball players using a single leg (SL) and double-leg landing (DL) following a spike shot. The data for vertical ground reaction forces (VGRF) and sagittal plane were collected. SPM analysis revealed that SL depicted a smaller knee flexion angle (about 13.8°) and hip flexion angle (about 10.8°) during the whole landing phase, a greater knee and hip power during the 16.83–20.45% (p = 0.006) and 13.01–16.26% (p = 0.008) landing phase, a greater ankle plantarflexion angle and moment during the 0–41.07% ($p < 0.001$) and 2.76–79.45% ($p < 0.001$) landing phase, a greater VGRF during the 5.87–8.25% (p = 0.029), 19.75–24.14% (p = 0.003) landing phase when compared to DL. Most of these differences fall within the time range of ACL injury (30–50 milliseconds after landing). To reduce non-contact ACL injuries, a landing strategy of consciously increasing the hip and knee flexion, and plantarflexion of the ankle should be considered by volleyball players.

Keywords: volleyball spike landing; sagittal biomechanics; non-contact ACL injuries; statistical parametric mapping (SPM)

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1. Introduction

As a competitive sport, volleyball has a high rate of musculoskeletal injury [1]. The special movements of volleyball, such as jumping, landing, blocking, and spiking, need to be combined with fast movements, that put very high demands on the musculoskeletal system of volleyball players. This can cause considerable damage to the musculoskeletal system of players [2,3]. High-level volleyball players are prone to suffering lower limb injury following overuse caused by continuous blocking and attacking [3–5]. This is especially true for the knee joint which is easily injured in the energy transmission process because it comprises of a joint capsule with multiple joints. Previous studies have shown that 63 percent of musculoskeletal injuries in volleyball occur during the basic jump and landing tasks of blocking and spiking. Most of the injuries are located at the knee joint and most injuries are anterior cruciate ligament (ACL) injuries [5]. Anterior cruciate ligament (ACL) injury is a common injury in sports, and 70–90% of ACL injuries are caused by non-contacts [6–8].

From a mechanical point of view, ACL injury is caused by excessive ACL tension. The non-contact injury of ACL is generally caused by the large force or moment generated by the player himself on the knee joint, so that the force of ACL exceeds its ultimate load capacity. The kinematics and kinetics of the sagittal plane are critical to ACL injury during landing tasks, with the increase of the anterior tibial shear force (mainly from the sagittal plane), the load on the ACL (the risk of ACL injury) will increase [9–11]. Previous studies have also shown that the sagittal hip and knee joints play a major role in energy dissipation and buffering during landing, and smaller knee and hip flexion will increase the impact pressure on the knee (the anterior tibial shear force), leading to an increased risk of ACL injury [12–14]. At the same time, the knee and hip flexion angle during the landing phase were associated with ground reaction forces during jump-landing tasks. Increasing the knee and hip flexion angle might reduce impact force during landing tasks [15]. Ryan et al. found that the joint angle and joint moment were strongly correlated with the ACL strain force. When the knee and hip flexion angles decrease and the knee and hip extension moment goes up, the ACL strain force is significantly increased [9,16]. In the sagittal plane, the lower limb joint has a wide range of motion (ROM) and can provide enough shock absorption without excessive compression of the fragile soft tissue structures such as ligaments [17]. For the total energy dissipation of the lower limb joint during a jump-landing task, previous studies have shown that the sagittal plane is about 10 (landing with double-leg) and 20 (landing with single-leg) times larger than the frontal plane [17]. Therefore, the sagittal plane plays a major role in reducing lower limb injuries.

Different landing modes will have a greater influence on the load of the lower limb joints and ligaments [18]. Unreasonable landing patterns include smaller knee and hip flexion angles in the sagittal plane, and greater vertical ground reaction forces (VGRF), which significantly increase the risk of ACL and lower limb injuries [12,13,19]. Volleyball players often land with a single leg after a spike due to a shift in the center of gravity and loss of balance [20]. This will increase the risk of lower limb injury when landing on a single leg, because the impact and energy transfer can only be absorbed by the muscle tissue of the individual leg in contact with the ground. Compared with landing using both legs, previous studies have shown that landing with a single leg will increase knee and hip flexion and extension moments, peak VGRF and power, which result will increase tension and stress of the knee ligament, thus increasing the risk of ACL injury [21–23]. However, there is a lack of research on the biomechanical differences caused by volleyball players landing on a single-leg after a spike, and it is also not clear that landing following a volleyball spike would make a greater difference than a typical landing.

The injury risk of volleyball players while jumping and landing is determined by the spatial and temporal characteristics of the sport. This has been observed especially during landing after blocking and spiking and increases the probability of ACL injury of volleyball players [24]. In landing tasks, previous studies have shown that ACL injury is believed to occur 30–50 ms after initial ground contact [21]. In the biomechanical analysis of landing tasks that may cause damage, the timing of post-landing damage needs consideration. However, current kinematic and kinetic analyses only partially focus on the peak point data and assume that the peak point time in line with the critical time at the injury occurs. Jeffrey and Michael et al. compared and analyzed the peak joint angle and moment (knee and hip flexion, knee and hip moment) between single-leg and double-leg landing tasks [25–28]. Most of them focus on analyzing peak data. Wang and Rachael et al. also compared the peak VGRF and peak knee proximal tibia shear force during different landing tasks [19,29,30].

This analysis method is widely accepted in research, but the comparative peak misses the important time fluctuations that may happen in the whole landing phase. At present, great progress has been made in the study of biomechanics spatial and temporal variability, and a relatively complete model framework has been presented [11,31,32]. Statistical parametric mapping (SPM) is a method, which can test the statistical differences of continuous data such as kinematics and kinetics over the whole period of motion to calculate accurately the significance threshold [33,34]. Therefore, based on the one-dimensional character of joint motion changing with time, combining traditional discrete analysis with statistical parametric mapping of one-dimensional (SPM 1d) was used to conduct statistical analysis on the sagittal biomechanical data of the landing task of volleyball players. This can reveal ACL injury mechanics of volleyball players during landing tasks after a spike and provide injury prevention strategies.

Therefore, the main aim of our study was to investigate the VGRF and kinematics and kinetics of the lower extremity sagittal plane movement in the whole landing stage of volleyball players following a spike landing with single-leg landing (SL) and double-leg landing (DL). This will be achieved using traditional discrete kinematic and kinetics analysis and statistical parametric mapping of one-dimensional (SPM 1d) analysis. We hypothesized that significant biomechanical differences may be observed in the sagittal plane between SL and DL after a spike. Compared with DL, the SL could show smaller knee and hip flexion angles and angular velocities as well as greater moment, power, VGRF and loading rates. Meanwhile, most of these differences may be observed within the 30–50 milliseconds time frame of initial contact with the ground. Our hypothesized results may provide some clues to reveal the injury mechanism of ACL from the temporal kinematic and kinetic differences during landing phase.

2. Materials and Methods

2.1. Participants

The sample size was calculated using the power package RStudio 3.6.1. A power analysis for Paired-samples *T*-test analysis of variance was performed for an effect size of 0.8 (significance level: 0.05; power: 0.8; alternative: two sided). The power analysis indicated that thirteen subjects (the minimal detectable difference threshold values ranged from 11.2 to 14.8) should be recruited for the study, but we only recruited twelve subjects who were eligible for this study. Therefore, twelve healthy semi-professional male volleyball players were recruited to participate in this experiment (age: 23.3 ± 2.2 years, height: 1.87 ± 0.04 m, weight: 80.2 ± 6.3 kg). All were members of the volleyball team from the local university. They were also National Division 2 athletes or above. Subject exclusion criteria included: (1) a history of foot diseases and deformity or lower limbs surgery; (2) lower extremity injuries or medical problems that might affect athletic performance in the six months prior to the trial. All participates had the objectives, demands, experimental procedures and other details of this study explained to them in detail, and all participates provided written informed consent prior to the experiment. The study protocol was approved by the Human Ethics Committee of Ningbo University (Approval Number: RAGH20200121). All experimentation was performed at the same day, from 8:00 to 21:00 (about 13 h).

2.2. Instrumentation

Before the formal test, the height and weight of all subjects were recorded with a calibrated weighing scale and stadiometer. A Vicon system with 8 cameras (Oxford Metric Ltd., Oxford, UK) was used to collect kinematic data during the test, and the data were sampled at a frequency of 200 Hz. An in-ground force plate (AMTI, Watertown, MA, USA) was used to obtain kinetic data, and the sampling frequency was set at 1000 Hz. Vicon Nexus software was used to synchronously capture force plate data and kinematics data. As shown in Figure 1, a total of thirty-six reflective markers with diameter of 12.5 mm were placed on the lower extremities and pelvis to track the motion trajectory according to the experimental requirements and details from a previous study [35]. The reflective markers were located in the bilateral lower limbs and pelvis, which included anterior and posterior (left and right) superior iliac spine, medial and lateral (left and right) condyle, medial and lateral (left and right) malleolus, first and fifth (left and right) metatarsal heads, distal interphalangeal joint (left and right) of the second toe, and six clusters were placed on the middle and lateral (left and right) heel, shank and thigh. The data set of initial coordinates were obtained via collected static experimental data on the subjects in an anatomically neutral position. All negative sagittal data represent flexion or plantarflexion, and all positive data represent extension or dorsiflexion. The data for joint motions were calculated using the Euler equation [36].

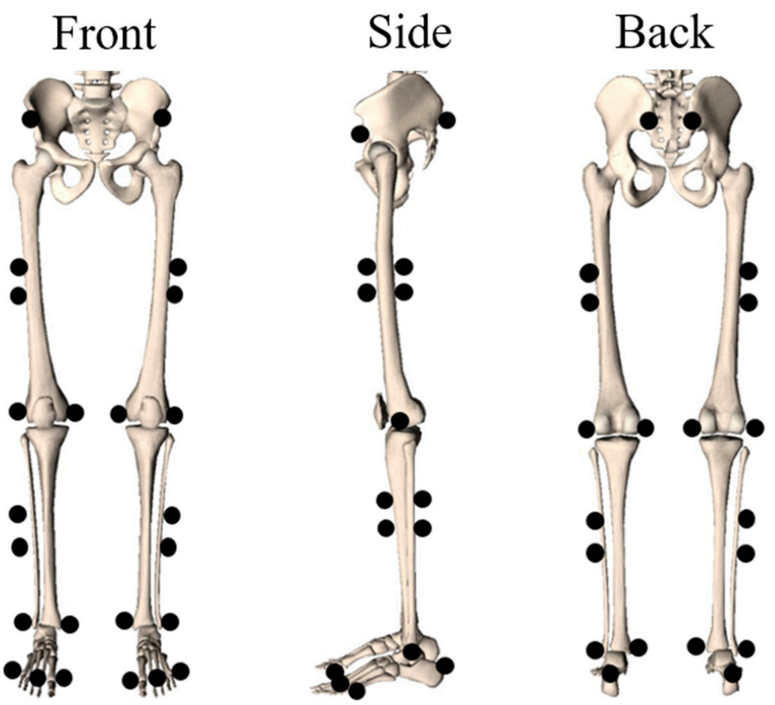

Figure 1. Illustration of lower limb and pelvis reflective markers placement.

2.3. Experimental Procedure

Figure 2 outlines the experimental process. The whole experimental environment was surrounded by eight infrared cameras and the force plate was located between the middle line and the attack line as shown in the figure. Before the formal experiment, participants warmed up for 10 min in a manner of their choice. Following the warm-up, the participants were familiarized with the experimental environment and experimental process and spike landing practice preceded the formal experiment. To avoid the error caused by the differences of participants' flight-time and flight-distance in the air, all participants were asked to take off from force plate and also land on the force plate. All participants were asked to find the most appropriate way (including the most appropriate way to take off, the most appropriate distance between the volleyball and participants) to straight spike during the spike landing practice.

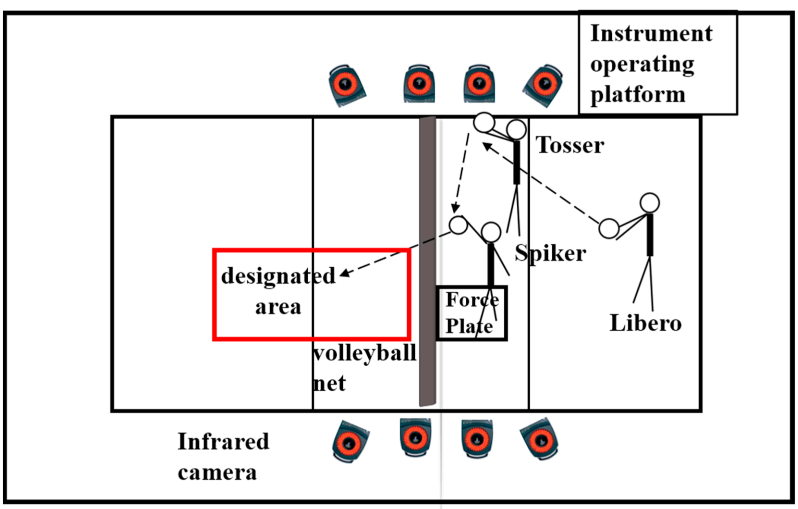

Figure 2. Illustration of the spike landing process.

The average spike height of each subject during the match was recorded, and the height was then marked as the subject′s spike height in the experiment. In the spike landing task, the libero held the ball and passed it to the tosser, and the tosser then passed the ball directly in front of the participant (the height of the ball is the height recorded). The participant took off in their own individual way and then spiked the ball with maximal effort on the top of the force plate. After spiking the ball, the subject then landed on the force plate. All participants were asked to use the straight spike and spike the ball into a designated area (Figure 2, red frame area).

The SL is outlined in Figure 3, the participants were asked to land on the force plate with their supporting leg (the supporting leg defined as a leg that is often used for support after landing [37]). The other leg did not contact with the ground to simulate landing with single leg as in an official match. For the DL task (Figure 3), the participants were asked to use the supporting leg to land on the force plate, and the other leg to be placed outside the force plate simultaneously. A successful trial of landing tasks was defined as the subjects not falling after landing, the complete spike sequence agreed with the rules of volleyball, and the ball landed in the designated area. Between each spike landing, participants rested for 30 s to reduce the error due to any acute fatigue caused by successive jumping and landing tasks. Participants were asked to wear professional volleyball shoes during the experiment, and they were randomly (by simple randomization: the subjects were randomly assigned directly) assigned to SL or DL tasks, a total of ten successful experiments were carried out (SL: five times, DL: five times).

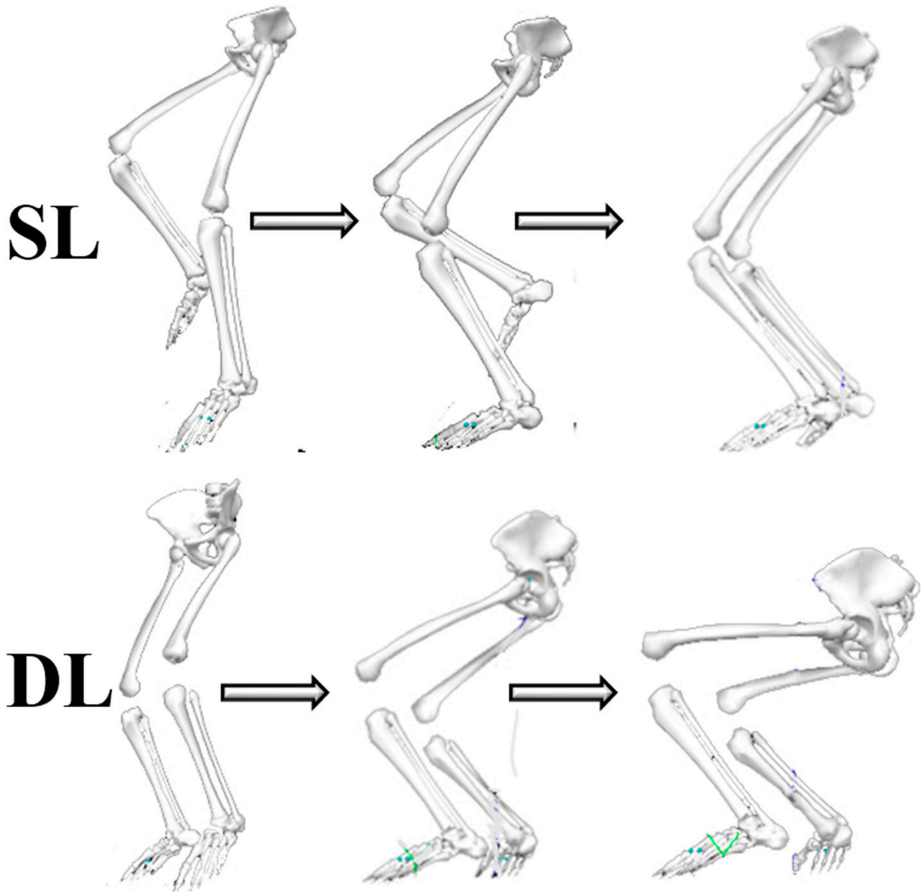

Figure 3. The process of single leg landing (SL) and double-leg landing (DL) following a spike.

2.4. Data Collection and Processing

The force platform had a value of vertical ground reaction force (VGRF) surpassing 10 N, and we defined this as the initial point [11]. Data capture started 2 s prior to the initial contact point and ended 3 s after landing, a total of 5 s of data was collected. The filter frequency selection was based on previous studies by Winter [38]. The dynamic residual analysis was executed in the subsets to confirm the most suitable SNR (signal to noise ratio). Finally, the data were filtered using Butterworth low-pass filters (order: fourth, phase lag: zero, kinematics: 10 Hz, kinetics: 20 Hz). The C3D format file was generated using Vicon Nexus software, and then imported into the C3D file into Visual 3-D software (C-Motion Inc., Germantown, MD, USA) for static modeling and further processing.

The joint angle, joint moment, joint angular velocity and joint power were calculated using Visual 3-D software by inverse kinetics algorithm. Joint power was defined as the product of the joint moment and joint angular velocity. The range of joint motion (ROM) was calculated from the difference between the initial contact joint angle (minimum flexion or plantarflexion) and the joint angle at maximum flexion (dorsiflexion). The jump height was calculated by the change of the center of gravity, and the calculation process also conducted using Visual 3-D software. The flight-time was calculated by the change of time point (the starting point: taking off from the force plate; the end point: landing on the force plate), which was conducted using the Vicon system.

The loading rate of VGRF was defined as the growth rate of VGRF over time. Then, we imported all data into MATLAB (Version: R2019a, The MathWorks, Natick, MA, USA) to process by written script. A custom MATLAB script was used to intercept the time point between the initial contact and the peak knee joint flexion during the landing process.

2.5. Statistical Analysis

The landing phase was defined as the maximum knee flexion after initial contact with the ground. Prior to statistical analysis, all data were subjected to the Shapiro-Wilk normality test. If the data showed nonconformity, the Wilcoxon matched-pairs signed-rank test was executed for data analysis.

For traditional analysis (discrete variable), the MATLAB script was written to extract and calculate the ROM of joint angle, peak data of angle, moment, power, angular velocity, and VGRF. Paired-samples T-test was employed to test differences of ROM of joint angle, peak data of angle, moment, power, angular velocity and VGRF. All traditional analyses were conducted using SPSS 24.0 (SPSSs Inc., Chicago, IL, USA), and the statistical significance was set as 0.05. Cohen's d effect sizes (ES) were split into large (ES > 0.8), medium (0.2 < ES < 0.5), and small (ES < 0.2) benchmark sizes [39].

For SPM 1d analysis, each task generated a separate curve prior to analysis. Then, all data were extracted to expand into 101 data points time series curves (using a custom MATLAB script). The 101 data points represent the landing phase (0–100%). The open-source script of SPM 1d (Paired-samples *T*-test, significance threshold was 0.05) was used in the statistical analysis [34].

3. Results

For the time of landing phase, there were no differences (p = 0.31) between SL (193.6 ± 29.8 ms) and DL (197.1 ± 33.4 ms) landing task. For the jump height, there were no differences (p = 0.22) between SL (0.59 ± 0.08 m) and DL (0.62 ± 0.10 m) landing task. For the flight-time in the air, there were no differences (p = 0.09) between SL (0.66 ± 0.09 s) and DL (0.71 ± 0.12 s) landing task.

For the results of ankle joint. SPM analysis revealed that SL depicted a significantly greater plantarflexion angle than DL during the 0–41.07% (SL: −19.8~24.6°, DL: −8.6~29.7°, p < 0.001) landing phase, a significantly greater plantarflexion moment than DL during the 2.76–79.45% (SL: −1.01~−1.41 Nm/kg, DL: −0.45~−0.52 Nm/kg, p < 0.001) landing phase, a significantly greater joint power than DL during the 2.48–9.66% (SL: −5.20~−31.37 W/kg, DL: −2.12~−10.25 W/kg, p < 0.001) and 12.92–26.95% (SL: −19.71~−2.89 W/kg, DL: −6.14~−0.65 W/kg, p < 0.001) landing phase, a significantly greater dorsiflexion angular velocity than DL during the 0–23.54% (SL: 756.3~102.4°/s, DL: 509.9~18.2°/s, p < 0.001) and 44.73–47.63% (SL: 16.2~12.0°/s, DL: 17.4~21.1°/s, p = 0.016) landing phase (Figure 4).

SL (Table 1) depicted greater peak plantarflexion angle (p = 0.001), ROM (p = 0.023), peak moment (p = 0.001), peak power (p = 0.001), and peak angular velocity (p = 0.001).

For the results of the knee joint. SPM analysis revealed that SL depicted a significantly smaller flexion angle than DL during the 0–100% (SL: −7.0~−72.2°, DL: −20.7~−97.6°, p < 0.001) landing phase, a significantly smaller flexion angular velocity than DL during the 0–32.69% (SL: −293.9~−157.9°/s, DL: −407.3~−275.9°/s, p < 0.001) and 44.36–55.40% (SL: −107.1~−43.9°/s, DL: −195.8~−133.6°/s, p = 0.014) landing phase, a significantly greater extension moment than DL during the 3.86–4.27% (p = 0.048) and 11.73–12.32% (p = 0.045) and 17.74–21.82% (3.03~4.16 Nm/kg, DL: 2.73~2.49 Nm/kg, p < 0.001) landing phase, a significantly greater joint power than DL during the 16.83–20.45% (SL: −25.9~−29.3 W/kg, DL: −16.4~−23.8 W/kg, p = 0.006) landing phase (Figure 4). SL (Table 1) depicted a smaller peak flexion angle (p = 0.001) and peak angular velocity (p = 0.001), greater peak moment (p = 0.005), and peak power (p = 0.002).

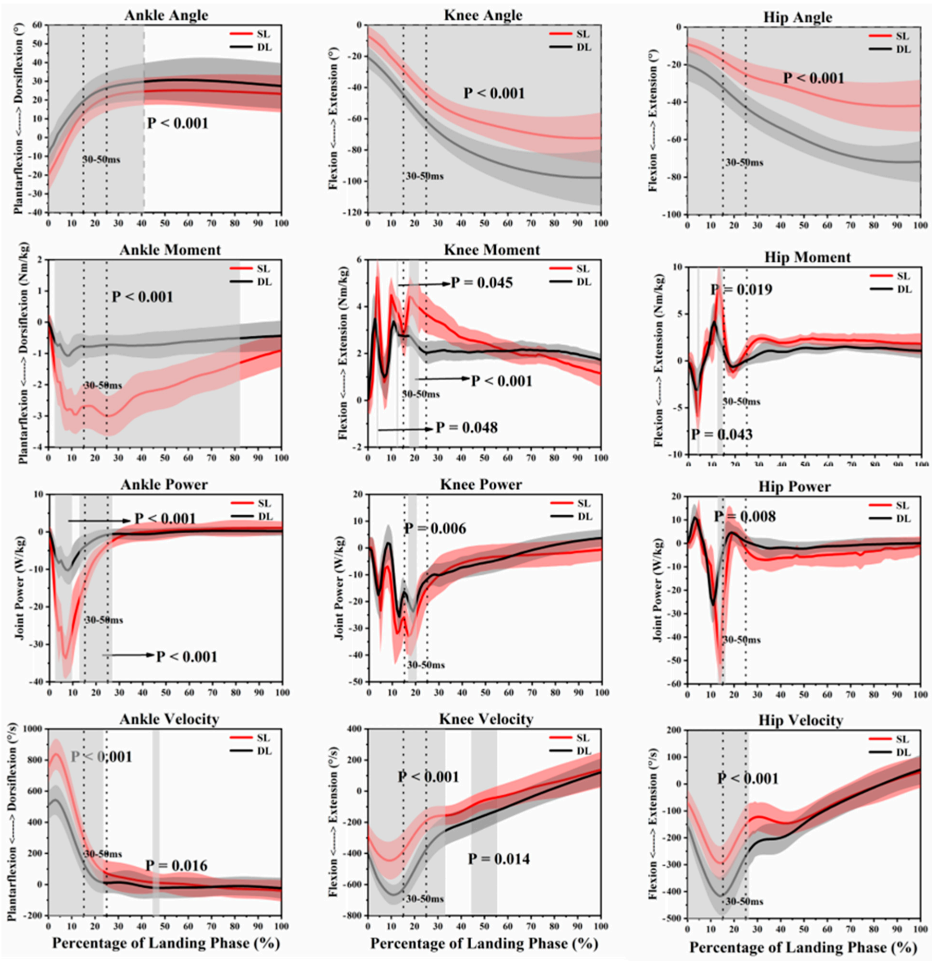

Figure 4. The Statistical Parametric Mapping (SPM) results between single-leg landing (SL) and double-leg landings (DL) tasks following a spike, depicting the mean angle moment power and angular velocity and standard deviation of ankle knee and hip sagittal plane. Two dashed lines are the 30 and 50 ms time frame of initial contact. Grey shaded areas indicate that there are significant differences (p < 0.05) between SL and DL during the spike landing phase.

Table 1. Means ± standard deviations for the peak angle moment power and angular velocity of ankle knee and hip sagittal plane, peak VGRF and loading rate between SL and DL.

		SL	DL	*p*
Ankle	Peak Ankle Plantarflexion Angle (°)	−19.88 ± 8.80 * (initial contact)	−8.58 ± 8.28 * (initial contact)	0.001
	Peak Ankle Dorsiflexion Angle (°)	25.93 ± 7.94 (193.6 ± 29.8 ms)	34.31 ± 14.49 (197.1 ± 33.4 ms)	0.058
	Ankle ROM (°)	45.81 ± 6.38 *	42.89 ± 7.98 *	0.023
	Peak Ankle Moment (Nm/kg)	−3.26 ± 0.55 * (24.2 ± 3.2 ms)	−1.12 ± 0.37 * (18.5 ± 2.3 ms)	0.001
	Peak Ankle Power (W/kg)	−37.97 ± 5.44 * (16.7 ± 2.8 ms)	−11.20 ± 4.15 * (18.7 ± 3.3 ms)	0.001
	Peak Ankle Angular Velocity (°/s)	844.91 ± 95.22 * (8.6 ± 1.7 ms)	559.49 ± 96.93 * (8.4 ± 1.3 ms)	0.001
Knee	Peak Knee Flexion Angle (°)	−74.26 ± 15.18 * (193.6 ± 29.8 ms)	−99.83 ± 17.58 * (197.1 ± 33.4 ms)	0.001
	Peak Knee Extension Angle (°)	−6.96 ± 6.39 * (initial contact)	−20.73 ± 6.84 * (initial contact)	0.003
	Knee ROM (°)	67.30 ± 10.99	79.10 ± 17.33	0.056
	Peak Knee Moment (Nm/kg)	5.52 ± 1.79 * (9.7 ± 3.5 ms)	3.88 ± 1.50 * (8.5 ± 1.9 ms)	0.005
	Peak Knee Power (W/kg)	−42.77 ± 5.77 * (26.3 ± 4.1 ms)	−29.44 ± 5.57 (27.8 ± 4.4 ms)	0.002
	Peak Knee Angular Velocity (°/s)	−464.22 ± 80.49 * (20.6 ± 3.6 ms)	−671.56 ± 77.83 * (24.3 ± 2.8 ms)	0.001
Hip	Peak Hip Flexion Angle (°)	−43.01 ± 12.82 * (193.6 ± 29.8 ms)	−74.31 ± 10.87 * (197.1 ± 33.4 ms)	0.001
	Peak Hip Extension Angle (°)	−9.21 ± 4.09 (initial contact)	−20.04 ± 8.61 (initial contact)	0.021
	Hip ROM (°)	33.80 ± 9.52 *	54.27 ± 13.04 *	0.001
	Peak Hip Moment (Nm/kg)	7.98 ± 2.93 * (28.2 ± 5.7 ms)	4.35 ± 1.99 * (24.3 ± 4.5 ms)	0.007
	Peak Hip Power (W/kg)	−57.90 ± 10.48 * (30.1 ± 6.2 ms)	−30.49 ± 6.91 * (24.7 ± 4.8 ms)	0.001
	Peak Hip Angular Velocity (°/s)	−302.56 ± 68.56 * (30.2 ± 3.7 ms)	−426.14 ± 75.13 * (29.7 ± 4.3 ms)	0.001
VGRF	Peak VGRF (BW)	5.08 ± 0.68 * (48.3 ± 6.6 ms)	3.31 ± 0.55 * (49.8 ± 7.8 ms)	0.001
	Loading Rate (BW/s)	187.23 ± 21.11 *	118.49 ± 14.78 *	0.001

Note: The time corresponding to the occurrence of the peak data point is in parentheses, and the unit is milliseconds (ms). * refer to significance with $p < 0.05$.

For the results of the hip joint, SPM analysis revealed that SL depicted a significantly smaller flexion angle than DL during the 0–100% (SL: −9.2~−41.8°, DL: −20.0~−71.7°, $p < 0.001$) landing phase, a significantly smaller flexion angular velocity than DL during the 0–26.34% (SL: −69.2~−150.1°/s, DL: −157.2~−267.7°/s, $p < 0.001$) landing phase, a significantly greater joint moment than DL during the 3.80–4.60% ($p = 0.043$) and 12.45–14.49% (3.14~7.98 Nm/kg, DL: 4.17~2.37 Nm/kg, $p = 0.019$) landing phase, a significantly greater joint power than DL during the 13.01–16.26% (SL: −29.2~−30.1 W/kg, DL: −21.9~−4.2 W/kg, $p = 0.008$) landing phase (Figure 4). SL (Table 1) that SL depicted a smaller peak flexion angle ($p = 0.001$) and ROM ($p = 0.001$) and peak angular velocity ($p = 0.001$), greater peak moment ($p = 0.007$) and peak power ($p = 0.001$).

For the results of VGRF. SPM analysis revealed that SL depicted a significantly greater VGRF than DL during the 5.87–8.25% (SL: 1.62~1.43 BW, DL: 1.43~1.09 BW, $p = 0.029$), 19.75–24.14% (SL: 3.43~4.71 BW, DL: 2.28~2.94 BW, $p = 0.003$) and 47.09–52.46% (SL: 1.79~1.93 BW, DL: 1.15 ~1.18 BW, $p = 0.004$) landing phase (Figure 5). Table 1 shows that SL depicted greater peak VGRF ($p = 0.001$) and loading rate of VGRF ($p = 0.001$) than DL.

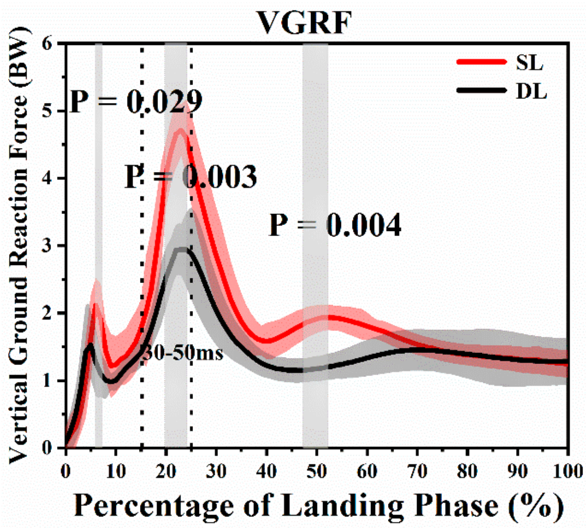

Figure 5. The Statistical Parametric Mapping (SPM) results between single-leg landing (SL) and double-leg landings (DL) tasks following a spike, depicting the mean vertical ground reaction force (VGRF) and standard deviation. Two dashed lines are the 30 and 50 ms time frame of initial contact. Grey shaded areas indicate that there are significant differences ($p < 0.05$) between SL and DL during the spike landing phase.

4. Discussion

We hypothesized that compared with DL, the SL will show smaller knee and hip flexion angles and angular velocities as well as greater moment, power, VGRF, and loading rates. Meanwhile, the SL will show greater ankle plantarflexion angle and dorsiflexion angular velocity, and most of these differences were within the 30–50 milliseconds time frame of initial contact with the ground. Our results agree in part with the proposed hypothesis. SPM demonstrated that most differences occur in the first third of the landing phase. Our results are consistent with ACL injury mechanism in landing tasks, and SL may increase the probability of ACL injury risk.

The wide ROM of the knee and hip sagittal plane are key to lower limb cushioning during landing [17]. Smaller knee flexion during landing can result in lower limb shock absorption, leaving the ACL at a higher risk of injury [40–43]. Our results revealed that SL depicted a smaller knee flexion angle (about 13.8°) and hip flexion angle (about 10.8°) than DL during the landing phase, and SL also has resulted in a flexion angle of less than 45° within the 30–50 ms time frame of initial contact. Markolf and Drahanich et al. both found that the quadriceps force will augment ACL loading, when the knee joint flexion angle is from 0° to 45° [18,44]. Therefore, our current evidence pointed out that the quadriceps force of SL seems to increase the load on the ACL. Meanwhile, SL appears to be a hard landing with the smaller hip flexion angle, and the results of hard landing patterns are followed by an increase in knee and hip moments. Within the 30–50 ms time frame of the initial landing phase, we found that the knee moment of SL is about 2 Nm/kg greater than DL during the 17.74–21.82% (about 35–43 ms after initial ground contact) landing phase. Consistent with previous studies, smaller flexion angles are accompanied by larger moments, and this increases the risk of ACL injury [45]. At the same time, SL depicted a greater negative knee joint power and hip joint power during the landing phase. Previous studies have shown that the case of the negative knee and hip joint power suggests that eccentric work is done by the extensor muscles of the knee and hip joint to consume the impact of energy [12]. With the increase of negative joint power, this means that the eccentricity to the joint extensor increases. The strain on the ACL will increase with the eccentricity to the joint extensor (quadriceps femoris and anterior tibia) then the risk of ACL injury will increase [13,45].

The knee joint plays a significant role in absorbing shock during contact between the body and the ground [12]. During the landing task, the increased VGRF will also increase the contractile force of the quadriceps femoris, thus increasing the load on the ACL [13]. Previous studies have shown that higher VGRF and insufficient VGRF dissipation during the landing phase may also be a risk factor for ACL injury [12,46]. Our results revealed that the SL depicted a significantly greater VGRF than DL during the 19.75–24.14% (38–47 ms after initial ground contact) landing phase, and SL also depicted greater peak VGRF (p = 0.001) and loading rate of VGRF (p = 0.001) than DL. Cerulli and Lamontagne found that the maximum ACL tension occurred at the moment of the peak VGRF [44,47]. A higher VGRF loading rate indicates a greater impact in a shorter period, thus increasing the risk of ACL injury with a high probability [3,48,49]. Yu et al. found that the angular velocity of the knee and hip flexion was negatively correlated with the peak VGRF during landing tasks, the active knee and hip flexion may be the key factor to reduce the impact force [13]. Consistent with our results, SL depicted a smaller knee and hip flexion angular velocity during the 0–32.69% and 0–26.34% landing phases respectively. It has been shown that increasing the angular velocity of hip and knee flexion during landing can be reduce VGRF, especially in the early landing phase. The difference in VGRF and angular velocity during the landing phase also occurred within the 30–50 ms time frame. This also seems to be associated with ACL injury mechanisms.

During the landing task, the lower limb undergoes a load pattern of distal-to-proximal in which the foot and ankle joint are loaded primarily, and then the load is dissipated to the knee and hip joints. At the moment of initial contact during the landing phase, a greater ankle plantarflexion angle will increase the time to reach the peak VGRF, and a larger plantarflexion landing method can use plantar flexor muscles for attenuation buffering according to previous studies [50,51]. It is also thought to increase lower extremity shock absorption and reduce non-contact ACL injuries [50]. Our results revealed that SL depicted a greater plantarflexion angle (within 83 ms after initial ground contact), power (about 5–54 ms after initial ground contact) during the landing phase. It is worth noting that the time at which these differences occur is also within the framework of the time at which the ACL is thought to occur. The ankle joint in the SL shows higher energy dissipation due to a larger plantarflexion landing method, thus the load impact on the knee joint is relatively reduced. Although the initial contact plantarflexion angle of the ankle joint affects the ROM and peak angular velocity of the knee joint, the knee joint always maintains a constant energy dissipation contribution rate [52,53]. Previous studies have shown that the total energy dissipation of the lower extremities is greater with soft landing (larger knee and hip flexion, ankle plantarflexion) techniques than with hard landing techniques, increasing the total energy dissipation of the lower limb during landing will reduce the potential injury risk of passive joint structure [12,14,54]. Therefore, a landing strategy of consciously increasing the knee and hip flexion, plantarflexion of ankle should be considered by volleyball players in order to reduce non-contact ACL injuries.

There are some limitations that should be considered in the present study. Firstly, we did not use electromyography (EMG) to compare and analyze the muscle activity during the spiking landing task. The muscle activity of the knee extensor also has great significance to this study, and EMG data should be considered for subsequent studies. Secondly, the subjects in our experiment did not include female volleyball players. Previous studies have shown differences between male and female athletes during landing [41,53]. Therefore, female subjects should be considered in future studies. Finally, although we attempted in the experiment to simulate real volleyball match conditions, it is unavoidable that some parameters (e.g., the distance between the ball and participant's hand, the spiked volleyball speed in the air) cannot be quantified. The sample size of our study included twelve semi-professional male volleyball players. This may have contributed to the limitations, so increasing the sample size in the future studies could further verify the results presented here.

5. Conclusions

In summary, this study presents the kinematic and kinetic differences of the lower limb sagittal plane of motion throughout a volleyball players spike landing phase of single and double leg landing. Appendix A exhibits the STROBE Checklist for reporting this study (Appendix A: Table A1). The results of our study are partially consistent with previous studies about typical landings. Following a spike, landing with a single leg shows a smaller flexion angle and angular velocity of the knee and hip joints, a greater joint moment, joint power, VGRF and loading rate. Most of these differences may increase the risk of ACL injury and fall within the time range of ACL injury. SL depicted a smaller knee flexion angle (about 13.8°) and hip flexion angle (about 10.8°), and SL also showed a flexion angle of less than 45° within the 30–50 milliseconds time frame of initial contact with the ground. Therefore, a landing strategy of consciously increasing knee flexion and hip flexion, and plantarflexion of the ankle, should be considered by volleyball players in order to reduce non-contact ACL injuries.

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Data Availability Statement: Data available in a publicly accessible repository.

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

Appendix A

Table A1. An adapted STROBE Checklist for reporting this study.

Section/Topic	Item Number	Recommendation
Title and abstract	1	Provide in the abstract an informative and balanced summary of "The Risk of Anterior Cruciate Ligament Injury (ACL) Following Different Volleyball Spike Landing Ways"
Introduction		
Background/rationale	2	Explain the scientific background and rationale about ACL injury following different volleyball spike landing ways for the investigation being reported
Objectives	3	State specific objectives, including any prespecified hypotheses
Methods		
Participants	4	Report numbers of individuals of study (e.g., numbers potentially eligible, examined for eligibility, confirmed eligible, included in the study)
Instrumentation	5	Describe the hardware and software equipment used to complete the study (e.g., Vicon motion capture system, AMTI force plate, MATLAB R2019a)

Table A1. *Cont.*

Section/Topic	Item Number	Recommendation
Study design	6	Present key elements of this experimental procedure
Data collection and processing	7	Describe the data collection and processing process of this study
Statistical analysis methods	8	Describe all statistical analysis method
Results		
Main results	9	Present main results of this study
Visual data	10	Present figures and tables of key data
Discussion		
Key results	11	Summarize key results with reference to study objectives
Limitations	12	Discuss limitations of the study, taking into account sources of potential bias (e.g., lack of female subjects, small sample size)
Conclusion	13	Provide a general interpretation of results and discuss a landing strategy to avoid ACL injury
Other information		
Author contributions	14	Describe author contributions of this study
Funding	15	Present the funding of this study

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