Applied Bionics and Biomechanics

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Research Article

The Application of Statistical Parametric Mapping to Evaluate Differences in Topspin Backhand between Chinese and Polish Female Table Tennis Players

Ziemowit Bańkosz and Sławomir Winiarski

Department of Biomechanics, Faculty of Physical Education and Sports, University School of Physical Education in Wroclaw, Poland

Correspondence should be addressed to Sławomir Winiarski; slawomir.winiarski@awf.wroc.pl

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The research is aimed at comparing the kinematics (the movement pattern in the most important joints and accelerations of the playing hand) between female table tennis players coached in Poland (POL) and China (CHIN) during the performance of a topspin backhand stroke (so-called quick topspin). The study involved six female table tennis players at a high sports skill level, playing in Poland's highest league. Three were national team members of Poland (age: 20.3 ± 1.9), while the other three were players from China (age: 20.0 ± 0.0). Kinematics was measured using MR3 myoMuscle Master Edition system—inertial measurement unit (IMU) system. The participants performed one task of topspin backhand as a response to a topspin ball, repeated 15 times. Statistical parametric mapping (SPM) was calculated using SPM1D in a Python package that offered a high-level interface to SPM1D. The SPM method allowed for the determination of differences between the Chinese and Polish female athletes. The differences found are probably mainly due to differences in the training methodologies caused by different coaching systems. The observed differences include, among others, greater use of the so-called small steps in order to adapt and be ready during the back to ready position and backswing phases, which gives the CHIN players slightly better conditions for preparation for the next plays. The CHIN players' position compared to that of the POL players favours a quicker transition from the backhand to the forehand play. This difference is probably related to the difference in the dominant playing styles of the groups studied. Despite the differences in movement patterns in both groups, the exact value of playing hand was achieved. This may be a manifestation of the phenomenon of equifinality and compensation. All the differences found are probably mainly due to differences in the training methodologies caused by different coaching systems.

1. Introduction

As a very complex and multifaceted sport, table tennis is characterized by various strokes, legwork techniques, tactical solutions and playing styles, and a multitude of solutions for an almost infinite number of game situations. The main groups of strokes that yield points are topspin strokes, introduced to the game in the 1950s by Japanese players [1]. Players use many variations of topspin strokes in their game (e.g., backhand and forehand strokes, differing in strength involved, speed of ball rotation, flight trajectory, ball speed, and the moment of hitting the ball) depending on the solution used or the need to adjust to ball parameters. Table tennis players must also adjust their position to the ball using a different kind of footwork, changing kinematics and kinetics characteristics of body segments [2, 3]. These differences lead to a large variety and variability of movement in this sport. Issues related to movement variability have recently been quite often addressed in the literature. Traditionally, movement variability is considered to reflect the "noise" in the system of human movements, while learning a given activity requires decreasing variability as it is perceived as incorrect [4, 5]. Movement variability is also viewed and considered a normal phenomenon, resulting from the diversity and variability present in the entire biological system used by humans, and its occurrence is associated with adaptive and functional processes [6]. Movement variability has been explained using many theories available in the literature, such as generalized motor program [7, 8], GMP-uncontrolled manifold (UCM) [9], and dynamic systems theory [5]. Assessing the occurrence and scale of movement variability appears to be extremely important in the sports training process. It seems to be also critical in the process of improving skills of purposive movements and explaining how to control human movements. Linear measures have been used in the assessment of variability, such as standard deviation, range, or coefficient of variation. Taking into account discrete (numerical) or serial data, i.e., continuous and changing over time, would improve the assessment of variability. This is because when assessing movement coordination, for instance, the change of the angle in a given joint over time, and comparing the repetitions by one or many people, a method should be used to compare time waveforms rather than just single, selected parameters. Such criteria are met by the statistical parametric mapping (SPM) method. It is the gold standard statistical method dedicated to numerical signal data analysis. For the one-dimensional variables recorded with the motion analysis system, the general SPM model can be simplified to the one-dimensional model SPM1D. This method and its characterization were presented in previous studies [10].

The assessment of variability of movement seems to be important in table tennis, which is a very complex sport, where technique and its improvement are the essential elements used to achieve the champion level, with the basis of the technique being a stroke and precise hitting the ball with the racket. The few available studies on table tennis and the variability of movement have been based on the methods of evaluation of standard deviation, correlation, and analysis of variance (ANOVA and least significant difference (LSD)) and presented UCM calculations. Iino et al. emphasized that the possibility of using different configurations in the evaluated joints to stabilize the vertical angle of the racket in table tennis strokes can be a critical factor in playing performance [11]. A previous study by Bańkosz and Winiarski also evaluated the variability of movement by analyzing the coefficient of variation of kinematic parameters in selected important moments of the hitting movement [10, 12]. However, the coordination of movements in individual joints was taken into account to a small extent. The variability of temporal and spatial coordination of movements, the possibility of compensation, and functional variability are significant problems in the coaching practice and in the process of teaching and improving technique and its monitoring. Making the coaches and players aware of the different variants of strokes even for a specific solution (e.g., playing with the right strength, speed, and rotation to the same place) seems to be very important and necessary for improving the training process. Therefore, copying and imposing a single pattern of performing the movement seem to be a wrong way. Considering the differences between athletes and looking for individual technical solutions instead would be a better choice [10].

Interpersonal variation of the sports technique may result, for example, from gender differences, differences in anatomical structure, and differences in sports skill level. The diversity of techniques due to the training system also seems to be an interesting issue. Identification of differences and, at the same time, similar or perhaps unchangeable elements of table tennis stroke techniques in athletes coached using different training systems can provide important insights into the technique of performing a given stroke. Some differences in the technique may indicate the possibility of using different solutions in the performance of the stroke, while the same, similar, and unchanging elements may highlight their importance in table tennis. Therefore, the aim of the research was to compare the kinematics between female players coached in Europe (Poland) and Asia (China) during the performance of a topspin backhand stroke. In accordance with the findings of other authors and previous studies [12, 13], it was assumed that, despite the comparable level of players, there are many differences in the kinematics of topspin backhand between them. The greater differences between the players would occur in the joints and segments located farther from the place of the racket contact with the ball (upper body and shoulder joint) than in those closer, located in the playing hand (wrist joint). It can also be assumed that at the key instant of the stroke, which is the moment of maximum acceleration, occurring at around the contact between the racket and the ball, the least differences are observed in players' kinematics.

2. Material and Methods

2.1. Study Design. It was an observational study with adopted retrospective convenience sampling. The minimal sample size of our data was determined in the planning stage of the experiment using the margin of error approach to get results as accurate as needed (with an assumed 5% margin of error at 95% level of confidence and α level of 0.05). The assumed standard deviation was taken from preliminary studies using the same population of interest.

2.2. Participants. The study involved six female table tennis players at a high sports skill level, playing in the highest league in Poland (Ekstraklasa table tennis league). Three of them were national team members of Poland (POL) in the category of adult players (age: 20.3 ± 1.9 y.), while the other three were players from China (CHIN, age: 20.0 ± 0.0 y.), coached within the Chinese training system (i.e., in China). All of the players had more than 10 years of experience in table tennis and presented the offensive style of the game. One player from China was a left hander. Average body height was 161.7 ± 4.5 cm in the group of Polish players and 162.7 ± 4.1 cm in the group of Chinese players, whereas body weight was 59.0 ± 6.9 kg and 56.7 ± 6.4 kg, respectively.

Before the study, all participants were informed about the purpose of the study and the possibility of withdrawing participation at any stage, without giving a reason. All the participants provided informed consent before the research. Pain or recent injury was the exclusion criterion for the study participants. All procedures performed in this study received positive approval from the Senate's Research Bioethics Commission at the University School of Physical Education in Wrocław, Poland (Ethics IRB number 34/2019).

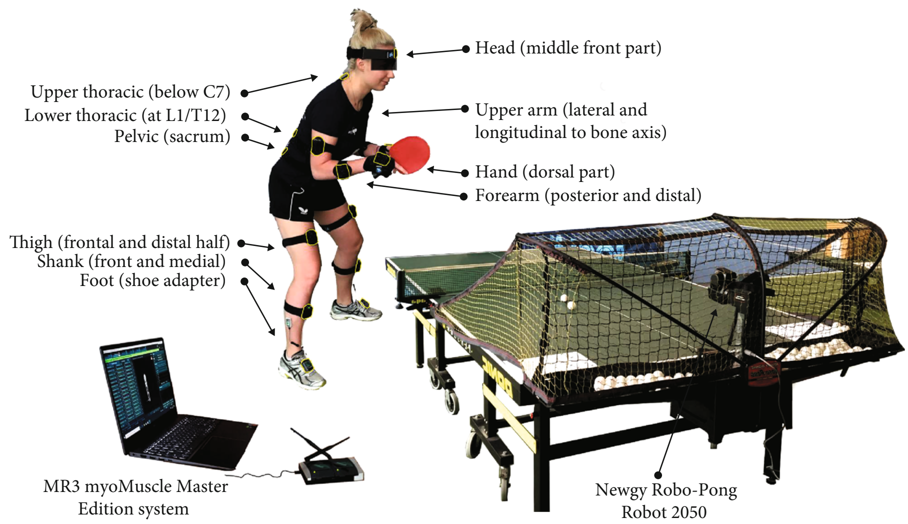

Figure 1: Measurement site.

2.3. Laboratory Set-Up. Kinematics was measured using the MR3 myoMuscle Master Edition system (myoMOTION™, Noraxon, USA, Figure 1). The myoMOTION system consists of a set of (1 to 16) sensors using inertial sensor technology. Based on the so-called fusion algorithms, the information from a 3D accelerometer, gyroscope, and magnetometer is used to measure the 3D rotation angles of each sensor in absolute space (yaw-pitch-roll, also called orientation or navigation angles, [12]). Inertial sensors were located on the body of the study participant to record the accelerations, according to the myoMOTION protocol described in the manual. The accuracy and validity of the inertial measurement unit (IMU) system in angle determination were the subject of the previous research [14, 15].

Sensors were attached with elastic straps and self-adhesive tape. The sensors were placed bilaterally so that the positive x-coordinate on the sensor label corresponded to a superior orientation for the trunk, head, and pelvis (Figure 1). For the limb segment sensors, the positive x -coordinate corresponded to a proximal orientation. For the foot sensor, the x-coordinate was directed distally (to the toes). At the beginning of the measurement, each participant was checked and the system was calibrated according to the manufacturer's recommendations. The recording speed of the piezoelectric sensor was adjusted to the maximal sampling rate for a given sensor (100 Hz per sensor) for the whole 16-sensor set. Noraxon's IMU technology mathematically combines and filters incoming source signals on the sensor level and transmits the 4 quaternions of each sensor. We used system-built fusion algorithms and Kalman filtering (digital bandpass finite impulse response filter (FIR)). This mode allowed direct access to all unprocessed raw IMU sensor data.

2.4. Experimental Procedures. The participants performed one task of topspin backhand (TBH) as a response to a topspin ball, repeated 15 times. Each player was asked to hit the ball in the early stage of its flight (so-called quick topspin) and to reach the marked area in the corner of the table (30×30 cm) diagonally (after instruction: "play diagonally, accurately, and as quick as you can"). After video analysis, only successful shot considered "on table" and played diagonally was recorded for further calculations (missed balls, balls hit out of bounds, and balls hit into the net were excluded). The balls were shot by a dedicated table tennis robot (Newgy Robo-Pong Robot 2050, Newgy Industries, Tennessee, USA, Figure 1) at constant parameters of rotation, speed, direction, and flight trajectory. The settings of the robot were as follows:

(i) Rotation type: topspin

(ii) Speed (determines both speed and spin, where 0 is the minimum and 30 is the maximum): 18

(iii) Left position (leftmost position to which the ball is delivered): 15

(iv) Wing (robot's head angle indicator): 7.5

(v) Frequency (time interval between balls thrown): 1.4 s

Each player had had three to five familiarization trials before the task. The same racket with the following characteristics was used for the experiment: blade, Jonyer-H-AN (Butterfly, Japan); rubber, Tenergy 05, 2.1 mm (Butterfly, Japan); Plastic Andro Speedball 3S 40+ balls (Andro, Germany); and a Stiga Premium Compact table (Stiga, Sweden).

2.5. Kinematics. A total of 90 cycles of topspin backhand stroke were studied. Based on the ISB recommendations concerning the definitions of the joint coordinate system of various joints for the reporting of human joint motion [16, 17], the following angles (measured in degrees) were chosen for both sides and sampled every 0.01 percent of cycle time:

(i) Ankle dorsiflexion/plantar flexion (AFE): rotation of the foot with respect to the tibia coordinate system in the sagittal plane; a negative sign denotes plantar flexion (extension) and positive sign dorsiflexion (flexion)

(ii) Ankle abduction-adduction: movement of the foot away or towards the midline of the body; a negative sign denotes adduction while positive sign abduction

(iii) Ankle inversion-eversion: rotation of the foot around its long axis; a negative sign denotes eversion (away from the median plane) while positive sign inversion (towards the median plane)

(iv) Knee flexion-extension (KFE): movement of the tibia with respect to the femur coordinate system in the sagittal plane; a negative sign denotes extension and positive flexion

(v) Hip flexion-extension (HFE): movement of the femur with respect to the pelvis coordinate system in the sagittal plane; a negative sign denotes extension while positive flexion

(vi) Hip abduction-adduction (HAA): movement of the femur with respect to the pelvis coordinate system in the frontal plane; a negative sign denotes adduction while positive abduction

(vii) Hip internal-external rotation (HIER): internal or external movement of the femur with respect to the pelvis coordinate system in the transversal plane; a negative sign denotes internal while positive external rotation

(viii) Lumbar internal-external rotation (LIER): internal or external movement of the loins in the transversal plane; a negative sign denotes internal while positive external rotation

(ix) Thoracic internal-external rotation (ThIER): internal or external movement of the thorax relative to global coordination system in the transversal plane; a negative sign denotes internal while positive external rotation

For the upper extremity (playing side), a simplified biomechanical model was adopted based on the predominant plane of movement as described by Wu et al. [17] with segments of interest being the thorax, clavicle, scapula, humerus, forearm, and carpus of the hand. Based on the adopted sequence of Euler angles, the following angles were computed:

(i) Shoulder flexion-extension (ShFE): movement of the humerus relative to the thorax in sagittal plane; negative sign denotes extension while positive flexion

(ii) Shoulder abduction-adduction (ShAA): movement of the humerus relative to the thorax in the frontal plane; negative sign denotes adduction while positive abduction

(iii) Shoulder internal-external rotation (ShIER): movement of the humerus relative to the thorax in the transversal plane; a negative sign denotes internal (medial) while positive external (lateral) rotation

(iv) Elbow flexion-extension (EFE): movement of the forearm relative to the humerus along the transversal axis; negative sign denotes (hyper)extension while positive flexion

(v) Wrist flexion-extension (WFE): movement of wrist relative to the radius along the transversal axis and measured between upper arm and hand sensors; a negative sign denotes extension while positive flexion

(vi) Wrist supination-pronation (WSup): movement of wrist relative to the radius along the axis and measured between the upper arm and hand sensors; pronation is a positive rotation and supination is a negative rotation

(vii) Wrist radial abduction-adduction (WRad): movement of wrist relative to the radius and measured between the upper arm and hand sensors; adduction (or ulnar deviation) is negative while abduction (or radial deviation) is positive

The movement of the playing hand was used to assess specific events of the cycle:

(i) Ready position, where the hand is not moving after the previous stroke, just before the swing

(ii) Backswing, which is the moment when the hand changes direction from backward to forward in the sagittal plane after the swing

(iii) Accmax, which is the moment of maximum acceleration of the hand and the moment when the hand reaches the maximum acceleration

(iv) Forward, which is the moment when the hand changes the direction from forward to backward in the sagittal plane after the stroke (the end of the cycle and the beginning of the next cycle)

The phases between defined events were as follows: back to ready position phase (between the forward and ready position), backswing phase (between ready position and backswing), hitting phase (between backswing and Accmax), and forward end phase (between Accmax and forward). The timing of events was analyzed and compared between the POL and CHIN players.

2.6. Statistical Analysis. Statistical calculations were performed using Statistica 13.1 (TIBCO Software Inc.). The sample size was estimated using recommendations postulated by Kontaxis et al. [18]. The statistical power was sufficient to detect the described differences. Power analysis of discrete data was performed to estimate the SPM test power. For the extracted data and for the significant changes ($\alpha = 0.05$), the partial η^2 effect size was found between 0.62 and 0.86. The SPM test was applied to identify the differences between groups in the movement patterns in individual joints and changes in the acceleration of the playing hand. The SPM was calculated using SPM1D in a Python package

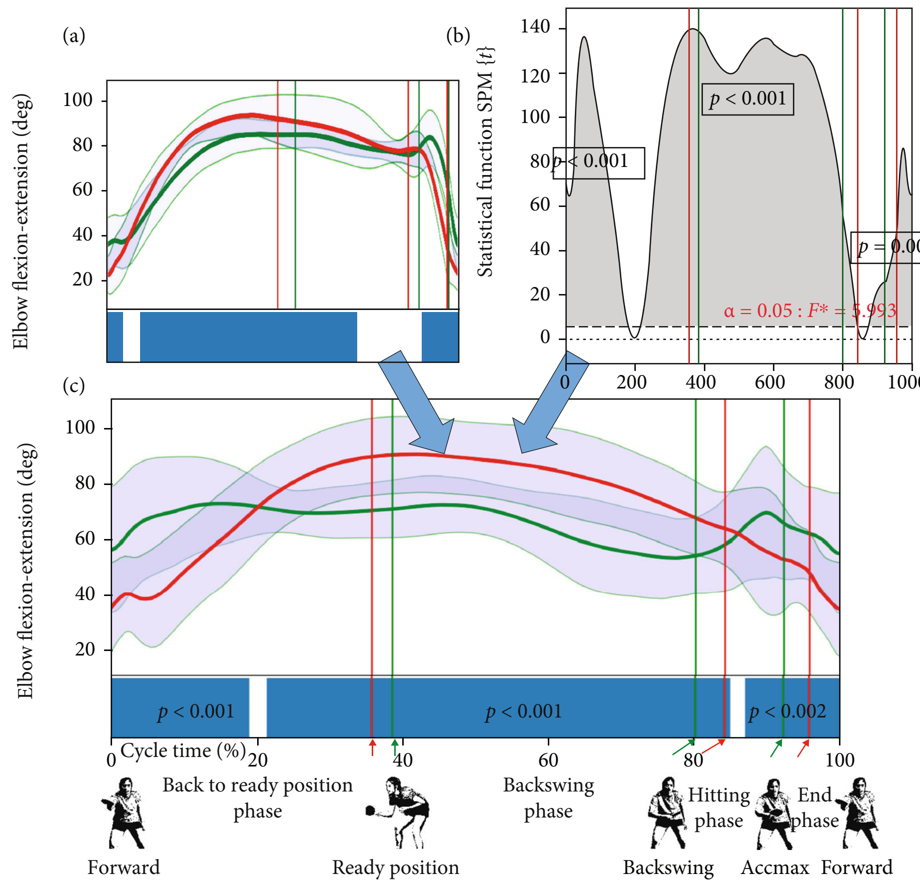

Figure 2: SPM procedure. The SPM, like other statistical methods, has assumptions. The assumptions for the SPM{t} paired sample t-test include continuous waveforms with an equal sample rate and a number of data points; the sample size (or data set size) should be greater than 5 in each group; each waveform should come from a random sample and be normally distributed over time; the waveforms of interest should be spread similarly between the two groups (homogeneity of variance that is maintained over time).

that offers a high-level interface to SPM1D. Angle-time numerical series were averaged over trials and reported against cycle time (Figure 2(a)). For each participant and selected time-dependent angular numerical data, a two-sample t-test SPM{t} function (with alpha = 0.05, non-sphericity correction, and assumption of unequal variances) was numerically computed to check the level of similarity between the movements [19, 20]. For each test, a statistical parametric map SPM{t} (Figure 2(b)) was created by calculating the conventional univariate t-statistic at each point of the gait curve [21–24]. When an SPM{t} crossed the assumed threshold, an additional threshold cluster was created, indicating a significant difference (a grey area) between two compared joint motion patterns in a specific location of the gait cycle. In the present study, because of the high number of statistical analyses, the SPM results are visualized in a summarised manner. Instead of SPM{t} curves, blue bars are shown, indicating the significance during the cycle (Figure 2(c)).

3. Results and Discussion

The study is aimed at evaluating the differences in movement kinematics using the SPM method between two different groups of female table tennis players. The application of the SPM test allowed for the identification of the differences between groups in the movement patterns in individual joints and changes in the acceleration of the playing hand. The basic difference that can be noticed is the time of occurrence of the beginnings and ends of the individual movement phases. For the POL players, the backswing phase starts slightly earlier (about 46% of the cycle duration for POL, 54% for CHIN players) similarly to the hitting phase (83% and 87%, respectively), whereas the average time of the maximum hand acceleration (Accmax) is very similar for both groups (about 96% of the cycle duration). The observation and description of the way of coordinating the movements when hitting the backhand topspin reveals that the average movement pattern (changes in joint angles throughout the cycle) is consistent with that described in previous studies [25, 26]. The following movements were observed in the backswing phase: lower limb flexion, upper body flexion (forward bend), adduction and internal rotation in the shoulder joint, elbow joint flexion, and flexion, pronation, and palmar flexion in the wrist joint. In the hitting phase (with different time of inclusion of individual segments into the movement, according to the principle of the proximal-to-distal movement sequence), the following movements were observed: extension in the lower limb and upper body joints, abduction, flexion, and external rotation in the shoulder joint, extension and supination in the elbow joint, and extension, supination, and radial abduction in the wrist joint.

The analysis of the SPM test results allowed for the observation of the differences in the movement patterns in the individual analyzed joints.

(1) Ankle joints: the movement pattern in the ankle joints is characterized by the occurrence of many periods that differ between the two groups studied. The lack of differences in the flexion-extension movement (dorsiflexion, Figure 3) in the nonplaying side ankle joint (i.e., throughout the hitting phase) and wave-like changes in ankle joint movement observed with higher frequency in CHIN female athletes (Figure 3) are noticeable

(2) Knee joints: in the flexion-extension movement of the knee joints, the wave-like character of the changes in the back to ready position phase and the backswing phase observed in CHIN is noteworthy. Significantly, more periods differing between the two groups occur in the right knee joint (Figure 3), in which the average flexion range is larger in POL compared to the CHIN group during the entire cycle

(3) Hip joints: in hip joint movements, there are more periods of differences concerning the right hip joint. CHIN players exhibit greater abduction and external rotation throughout the cycle in the right hip joint. It is noteworthy that there were no differences between the groups in the significant part of the backswing phase in the abduction movement in the nonplaying side hip joint and the part of the backswing and hitting phases in the rotation movement in these joints (Figure 3).

(4) Joints of the upper body: very few differences were observed in the flexion-extension movements in the lumbar region, in which flexion can be observed in the backswing phase and extension was found in the hitting phase (Figure 4). The range of rotation movement was slight (about 5 deg), more pronounced in CHIN, whereas in the POL group, it was characterized by high variability (high SD value throughout the cycle). The movement of the upper body (thoracic region) differentiates the two groups the most in the sagittal plane (flexion-extension). In CHIN players, this movement is used to a greater extent (about 30-40 deg), from slow flexion in the backswing phase, through faster flexion in the initial hitting phase, to the extension in the Accmax region and later (Figure 4). The rotation of this part of the upper body and lateral flexion in the backswing phase and most of the hitting phase does not show differences between the two groups. These movements take place in small ranges of several degrees

(5) Shoulder joint of the playing limb: in the shoulder joint of the playing limb, it can be observed that the differences mainly concern the back to ready position phase in all planes (Figure 5). In the flexion-extension movement, differences also occur at the end of the forward phase. Greater abduction and external rotation can be also observed in the part of the backswing and hitting phases in the discussed joints in the CHIN female players (Figure 5). It should also be emphasized that there is a period with no differences in the flexion-extension movement in a significant part of the backswing and hitting phases (up to the moment of reaching the maximum acceleration—Accmax)

(6) Elbow joint of the playing limb: the SPM test revealed differences in flexion-extension movement at the elbow joints in the major part of the back to ready position phase, part of the backswing phase, and the end of the hitting phase (Figure 5). Nevertheless, both groups showed elbow flexion in the backswing joint in the back to ready position phase (up to circa 70-90 deg), maintaining this flexion or very slow extension during the backswing phase, and quite a rapid extension during the hitting phase (up to circa 20-40 deg)

(7) Wrist joint of the playing limb: the fewest periods of differences between the two groups demonstrated by the SPM test occur in the movement of elbow flexion and radial abduction in the wrist joint (Figure 5). Maintaining the elbow flexion up to circa -20 to -30 deg can be observed in both groups in the back to ready position and backswing phases, and then, after the beginning of the hitting phase, quite a rapid movement towards radial flexion (up to circa -10-0 deg) was found. The maximum of radial flexion occurs at around Accmax, and there is a brief moment of differences between the groups during this period. The supination-pronation movement in the described joint differentiates between the two groups more. A period of no differences between the groups occurs in the back to ready position phase (from circa 5% to circa 30% of the cycle time) and in circa 91-93% of the cycle time in the hitting phase. Polish female players are characterized by using a greater range of this movement. The supination movement is rapid during the hitting phase, from the moment after the beginning of this phase to the moment of Accmax in both groups. In the extension-flexion movement in the wrist joint, it is noticeable that there are no differences in the back to ready position phase and before the Accmax moment. There is a slow flexion of the limb in the described joint in both groups during the back to ready position and backswing phases, accelerating during the hitting phase. At circa 90% of the cycle, the direction of movement changes to the extension (within circa 10 deg in both groups) at a high rate until reaching Accmax. The latter short period shows no differences between the groups

The observation that comes to mind is the occurrence of the longest periods of differentiation between the groups studied in the lower limb joints, which indicates their

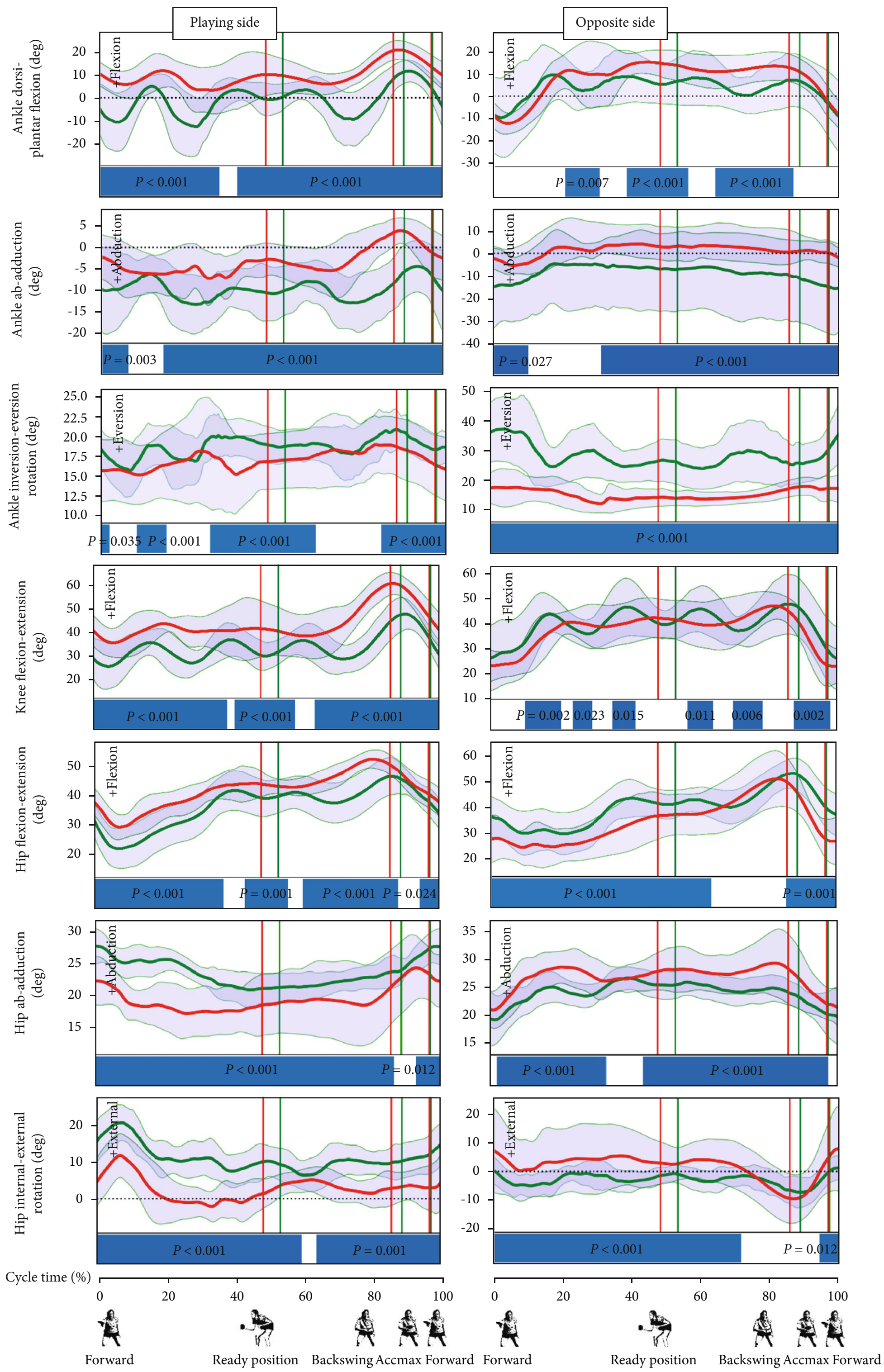

Figure 3: Lower extremity kinematics. Red line: average values of POL; green line: average values of CHIN; grey areas: SD values. Blue bars indicate the significance during the cycle.

different use by both groups of female players. Undoubtedly, a wave-like movement in the ankle and knee joints is more pronounced in CHIN players, which reflects the use of the so-called small steps, mainly in the back to ready position and backswing phases. These steps are used to adapt to the next stroke and keep the lower limbs in constant readiness. Therefore, it can be concluded that CHIN players use these steps more often than POL and perhaps this is due to differences in coaching. Differences can be observed in the ankle joints in all planes, and they affect the entire backswing and

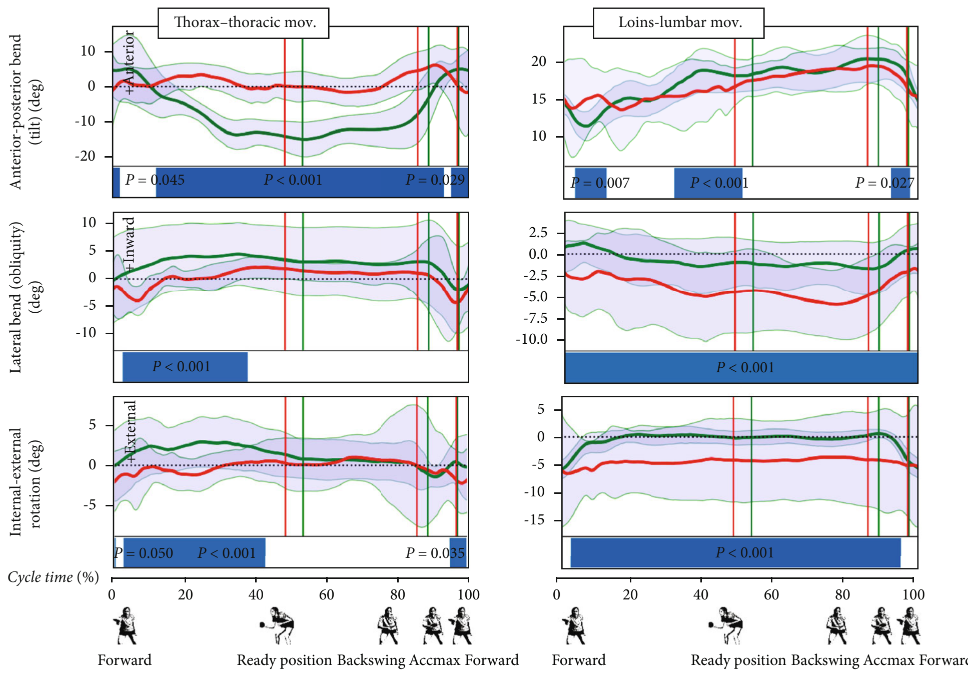

Figure 4: Torso kinematics. Red line: average values of POL; green line: average values of CHIN; grey areas: SD values. Blue bars indicate the significance during the cycle.

forward phases. It is noticeable that the directions of movement in the hitting phase are the same in both groups in the ankle joints, and the differences are in the degree values. The nonplaying side ankle joint in both groups in the forward phase shows no differences and the toe-raise movement (decreasing dorsiflexion, transitioning to plantar flexion), in an approximately 20-degree range. A similar movement, but differentiating between the two groups, can be observed in the right ankle joint. For both joints, the range of motion is smaller in CHIN player. The direction of this movement in the forward phase indicates the use of upward and forward transfer of the center of gravity as an action to support the hitting movement performed by the player. The importance of this movement while performing a stroke has been highlighted in the literature [26, 27]. Wang et al. also pointed out the differences between players at different sport skill levels in the performance of movements in the joints of the lower limbs, emphasizing that these movements can be used better by an economical work with simultaneous use of the energy generated by the elastic components of the joints and muscles (based on the stretch-shortening principle) [28]. Perhaps the differences in the movement in the ankle joints shown in this paper are related to this method. As mentioned above, a wave-like movement in CHIN players was reported in flexion-extension movements in the knee joints, indicating the use of small steps in the preparation phases (back to ready position, backswing). A greater flexion angle in the right knee joint was also observed in the POL group throughout the cycle. This is probably due to the transfer of center of gravity to the right leg, emphasized more in the POL group throughout the cycle. It can be assumed that this difference allows the CHIN players to switch to forehand play faster and more flexibly after performing a pivot and is probably due to the different playing styles prevalent in the two groups. In all players, the forward phase is accompanied by the extension of the knee joints within a range of several dozen degrees. The above findings provide helpful information for coaches and players with regard to the backhand topspin technique and its modifications regarding lower limb movements.

The movement in the hip joint showed long periods of differences between the groups studied. However, similar movement directions were found in individual phases in both groups. The small rotation range of a few degrees in the hip joints should be emphasized, which, according to many authors, greatly helps generate the stroke force and high racket speed in table tennis [26–29]. It is directly suggested that the range of this movement and its use differentiates between players of different sports skill levels. The lower use of rotation in these joints is related to the type of stroke analyzed in this study. It is a topspin backhand played early against a topspin ball, so it is a counterstroke from the group of strokes that utilize the energy of the flying ball and therefore does not require the involvement of great strength of the player. Similar aspects were pointed out by Marsan et al., who evaluated the mechanical energy generated from the hip joint during different variations of strokes, finding that backhand drive required the lowest hip mechanical work [30].

In the lumbar spine, the least differences were found in the flexion-extension motion. In the backswing phase, this is a few degrees of flexion, whereas in the hitting phase-extension in both groups. The lateral flexion movement indicates that the POL players are slightly leaning to the right, with the body weight shifted to the right lower limb, again indicating a more backhanded position than in the Chinese players. The CHIN players seem to stand more universally,

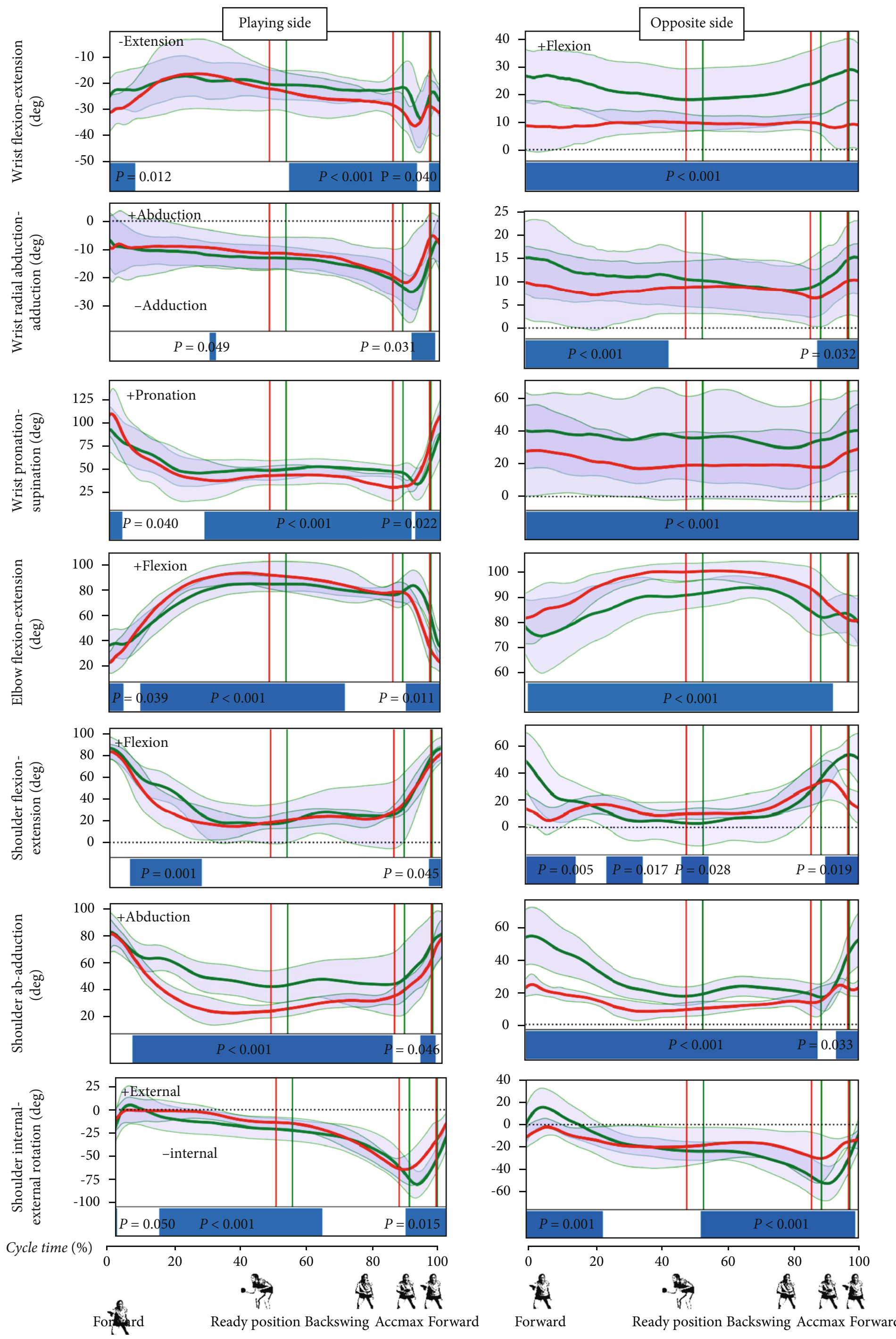

FIGURE 5: Upper extremity kinematics. Red line: average values of POL; green line: average values of CHIN; grey areas: SD values. Blue bars indicate the significance during the cycle.

with the ability to transition more easily from the backhand to the forehand playing, as discussed above. The CHIN players also use a certain amount of rotation in the lumbar region during the hitting phase in contrast to POL players, who hardly use any rotation in this body segment. It must be admitted, however, that the SD values in the POL group are high, indicating great variation in the way this segment is used in the topspin backhand stroke. Nevertheless, the small range of rotation (similar in both groups) in body trunk confirms previous observations concerning the small contribution of hip and trunk rotation resulting from the type of stroke assessed.

Regarding the playing upper hand, the most differences were found in the abduction-adduction of the shoulder, flexion-extension at the elbow joint, and supination-pronation at the wrist joint. In these three cases, the differences between the groups concern much of the back to ready position phase, the beginning of the backswing, and the end of the forward phase. Actually, the end of the forward phase (from Accmax to the end of this phase) differentiates between the groups in each movement in the joints of the playing upper limb. It must be admitted, however, that the directions of movements are very similar (the curves of the graphs have a very similar shape), and the differences demonstrated in the SPM test may be due to the different times beginning the individual phases in the groups. The SPM test showed no differences in flexion-extension and external-internal rotation in the shoulder joint, in radial abduction-adduction, and flexion-extension at the wrist joint during the second part of the backswing and the beginning of the hitting phase. Movement coordination in the female players studied is consistent with that reported in the literature [25, 29]. Furthermore, the description of basic movement, presented in our work, can provide more clarity in understanding the topspin backhand technique.

The values of hand acceleration and its changes over time demonstrated in the SPM test differentiate between the groups studied for most of the cycle and in all phases, with short exceptions of ca. 20% and 40%, and in the hitting phase, especially after reaching Accmax (Figure 6).

For most of the back to ready position phase and the backswing phase, the acceleration values are close to 0. After circa half of the backswing phase, acceleration values increase until they reach maximum values at the end of the forward phase, which are very similar in both groups (about 90 m/s^2). The pattern of acceleration values is then interesting. It is different for both groups in each phase, but it is similar at the Accmax point, and the maximum values obtained by both groups are also similar. Therefore, it can be concluded that despite the indicated differences in movement patterns in both groups, the same value of Accmax was achieved. This may be a manifestation of the phenomenon of equifinality and compensation, indicated in the literature as typical of dynamic systems and variability of movement [5, 10, 31]. Obviously, it should be added that just achieving the right amount of hand acceleration does not determine the accuracy of the play; the hitting angle, the direction of movement, and other factors are also important [32].

3.1. Limitations of the Study. Undoubtedly, from the standpoint of statistical calculations, the number of participants may seem to be a limitation of the study. It should be remembered, however, that the averaging of movement patterns (changes in joint angles over time) can lead to unavoidable errors in the observation of the activity of the human movement system, in which variability, differentiation, and compensation are normal and commonly occurring phenomena [5]. It should also be noted that the observations presented in this study concerned only women and one type of stroke; thus, generalization of the results should be made with caution.

4. Conclusions

The examinations carried out in this study allowed for a detailed description of the technique of performing a fast topspin backhand stroke, thus providing valuable information for table tennis coaches and players. The SPM method allowed for the determination of differences between the Chinese and Polish female athletes. The observed differences include, among others, greater use of the so-called small steps in order to adapt and be ready during the back to ready position and backswing phases, which gives the CHIN players slightly better conditions for preparation for the next plays. The position of the CHIN players compared to that of the POL players favours a quicker transition from the backhand to the forehand play. This difference is probably related to the difference in the dominant playing styles of the groups studied. The differences found are probably mainly due to differences in the training methodologies caused by different coaching systems. It can be also concluded that despite the indicated differences in movement patterns in both groups, the same value of Accmax was achieved. This may be a manifestation of the phenomenon of variability of movement, as well as equifinality and compensation.

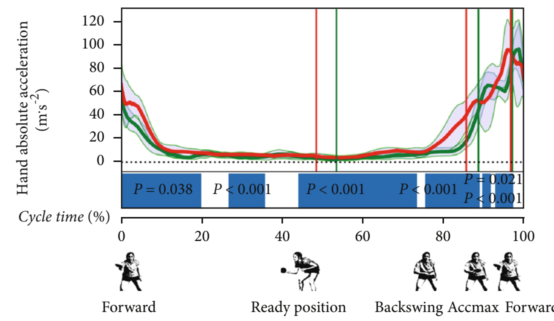

Figure 6: Hand acceleration. Red line: average values of POL; green line: average values of CHIN; grey areas: SD values. Blue bars indicate the significance during the cycle.

Data Availability

The supplementary data (containing angle waveforms and accelerations) used to support the findings of this study are included within the supplementary materials.

Conflicts of Interest

The authors have no conflict of interest to declare.

Supplementary Materials

Angles and accelerations. *(Supplementary Materials)*

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Review Article

Health View to Decrease Negative Effect of High Heels Wearing: A Systemic Review

Meizi Wang,[1,2] Ci Jiang,[1] Gusztáv Fekete,[3] Ee-Chon Teo,[1] and Yaodong Gu[1]

[1]*Faculty of Sports Science, Ningbo University, Ningbo, China*
[2]*Faculty of Engineering, University of Pannonia, Veszprem, Hungary*
[3]*Savaria Institute of Technology, Eötvös Loránd University, Hungary*

Correspondence should be addressed to Ci Jiang; davidmaowl@gmail.com

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Effective recommendations about how to decrease adverse effects of high heels (HH) need to be provided, since wearing HH is inevitable for most women in their daily life, regardless of their negative impacts on the foot morphology. The main purpose of this systematic review was to summarize studies which have provided specific information about how to effectively offset the negative effects of wearing HH, in the case of women, by means of examining heel height, insole, and heel base support (HBS). Some evidence indicate the following: (i) the range of appropriate heel height for HH shoes is 3.76 cm to 4.47 cm; (ii) compared to small HBS, the larger ones effectively increase gait stability, reduce risk of ankle injury, and improve comfort rating during HH walking; and (iii) the use of a total contact insert (TCI) significantly decreases plantar pressure and the impact on the foot, resulting in higher perceived comfort. It must be noted that these results are based on short-term research; therefore, any conclusions with regard to effects in the long term should be taken with a grain of salt. Nevertheless, future studies should be aimed at combining numerical and experimental methods, in order to provide personal recommendations for HH shoes by considering heel height and HBS size, based on the individual characters (weight, height, and age).

1. Introduction

The potential impact of HH shoes on women's health for over 50 years has been of concern in medical circles. Studies have shown that wearing HH can lead to slower self-selected walking speed, shorter step length, and smaller stance phase duration, while it increases ankle plantar flexion, knee plantar flexion, anterior pelvic tilt, and trunk extension [1–7]. Redistributing the plantar pressure, higher ground reaction forces (GRF), larger loading rate, higher peak knee external adduction moments, and higher peak patellofemoral joint stress have been detected during walking in HH [8–11]. It is worthy to note that substantial bodily adjustments have been observed due to wearing HH, e.g., change in the neuromuscular activation pattern, shortening of the gastrocnemius muscle fascicle muscles, increase in the Achilles tendon stiffness, and higher muscle activity of the soleus, tibialis anterior, and medial gastrocnemius [12–14]. These above-mentioned disturbances have been identified as negative implications for the human body. It is presumed that they contribute to several pathologies including metatarsalgia, hallux valgus, Achilles tendon tightness, knee osteoarthritis (OA), plantar fasciitis, and lower back pain, not to mention the elevated instability and imbalance, which can result in a greater risk of falling and slipping [15–19].

Despite widespread warnings from public health institutions and international medical societies [20], there is still a large proportion of the population wearing HH in their daily life. Regarding why women choose to wear HH, Broega et al. surveyed 574 females, between the ages of 24 and 45, who indicated that beauty and femininity were the key drivers of women's behavior [21]. Therefore, accurate suggestions must be provided about how to counter the adverse effects of HH using, instead of only giving a simple advice on not wearing it. Consequently, in the near future, the design of HH shoes must be associated to comfort and aesthetics in order to meet the requirements of beauty and health.

Up to now, researchers have made significant efforts to improve comfort of high-heeled shoes by suggesting a suitable heel height, an appropriate insert insole, sufficient support area of the heel, and even walking speed during HH gait. Studies have shown that an optimal range of height for maintaining postural balance and stability is between 3 and 5 cm [2, 17, 18]. Yung and Wei observed that a TCI, coupled with a metatarsal pad, arch support, and heel-cup mechanism, redistributed the plantar pressure, and as a consequence, it decreased the impact force by 33.2% in the case of HH [22]. It was also considered that small HBS increased the deviation of the center of pressure (COP), which on the one hand caused larger foot pressure in the rearfoot region, and on the other hand, it disturbed the muscle activity pattern [22].

However, the effects of physiology and ergonomics on HH design in terms of heel height, contact insole, and HBS have not yet been summarized. It is essentially needed to reach a consensus for shoe manufacturers and users on what kind of high-heeled shoes or insole is most optimal for women. Therefore, this systematic review is aimed at concluding studies that have provided a specific way to effectively offset the negative effects of wearing HH in the case of women. Our investigations included heel height, insole, and HBS as parameters.

2. Materials and Methods

2.1. Design Data Sources, Search Strategy, and Study Selection. This systemic review was carried out in accordance with the PRISMA statement [23]. To identify relevant papers, a bibliographical search was conducted in four databases: Web of Science, PubMed, Scopus, and Embase. A manual search was performed in OpenGrey literature in April 2020. In some cases, YDG was responsible for contacting author by e-mail to obtain supplement information. The detailed electronic search was as follows: “high heels”, “high-heeled shoes”, “women’s footwear”, “heel height”, “biomechanics”, “comfort height”, “heel base size”, “kinematic parameters”, “kinetic parameters”, “insole measurements”, and “insert”. These keywords are combined and searched on each database. The first and second author (MWZ and CJ) independently performed relevance article screening, which involved the title, abstract, full-text, and data extraction examination.

2.2. Eligibility Criteria. The eligibility of selecting papers was estimated according to the following inclusion criteria: (1) the articles had to focus primarily on healthy women wearing HH shoes, (2) the articles were published in English, (3) full-text, peer-reviewed, original scientific articles published in journals, (4) the presented data is associated with HH gait (including spatiotemporal, kinematic, kinetic parameters, and EMG), (5) the articles focused on how to alleviate harmful influences on female health with HH, and (6) the articles had to be retrievable. If the abstract did not present sufficient details for any of the eligibility criteria, the reviewers would browse the full text. Then, if the full text failed to comply with any of the eligibility criteria, it would be deleted.

2.3. Data Extraction and Quality Assessment. The important details of the selected articles were extracted by two independent reviewers (ECT and GF). The following data were retrieved from the selected articles: author, year of publication, participant characteristics, shoe condition, measured variables, purpose, and main result. In case of disagreement in data extraction, another reviewer (CJ) was included into the discussion to reach a consensus.

The principles of McMaster Critical Review Form were conducted to thoroughly estimate methodological quality of all selected studies [24–26]. This review form provided 15 separate elements to assess the various types of experimental studies. A 2-point scoring system has been established, where the rating was defined as follows: “yes” (1 point), “no or not measured,” or “not applicable” (0 point). This system can be utilized to appraise whether a study meets the standards for good methodological quality [26].

3. Results

3.1. Search Results and Validity. The bibliographical database search identified 906 citations: 276 in PubMed, 243 in Scopus, 187 in Web of Science, and 200 in Embase. Duplicates were deleted leaving a total of 362 articles for evaluation. 276 studies were eliminated since after scanning the titles and abstracts of the retrieved papers, it turned out that the content was inconsistent with the standards. 86 full-text studies were extracted for detailed review, and 78 studies were removed as these failed to meet inclusion criteria. A total of 8 studies were eventually eligible for all inclusion criteria. The detailed search strategy is shown in Figure 1, while the basic information of the selected articles is listed in Table 1. Quality evaluation of each article by the McMaster score form is presented in Table 2. All of the extracted papers were graded from moderate to high rating based on the McMaster critical appraisal tool.

3.2. Overview of the Included Studies. An accurate recommendation for offsetting negative impact on HH for women is to alter three important parameters, namely, heel height, HBS, and sole insert. The biomechanical investigation of these parameters commonly involve kinematic, kinetic, and perceived stability changes of the lower extremity, such as plantar pressure in a different region of the foot, COP deviation in a gait cycle, spatiotemporal variation, and comfort rating. One included article contained EMG testing to detect muscle activity in the tibialis anterior (TA), medial gastrocnemius (MG), quadriceps (QUA), hamstrings (HAM), and erector spinae (ES) during walking, and one study recorded heart rate as a physiological variable.

3.2.1. Heel Height. Two studies conducted several experiments to determine an appropriate height heel of high-heeled shoes in order to reduce disturbance of the locomotor pattern. Based on three different walking speeds, Nadège et al. assessed the effect of nine pairs of heel height (0 cm, 2 cm, 3 cm, 4 cm, 5 cm, 6 cm, 7 cm, 8 cm, 9 cm) on kinematic

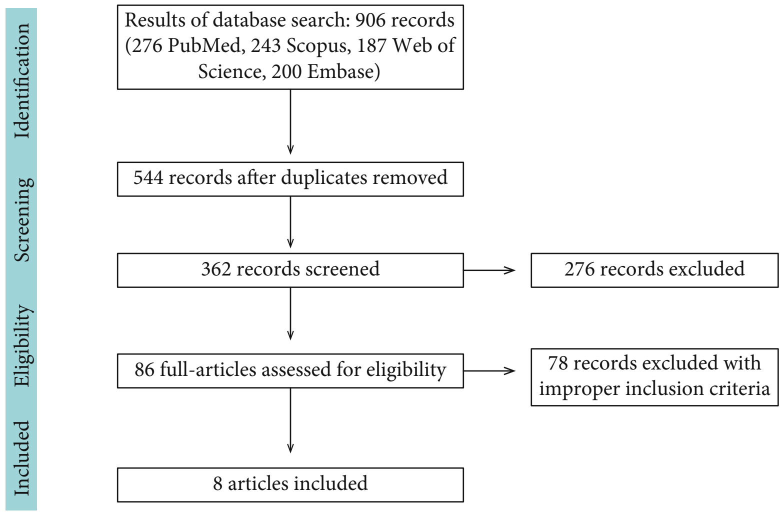

Figure 1: Flowchart of the search strategy.

Table 1: Basic information on selected articles.

Number [ref.]	Author/date	Title	Journal	Concentration
1 [27]	Nadège et al. 2015	Wearing high-heeled shoes during gait: kinematics impact and determination of comfort height	American Journal of Life Sciences	Heel height
2 [28]	Ko and Lee 2013	The changes of COP and foot pressure after one-hour walking wearing high-heeled and flat shoes	Journal of Physical Therapy Science	Heel height
3 [30]	Luximon et al. 2015	Effects of heel base size, walking speed, and slope angle on center of pressure trajectory and plantar pressure when wearing high-heeled shoes	Human Movement Science	HBS
4 [29]	Guo et al. 2012	Effect on plantar pressure distribution with wearing different base size of high-heel shoes during walking and slow running	Journal of Mechanics in Medicine and Biology	NBS
5 [31]	Hong et al. 2013	Effect of shoe heel height and total-contact insert on muscle loading and foot stability while walking	Foot and Ankle Society	Insert insole
6 [7]	Li et al. 2010	Biomechanical effects of foam inserts on forefoot load during the high-heeled gait: a pilot study	Journal of Mechanics in Medicine and Biology	Insert insole
7 [8]	Hong et al. 2005	Influence of heel height and shoe insert on comfort perception and biomechanical performance of young female adults during walking	Foot and Ankle International	Insert insole
8 [22]	Yung and Wei 2005	Effects of shoe inserts and heel height on foot pressure, impact force, and perceived comfort during walking	Applied Ergonomics	Insert insole

parameters, in which the stride length (SL), swing phase (DSwp), duration of the stance phase (DStp), swing phase (DSwP), and gait ratio were included, as well as heart rate [27]. The results indicated that the most comfortable heel height is 4.13 ± 0.34 cm, which is accompanied with less disruptive locomotor pattern and optimal heart rate, compared to other heights. Differently, Ko and Lee determined the most comfortable heel height for HH shoes by detecting the displacement of the COP and plantar pressure change after walking for 1 hour in 0.5 cm, 4 cm, and 9 cm shoes, respectively [28]. Results presented that 4 cm heel height is the most suitable, since this height is accompanied with stable COP tendency and less plantar pressure than walking in 0.5 cm or 9 cm shoes. Details are presented in Table 3.

3.2.2. HBS. Two studies were associated with the effect of HBS on distribution of plantar pressure patterns, COP trajectory, and perceived comfort. A large HBS demonstrated

Table 2: Methodological quality of included studies by using the McMaster critical appraisal form.

Number	Study design	Level	Items															Score
			1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	
1	CCT	III-2	√	√	√	√	√	√	x	√	n/a	√	√	√	√	x	√	12/14
2	CCT	III-2	√	√	√	√	√	x	√	√	n/a	√	√	√	√	x	√	12/14
3	CCT	III-2	√	√	√	√	√	x	√	√	n/a	√	√	√	√	√	√	13/14
4	CCT	III-2	√	√	√	√	√	x	√	√	n/a	√	√	√	√	x	√	12/14
5	CCT	III-2	√	√	√	√	√	x	√	√	n/a	√	√	√	√	x	√	12/14
6	CCT	III-2	√	√	√	√	x	x	√	√	n/a	√	√	√	√	√	√	12/14
7	CCT	III-2	√	√	√	√	√	x	√	√	n/a	√	√	√	√	x	√	12/14
8	CCT	III-2	√	√	√	√	√	x	√	√	n/a	√	√	√	√	x	√	12/14

Level of evidence (based on NHMRC hierarchy); CCT: control clinical trial; FU/RCT: follow-up study from randomized control trial. √: yes; x: no/not reported; n/a: not applicable. McMaster Items: (1) study purpose clearly stated; (2) background literature reviewed; (3) appropriate research design; (4) sample described in detail; (5) sample size justified; (6) outcome measure reliability reported; (7) outcome measure validity reported; (8) intervention described; (9) contamination avoided; (10) cointervention avoided; (11) results reported in terms of statistical significance; (12) appropriate analysis method; (13) clinical significance reported; (14) dropouts reported; (15) appropriate conclusion.

smaller maximal peak pear pressure in the rearfoot, midfoot, and forefoot compared to small HBS [29, 30]. It must be noted that the scale of HBS affects the COP location in the anterior-posterior direction at the end of the stance phase. The COP deviations are increased with a small HBS when compared to a large HBS [30]. Only one study reported information about the stability as a function of HBS. It can be concluded that a large HBS can lead to a more stable gait during walking with HH [30]. Details are presented in Table 4, and different sizes of HBS are shown in Figure 2(b).

3.2.3. Insert Insole. Four included studies evaluated the effect of insert insole on kinematic, kinetic, EMG, and comfort rating of the lower extremity, but different types of insole were used for each study. One study investigated how subject's rearfoot kinematic, muscle activity, and subjective comfort were affected by TCI which were designed from rearfoot to metatarsal head during walking with HH (Figure 2(b) [31]. When compared with a noninsert condition, results showed that the use of a TCI could reduce peak MG by 19.0% and peak ES by 21.5% in HH with 7.6 cm, and rearfoot inversion angle was significantly decreased. But this study did not present kinetic variation of the foot. Another included article used an insole from the rearfoot to metatarsal head (TCI II) that was designed by the orthotist to fit each participants' foot to determine the effect of shoe inserts on plantar pressure, GRF, and perceived comfort during walking in different heel height shoes (1.0 cm, 5.1 cm, 7.6 cm) (Figure 2(b)) [8]. Results showed that the peak pressure of the medial forefoot reduced by using TCL compared with noninsert shoes, and it was more effective to use TCI in the higher heels than lower and flat heels. Furthermore, Yung and Wei also found that a TCI decreased heel pressure by 25%, medial forefoot pressure by 24%, and impact force by 33% [22].

While the heel cup pad could decrease the heel pressure and impact force and the use of single arch support inserts can attenuate the medial forefoot pressure, no special changes to the metatarsal pad using measured parameters were found [22]. The medial forefoot (MF) has been considered the most sensitive area in response to heel height variation [7, 22, 32, 33]. The effect of four different types of foam insoles (soft 5 mm, soft 10 mm, hard 5 mm, hard 100 mm) in the targeted MF region on plantar pressure was tested. There was a great advantage in soft 5 mm to reduce peak pressure by 26%, impact force by 27% in MF region compared to the noninsert condition [7]. All the above insole types are presented in Figure 2(a). More details on female insert insoles are presents in Table 5.

4. Discussion

To reach a full understanding of how the design factors of high-heeled shoes affect locomotor pattern, disturbance of plantar pressure, and perceived comfort is crucial. This review identified 8 articles, which appraised either the effect of heel height, HBS, or insert insole on lower limb kinematic, kinetic, or EMG during waking with HH, as well as perceived comfort.

4.1. Heel Height Studies. There are only two studies that evaluated the comfort heel height of HH by using different methods, and a consensus has been formed that shows that the appropriate heel height in high-heeled shoes is 4.13 ± 0.34 cm [27, 28]. In addition, this result is also consistent with Ko who reported that the preferable heel height was between 3 cm and 5 cm, but this article as a conference paper failed to be selected in this review [34]. A growing number of researches indicate that musculoskeletal systems are directly modified from wearing HH. On the other hand, the human foot naturally presents a moderate imbalance in body weight (BW) distribution; i.e., 43% of BW is loaded to the foot front, with the remaining 57% at the heel portion when walking barefoot. For that reason, a slight heel height shoe (2 cm) is recommended to be used by orthopedic specialists, since it can balance the distribution of plantar pressure to relieve rearfoot load [21]. But these outcomes depend on short-term testing rather than for a long period. Therefore, the results of a suitable heel height in included articles may not predict the impact of wearing high-heeled shoes in the long term.

Table 3: Summary of articles related to the effect of appropriate heel height among females.

Number	Participant characteristics (mean ± standard deviation)				Shoe condition		Walking speed (cm/s)	Variables measured	Purpose	Main findings
	N	Age (years)	Height (m)	Weight (kg)	HBS (cm^2)	Height of heel (cm)				
1	15	22.40 ± 2.56	1.63 ± 0.04	59.07 ± 5.15	5.3	0, 2, 3, 4, 5, 6, 7, 8, 9	0% Ffcwh +20% Ffcwh -20% Ffcwh	(i) SL (ii) DStP (iii) DSwP (iv) Gait ratio (v) Heart rate	To determine the comfort heel height for the shoe	(i) The comfort heel height was 4.13 cm ± 0.34
2	15	20.90 ± 1.30	1.60 ± 3.30	52.10 ± 5.0	—	0.5, 4, 9	1.17	(i) Foot pressure (ii) COP	To determine the most appropriate height for shoe heels	(i) The distribution of foot pressure and COP did not change significantly in 4 cm heel height after walking (ii) 4 cm heel height was preferable for health and comfort

Note: Ffcwh: step frequency freely chosen in shoes without heel; SL: stride length; DStP: duration of the stance phase; DSwP: swing phase; COP: central of pressure.

Table 4: Summary of articles related to the effect of HBS on female.

Number	Participant characteristics (mean ± standard deviation)				Shoe condition		Walking speed (cm/s)	Variables measured	Purpose	Main findings
	N	Age (years)	Height (m)	Weight (kg)	HBS (cm2)	Height of heel (cm)				
3	15	22.50 ± 4.70	1.61 ± 0.04	51.30 ± 4.90	-0.88, -8.92	3	112, 143	(i) COP (ii) Perceived stability (iii) Plantar pressure	Evaluated the effects of the HBS of HH shoes	(i) Smaller COP deviations in larger HBS (ii) The walking speed mainly affected the locations of the COP in the anteroposterior direction during gait (iii) A maximal peak pressure increased over the forefoot, midfoot, and rearfoot in larger HBS (iv) The participants felt more stable in larger HBS
4	13	22.0 ± 0.8	1.60 ± 3.30	52.10 ± 5.0	-1.44, -7.7	7.8	100, 115, 200	(i) Plantar pressure	Be better to select one with a wide-based heel	(i) Smaller plantar pressure in medial, central, and lateral of forefoot and toe regions in larger HBS

Note: COP: central of pressure; HBS: heel base support; HH: high heels.

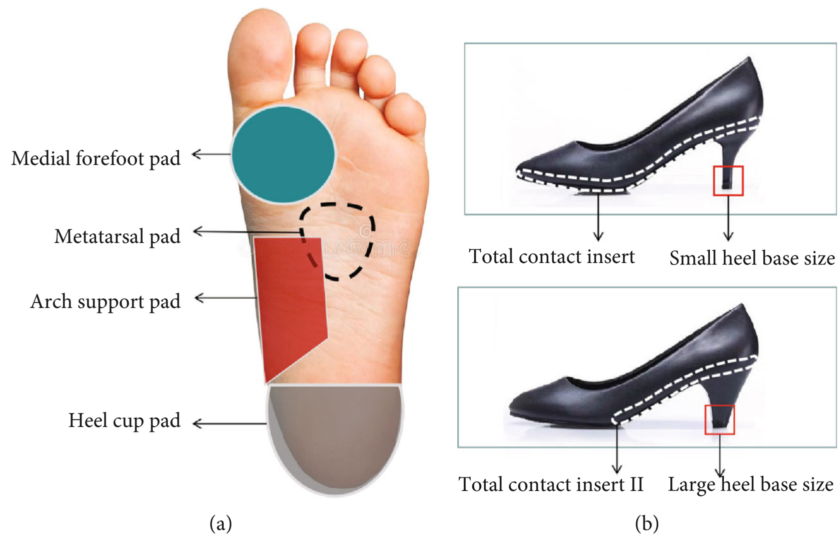

Figure 2: (a) The different insert pads mentioned in included studies. (b) The total contact insert, total contact insert II (from rearfoot to metatarsal head), and small and large heel base size mentioned in included studies. Note: (a) and (b) do not represent the actual ones used in the included study.

4.2. HBS Studies. The scale of HBS is another important factor influencing locomotor pattern during gait with HH. The narrow heels are the most commonly used design in HH which increase plantar pressure, especially in the heel region and lead to instability [22, 35, 36]. Luximon et al. noted that the maximal peak pressure uniformly increased over the whole plantar in large HBS, whereas a narrowed HBS presented higher maximal peak pressure in the toe region [30]. This result is partially similar to Guo et al. whose research showed that the plantar pressure of hallux was significantly increased in small HBS compared to large HBS [29]. Except for the impact of heel height, the narrow HBS may be a direct reason contributing to hammer toe which is caused by excessive pressure on the metatarsal-phalangeal region during walking with HH. Additionally, a smaller HBS presented a larger COP deviation which triggered gait instability, where the toes had to grip the sole of shoes to keep stable. This scenario may be another reason that could lead to the development of a hammer toe when wearing HH [30].

It must be noted that only one heel height was used to measure the function of using different HBS in two selected studies. Although previous researchers suggested that a decreased HBS rather than an increased heel height was the main element for reducing stability during walking with HH [37], the different sizes of HBS combined with diverse heel height should be assessed in the future to further confirm the effectiveness of HBS on gait stabilization.

4.3. Insert Insole Studies. Insert insole has been widely used in footwear to improve perceived comfort, absorbing energy attenuating impact forces, redistribute the plantar pressure, and reduce the risk of movement-related injury [38–40]. The various insert designs presented different kinetic modifications during walking with HH. For instance, Yung and Wei indicated that a heel cup pad reduced pressure by 24.3% and impact force by 18.6% in the heel region when wearing HH [22]. An arch support insole decreased peak pressure by 15% in the medial forefoot region and raised the pressure by 125.6% in the midfoot region, since it was used to prevent depression of the longitudinal arch during weight bearing, thereby alleviating the tension of the plantar aponeurosis [22, 41, 42]. Weight bearing can be transferred from the forefoot to the longitudinal and metatarsal arches by the metatarsal pad; however, no changes in pressure and impact forces were found in the medial forefoot region reported by Yung and Wei [22]. Furthermore, medial forefoot pads with different foam hardness and thickness were utilized; and the results showed that the thick soft pad can effectively reduce larger pressure and impact force caused by HH in the medial forefoot when compared to other types [7, 43].

In terms of using TCI, three included studies indicated that TCI relieved pressure and impact force on multiple areas of the foot simultaneously and significantly improved perceived comfort during walking with HH [8, 22, 31]. The TCI provides a highly conforming fit between the foot and the contact surface of the insole, as well as spreading and redistributing pressure over the rearfoot, midfoot, and forefoot. Notably, the current research notes that the use of TCI is the most effective way to attenuate pressure in comparison to other single pads during walking with HH. Further studies are needed to evaluate the effect of different thicknesses and material properties of TCI on load and pressure redistribution during walking with HH shoes. What is more, the effectiveness of insoles also needs to be estimated in the long term to determine whether this type of intervention should be recommended for women with high heels-related foot problems.

4.4. Limitations and Future Direction. The most obvious limitation in this review is the small sample size. Only 8 studies met our inclusion criteria hence the reason why a meta-analysis was not conducted. In addition, the effects of

Table 5: Summary of articles related to the effect of insert insole on female.

Number	Participant characteristics (mean ± standard deviation)				Shoe condition	Type of insole	Variables measured	Purpose	Main findings
	N	Age (years)	Height (m)	Weight (kg)	Height of heel (cm)				
5	15	24.5 ± 4.5	159.3 ± 6.5	49.6 ± 6.4	1.0, 5.1, 7.6	(i) TCI II (from rearfoot to metatarsal head)	(i) Kinematic (ii) EMG (iii) Comfort rating	To investigate how shoe heel height and use of TCI in high-heeled shoes affect the wearer's rearfoot complex, muscle loading, and subjective comfort	(i) The use of TCI reduced the rearfoot inversion angle and in both QUA and ES muscles activity (ii) Comfort rating increased by using TCI
6	8	22.0 ± 2.0	163.0 ± 3.0	51.0 ± 4.5	0, 4.5, 8.5	(i) MF foams insole with soft (5 mm), (ii) MF soft 10 mm (iii) MF hard 5 mm (iv) MF soft 10 mm	(i) Plantar pressure	To determine the effect of foam inserts in targeted medial foot region on the foot pressure distribution during normal walking	(i) Thicker soft foam sole significantly reduced peak pressure in MF than thinner hard foams (ii) The maximum force and peak pressure reduced in the MF region bu using foams than no insole in HH
7	20	25.4 ± 3.8	157.8 ± 5.0	50.5 ± 4.2	1.0, 5.1, 7.6	(i) TCI insole	(i) Perceived stability (ii) Plantar pressure (iii) GRF	To evaluate the effect of TCI on comfort perception and plantar pressure	(i) The comfort ratings were significantly increased when wearing with TCI (ii) Use of the TCI reduced the peak pressure in the medial forefoot (iii) Mean GRF decreased to 39% BW in the highest shoes with the TCI (iv) It was more effective to use TCL in higher heels than in the lower heels and in flat heels
8	10	23 ± 3.4	160 ± 3	50 ± 3	1.0, 5.1, 7.6	(i) Heel cup pad (ii) Arch support (iii) Metatarsal pad (iv) TCI	(i) Plantar pressure (ii) Comfort rating	Whether use of shoe inserts change foot pressure distribution, impact force, and perceived comfort during walking	(i) Heel cup insert reduced the heel pressure and impact force (ii) Arch support insert reduced the medial forefoot pressure and improved comfort (iii) TCI reduce heel pressure, medial forefoot pressure, and impact force and provided a higher perceived comfort

Note: TCI: total contact insert; MF: medial forefoot; GRF: grand reaction force; BW: body weight; QUA: quadriceps; ES: erector spinae; EMG: electromyography.

walking/running speed on locomotor pattern during high-heeled gait were not studied because there are a wide range of variables and different experimental conditions that would need to be taken into consideration.

It is worth thinking about HH in relation to finite element model analysis and laboratory tests to determine what kind of material properties, hardness, and thickness of insert insole are optimal to minimize the negative impacts of wearing HH. On the other hand, the age, height, and body mass are important parameters in wearing HH; the age affects muscle strength, the height may affect joint moment, and bodyweight directly related to load increase. Future studies should be aimed at providing personal recommendations for HH in terms of choosing the heel height and HBS size based on the individual characteristics that involve weight, height, and age. It seems likely achieved by conducting a comprehensive study that combines the numerical simulation, finite element model analysis, and a large number of sample experiments in the long term.

5. Conclusions

We have systematically reviewed studies focused on factors that aim to counter the adverse impacts on high-heeled shoes. The effects of heel height, heel base size, and insert insole on the biomechanical of lower extremity and perceived comfort are concluded. Some evidence demonstrates that (i) the range of appropriate heel height for wearing HH is 3.76 cm to 4.47 cm; (ii) compared to small heel base size, the larger ones effectively increase gait stability, reduce risk of ankle injury, and improve comfort rating during walking with HH; and (iii) the use of a total contact insert significantly decreases plantar pressure and impact forces on the foot so that a higher perceived comfort is achieved. However, there were some limitations in the data presented in the included articles due to the different methodologies used and a limited number of studies. All the above conclusions need to be further tested in a longer duration experiment. In the future, numerical simulation, finite element model analysis, and a large number of sample experiments should be combined to offer personal recommendations for wearing HH based on the individuals' characteristics.

Conflicts of Interest

The authors declare that they have no conflicts of interest.

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Research Article

Comparison of Modular Control during Side Cutting before and after Fatigue

Naoto Matsunaga,[1,2] Kenta Aoki,[3] and Koji Kaneoka[4]

[1]General Education Core Curriculum Division, Seigakuin University, 1-1 Tosaki, Ageo, Saitama, Japan
[2]Waseda Institute for Sport Sciences, Waseda University, 2-579-15 Mikajima, Tokorozawa, Saitama, Japan
[3]School of Sport Sciences, Waseda University, 2-579-15 Mikajima, Tokorozawa, Saitama, Japan
[4]Faculty of Sport Sciences, Waseda University, 2-579-15 Mikajima, Tokorozawa, Saitama, Japan

Correspondence should be addressed to Naoto Matsunaga; n_matsunaga@seigakuin-univ.ac.jp

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The purpose of this study was to clarify the coordination between the trunk and lower limb muscles during sidestep and to compare this coordination before and after fatigue intervention. The intervention was lateral jump until exhaustion. Nonnegative matrix factorization (NMF) was used to extract muscle synergies from electromyography. Subsequently, to compare the muscle synergies, a scalar product that evaluates the coincidence of synergies was calculated. Three muscle synergies were extracted before and after the intervention from the NMF analysis. In accordance with the evaluation of the scalar product, these synergies were the same before and after the intervention. One of these synergies that engaged the internal oblique/transversus abdominis, rectus femoris, and adductor muscle was activated from before landing to midstance during sidestep motion; therefore, this synergy is thought to suppress excessive hip abduction. However, the activation timing of this synergy was delayed after the intervention ($P = 0.028$, effect size: 0.54, Wilcoxon test). This delay is considered to decrease hip stability. Thus, this change may induce a reduction in hip control function.

1. Introduction

Muscle synergy is defined as a group of muscle activities in synchrony and is extracted using mathematical methods, such as principal component analysis and nonnegative matrix factorization (NMF) [1]. In synergy analysis, electromyography is divided into two factors: (i) muscle synergy, which represents the relative weighting; and (ii) the time-varying component, which represents the relative activation of the muscle synergy [1]. Research on muscle synergy has been widely conducted in the fields of rehabilitation and neuroscience to evaluate activities of daily living, such as reaching tasks and locomotion [2–4]. In recent years, it has expanded to sports science and is expected to improve sports performance and prevent sports injuries [5–13].

Groin pain syndrome is common in soccer and rugby, wherein repetitive sidestep or side cutting motions are frequent [14–17]. It is defined as a dysfunction around the pelvis that results from poor mobility, stability, or coordination between the trunk and lower limbs, causing pain around the inguinal area [18]. Matsunaga et al. reported that the muscle synergies between the trunk and lower limb muscles during side cutting were different with and without groin pain [12]. However, whether the alteration of the synergy occurs because of groin pain or whether groin pain occurs because of the alteration of the synergy is unclear. In addition, groin pain is believed to be caused by overuse [14, 19, 20]. Overuse injury in sports occurs when the musculoskeletal structure receives a repetitive force, producing a combined fatigue effect over a period beyond the capacities of the specific structure [21, 22]. A previous study reported that the synergy changes due to transitory fatigue [11], and we hypothesized that it might be one of the risk factors for overuse injuries. Therefore, the purpose of this study was to compare the synergies during side cutting before and after transitory fatigue intervention. We hypothesized that the delay of the time-varying component will occur because the synergy time-varying component of the subjects with groin pain will be delayed.

2. Materials and Methods

2.1. Subjects. We recruited young healthy men with an exercise routine. Nine men (mean ± SD: age, 20.8 ± 2.2 years; height, 1.74 ± 0.06 m; and weight, 67.3 ± 5.7 kg) participated in this study. They had performed hard training as high school students and had been performing physical activities two to three times per week at the recreational level when they participated in this study. Their sports included soccer, badminton, baseball, and rugby. The exclusion criteria included a history of lower limb disorders that included groin pain syndrome, neurological disorders, or lower limb surgery. This study was approved by the ethics committee of our university (2010–270). All the subjects read and signed an informed consent form prior to participation.

2.2. Test Exercises. The test exercise was 5 repeated sidesteps with a width of 1.1 times the subject's height. The protocol was performed in the following order: first, five repeated sidestep motions at full effort, followed by the fatigue intervention, and finally, 5 repeated sidestep motions at full effort in the fatigue state. The subjects performed lateral jumps as a fatigue intervention. Lateral jumps can impose a large load on many muscles around the hip [23]. The width of the lateral jump was the same as the subject's height. The fatigue intervention required continuous lateral jumping, and fatigue was reached when the subjects could not keep jumping in sync with the metronome (DB-60, BOSS Co., Japan) rhythm at 60 Hz or jumping the distance of their height.

2.3. Data Measurement. To measure muscle activity, a wireless electromyography (EMG) system (EMG-025, Harada Electronic Co., Japan) was used. The activities of the following 8 muscles were measured as described by Matsunaga et al. [11]: rectus abdominal (RA), external oblique (EO), internal oblique/transversus abdominal (IO/TrA), erector spinae (ES), rectus femoris (RF), semitendinosus (ST), gluteus medius (Gmed), and adductor (ADD). All the muscles measured were on the right side. The RA electrodes were placed 3 cm lateral to the umbilicus. The EO electrodes were placed midway between the costal margin of the ribs and iliac crest, approximately 45° from the horizontal plane. The IO/TrA electrodes were placed approximately 2 cm medial and inferior to the anterior superior iliac spine (ASIS), and the ES electrodes were placed 3 cm lateral to the L4 spinal process. The RF electrodes were placed on the belly of the muscle corresponding to the center point between the ASIS and the upper margin of the patella. The ST electrodes were placed on the center point between the ischial tuberosity and medial epicondyle of the femur. The Gmed electrodes were placed 3 fingerbreadths from the lower region of the iliac crest. The ADD electrodes were placed 4 fingerbreadths from the lower region of the pubic symphysis. Before the surface electrodes were attached, the skin was rubbed with a skin abrasive and alcohol to reduce skin impedance to a level <2 kΩ. Pairs of disposable Ag/AgCl surface electrodes (Vitrode F-150S; Nihon Kohden Co., Japan) were attached parallel to the muscle fibers. The sampling frequency was set to 1000 Hz. To normalize the EMG data, a maximum voluntary contraction (MVC) test was performed on each muscle before the test exercise. The MVC for the RA was obtained, while the subjects performed a partial sit-up with their knees flexed and manual resistance was applied. For the EO, the subjects were in a supine position, with their knees and trunk flexed and rotated to the left. Resistance was applied to the shoulders, with the trunk extending and rotating to the right. For the IO/TrA, the trunk was instead flexed and rotated to the right, with the resistance applied at the shoulders, and the trunk extending and rotating to the left. For the ES, the MVC task was trunk extension performed in the prone position with manual resistance applied to the upper thoracic area without leg movement. For the RF, the subject sat on a chair with the hip and knee flexed to 90° and performed knee extension. Resistance was applied at the shank in the knee flexion direction. For the ST, the subjects performed knee flexion in the prone position, with the knee flexed to 45°. Resistance was applied at the shank in the knee extension direction. For the Gmed, the subject was in a side-lying position with the right side up and the hip extended and performed hip abduction with resistance applied in the hip adduction direction. For the ADD, the subject was in the side-lying position with the right side down with the hip extended and performed hip adduction with resistance applied in the hip abduction direction. Manual resistance was gradually increased up to the subject's limit and then held for 3 s.

Three three-dimensional motion capture cameras (Oqus, Qualisys, Sweden) were used to determine the timing of foot contact and push-off. The laboratory (global) orthogonal coordinate system (frame) followed the right-hand rule and had the positive x-direction orientated toward the left, the positive y-direction orientated toward the back, and the positive z-direction orientated vertically upward. To obtain kinematic data, reflective markers (QPM190, Qualisys, Sweden) were attached to 10 landmarks as follows: bilaterally on the edge of the toe on the shoes, the internal and external pastern on the shoes, the anterior superior iliac spinae, and 2 markers were placed on the floor, on either side of the subject. Kinematic data were collected at 200 Hz and synchronized with the EMG system.

2.4. Data Analysis. We analyzed the third round of sidestep motion during the test exercise. We determined the timing of landing using the acceleration of the edge of the toe marker. The EMG data were obtained from 200 ms before landing to 200 ms after push-off based on the kinematic data. A custom MATLAB (MATLAB R2016, MathWorks, Inc., Natick, MA, USA) code was used to analyze the EMG data. The raw data were band-pass filtered between 20 and 450 Hz and full-wave rectified. Then, they were interpolated to 200 time points. The EMG data were normalized relative to the associated MVC data of the muscle. As previously described, NMF was performed to extract modules [1], as follows:

$$\mathrm{E} = \mathrm{WC} + \mathrm{e} \cdots \text{formula 1}$$

$$\min_{\substack{W>0 \\ C>0}} \|\mathrm{E} - \mathrm{WC}\|_{\mathrm{FRO}} \cdots \text{formula 2}, \tag{1}$$

where E is a p-by-n initial matrix (p is the number of muscles and n is the number of time points). The initial matrix comprised normalized EMG data and a cycle for each of the 8 muscles; therefore, E is a matrix with 8 rows and 200 columns. W is a p-by-s matrix (s is the number of synergies) and represents muscle synergy, C is an s-by-n matrix and represents the time-varying component, and e is a p-by-n residual error matrix. Formula 2 indicates that matrix "e" calculated using formula 1 reaches a minimum. W is a vector; therefore, the calculated W is written as $\vec{W}$. For each subject, we iterated the analysis by varying the number of synergies between 1 and 8, and then selected the least number of synergies that accounted for >90% of the variance accounted for (VAF) [5, 24, 25]. Global VAF was calculated based on the findings of previous studies.

$$\text{Global VAF} = \left(1 - \frac{\sum_{i=1}^{p}\sum_{j=1}^{n}\left(e_{i,j}\right)^2}{\sum_{i=1}^{p}\sum_{j=1}^{n}\left(E_{i,j}\right)^2}\right) \times 100\,[\%] \cdots \text{formula 3}, \tag{2}$$

where i goes from 1 to p and j goes from 1 to n. Thus, i increases from 1 to 8, and j increases from 1 to 200 in this study. In addition, to confirm the reliability of our analysis, we also calculated the local VAF based on Hug et al. [24]:

$$\text{Local VAF}\,[m] = \left(1 - \frac{\sum_{j=1}^{n}\left(e_{m,j}\right)^2}{\sum_{j=1}^{n}\left(E_{m,j}\right)^2}\right) \times 100\,[\%] \cdots \text{formula 4}, \tag{3}$$

where m represents the muscle. We defined the adoption standard for a local VAF >75% [24].

To compare W before and after the intervention, the scalar product (SP) was calculated based on the findings reported by Cheung et al. [4]:

$$\mathrm{SP} = \frac{\overrightarrow{W_{\text{before}}} \times \overrightarrow{W_{\text{after}}}}{\left|\overrightarrow{W_{\text{before}}}\right|\left|\overrightarrow{W_{\text{after}}}\right|}\,(0 \leqq SP \leqq 1) \cdots \text{formula 5}. \tag{4}$$

The SP for the use of the correlation coefficients can assess the similarity of W. We defined the module as similar if the SP was above 0.75 [4].

In addition, to evaluate the level of fatigue, wavelet transform was performed based on the findings reported by Smale et al. [8], and the instantaneous mean frequencies (IMF) were calculated.

$$\mathrm{IMF}\,(t) = \frac{\int \omega P(t,\omega)d\omega}{\int P(t,\omega)d\omega} \cdots \text{formula 6}, \tag{5}$$

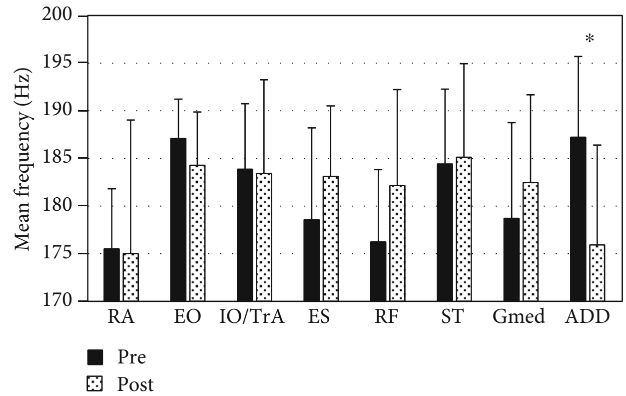

Figure 1: Comparison of the mean frequency before and after fatigue. RA: rectus abdominis; EO: external oblique; IO/TrA: internal oblique/transversus abdominis; ES: erector spinae; RF: rectus femoris; ST: semitendinosus; Gmed: gluteus medius; ADD: adductor muscle. *Significant difference ($P < 0.05$).

where $P(t,\omega)$ is the time-dependent power spectral density. After that, the IMF was averaged to calculate the mean frequency. The analyzed range of data was the same as that of the NMF analysis.

2.5. Statistics Analysis. Following the results of the Kolmogorov-Smirnov test, we used the Wilcoxon test to compare the mean frequency, peak timing of the time-varying component [7, 11], and muscle synergy weighting before and after fatigue using SPSS (ver 24.0, USA). The significance level was set at 0.05.

3. Results

Figure 1 shows the mean frequency of each muscle before and after fatigue. Among them, the mean frequency of the ADD decreased significantly ($P = 0.036$, effect size: 0.67). Table 1 shows the relationship between the number of modules and global and local VAF. When 2 synergies occurred, the mean global VAF exceeded 90% for the first time. However, the mean local VAF did not exceed 75%. On the other hand, when 3 synergies occurred, the mean global and local VAF exceeded 90% and 75%, respectively. Therefore, 3 synergies were compared before and after fatigue in this study.

Figure 2 shows the extracted modules before and after fatigue. The SPs of modules 1, 2, and 3, which indicate the coincidence of muscle synergies, were 0.96, 0.97, and 0.86, respectively. For synergy 1, we found no significant difference in the peak activation timing ($P = 0.909$, effect size: 0.15). The activation timing of synergy 2 was significantly delayed after fatigue ($P = 0.028$, effect size: 0.54). Before fatigue, the activity of synergy 2 was observed in the first half of the sidestep sequence; however, it was seen in the second half after fatigue. The activation timing of synergy 3 did not differ before or after fatigue ($P = 0.353$, effect size: 0.36), and it occurred in the second half of the sidestep sequence. In addition, the weighting of ST decreased ($P = 0.023$, effect size:

Table 1: Relationship between the number of synergies and the global and local VAF. The number of synergies is decided when the global VAF exceeds 90% and the local VAF exceeds 75% for the first time.

		Number of synergies = 2		Number of synergies = 3	
		Pre	Post	Pre	Post
Global VAF (%)		91.8 ± 5.0	94.9 ± 5.3	96.4 ± 2.7	98.1 ± 1.8
Local VAF (%)	RA	35.2 ± 17.9	55.4 ± 40.0	81.1 ± 10.7	90.0 ± 12.2
	EO	64.6 ± 22.6	77.9 ± 16.9	86.5 ± 12.1	91.1 ± 10.1
	IO/TrA	54.6 ± 35.0	88.9 ± 10.1	95.6 ± 7.1	92.9 ± 7.9
	ES	69.0 ± 24.2	83.3 ± 19.7	88.3 ± 16.3	88.2 ± 11.9
	RF	50.7 ± 23.6	70.4 ± 17.4	89.8 ± 8.5	79.1 ± 15.4
	ST	48.9 ± 24.5	76.8 ± 25.7	86.7 ± 10.6	97.7 ± 3.9
	Gmed	44.4 ± 30.6	59.0 ± 34.1	88.7 ± 11.0	90.4 ± 12.2
	ADD	45.5 ± 24.5	69.7 ± 27.5	82.0 ± 13.4	88.7 ± 9.9

VAF: variance accounted for; RA: rectus abdominis; EO: external oblique; IO/TrA: internal oblique/transversus abdominis; ES: erector spinae; RF: rectus femoris; ST: semitendinosus; Gmed: gluteus medius; ADD: adductor.

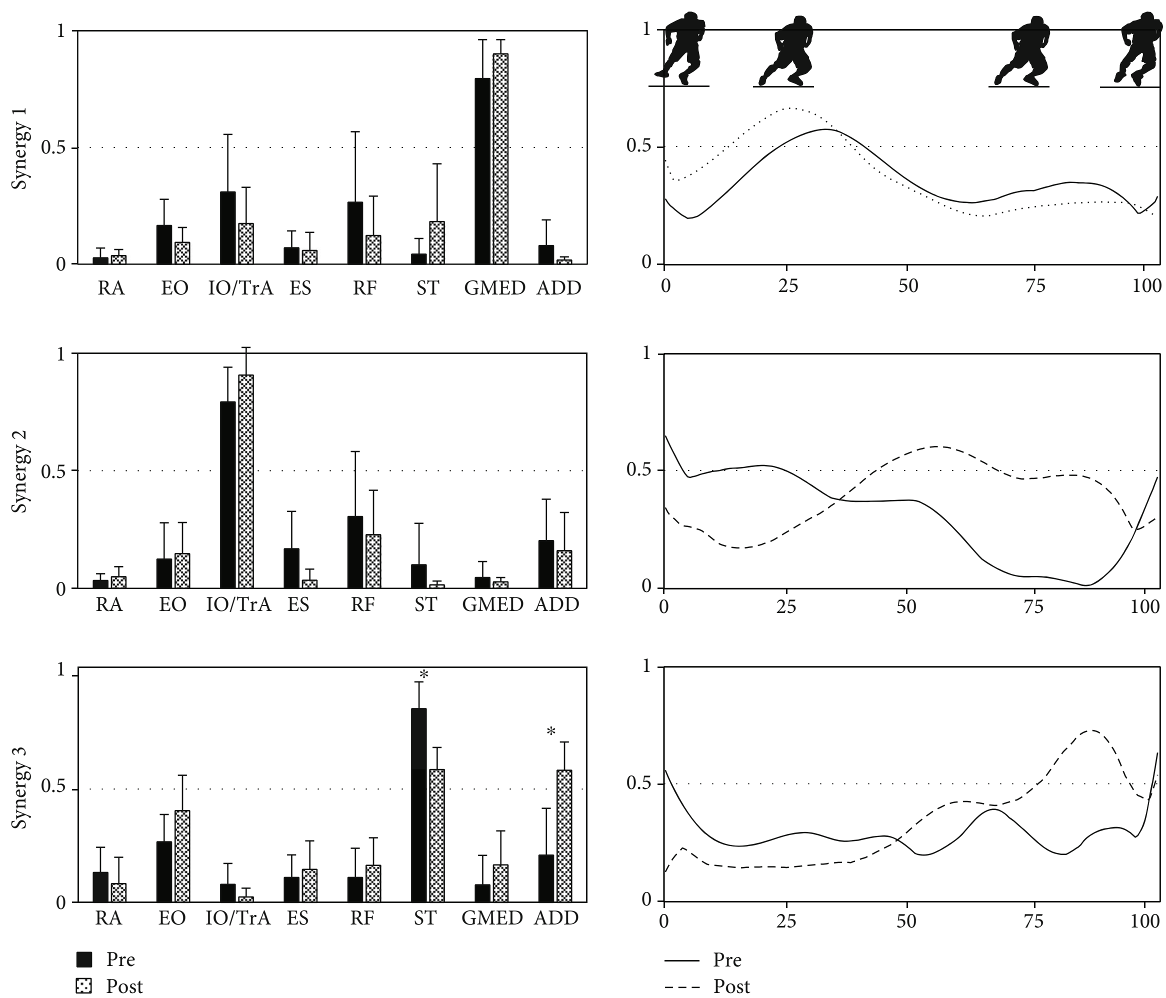

Figure 2: Extracted synergies (left) and the activation pattern (right) during side cutting before and after fatigue. RA: rectus abdominis; EO: external oblique; IO/TrA: internal oblique/transversus abdominis; ES: erector spinae; RF: rectus femoris; ST: semitendinosus; Gmed: gluteus medius; ADD: adductor muscle. *Significant difference ($P < 0.05$).

0.54), while that of ADD increased significantly ($P = 0.032$, effect size: 0.54).

4. Discussion

This study investigated modular control during a sidestep sequence before and after fatigue. The results indicate that the synergies were similar before and after the intervention, and the main finding of this study was that the activation timing of synergy 2 was delayed after fatigue.

Muscle fatigue was confirmed using wavelet analysis. The spectrum shifts toward lower values if muscle fatigue occurs [26, 27]. In our study, the median frequency of ADD decreased significantly after fatigue. This result indicated that ADD was fatigued after the fatigue intervention.

The data from synergy 1 mainly reflected the Gmed and IO/TrA activities during the first half of the sidestep sequence. The activation level of synergy 1 gradually increased, and the peak activation was 25–35% of the sidestep sequence corresponding to the time of landing. The Gmed was the highest contributing muscle, and the IO/TrA and RF activities were also notable in this synergy. The extensor muscles of the lower limb began activating before landing to enhance joint stability and prevent injury. Therefore, it is theorized that synergy 1 functions in response to weight loads [11, 28–34].

Synergy 2 indicated coordination among IO/TrA, RF, and ADD. Before fatigue, this synergy was activated before landing until the middle stage of the sidestep sequence. It is considered that this synergy has a role in hip control because the hip abduction moment occurs after landing [11, 35]. In contrast, the activation timing of this module is delayed after fatigue. Matsunaga et al. [11] reported that synergy 2 is delayed in subjects with groin pain when compared with healthy subjects. Therefore, we believe that this timing delay may be related to the development of groin pain.

Synergy 3 engaged mainly the EO and ST and was activated in the last one-third of the sidestep sequence, corresponding to toe off. The EO contributes to trunk stability and ST activates hip extension. Therefore, synergy 3 is involved in the movement of kicking off the ground. In addition, the weighting of the ST decreased after fatigue, and the contribution of the ADD may show compensatory increases because the main function of the ADD is hip adduction, with hip extension being a supplemental function. The risk factors for groin pain are overuse [14, 19, 20, 36], dysfunction of the ADD [37–42], and hip stability [18, 43]. These reports indicate that compensatory ADD activity may influence the pathogenesis of groin pain.

This study had some limitations. First, our sample size was small. However, the results showed a sufficient effect size; therefore, we believe that the impact on the data in this study was small. In addition, we did not include women as our subjects because we attached devices on the unclothed upper body. This might have affected the sample size. Second, we could not perform a full motion analysis; therefore, we focused only on the timing of foot contact and toe off the ground to divide the sidestep into phases. Third, we did not judge fatigue using other methods.

5. Conclusion

We examined trunk and lower limb muscle coordination during the sidestep sequence and compared it before and after the intervention. As a result, the same 3 synergies were observed before and after the intervention. However, the activation timing of the synergy that functioned as a hip adduction control was delayed due to the intervention. This motor control deficiency may be related to groin pain syndrome and other injuries to the hip.

Data Availability

The data will be used in a future study.

Conflicts of Interest

The authors declare that there is no conflict of interests.

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Research Article

Do Novice Runners Show Greater Changes in Biomechanical Parameters?

Wenjing Quan,[1,2] Feng Ren ,[1] Dong Sun ,[1] Gusztáv Fekete,[1,2] and Yuhuan He[1]

[1]*Faculty of Sports Science, Ningbo University, China*
[2]*Savaria Institute of Technology, Eötvös Loránd University, Hungary*

Correspondence should be addressed to Feng Ren; renfengnb@yeah.net

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Purpose. Examining and understanding the biomechanics of novice runners and experienced runners can further improve our knowledge within the field of running mechanics and running injuries. The purpose of this study was to classify the differences in lower limb biomechanics during a 3.3 m/s running task among both experienced runners and novice runners. *Method*. Twenty-four participants (12 experienced runners and 12 novice runners) ran at 3.3 m/s across a force plate; kinematics and kinetics data were collected by the Vicon motion system and Kistler force plate. Group comparisons were made using an independent samples t-test to identify differences in the impact peak, loading rate, contact time, ankle, knee, and hip joint kinematics and kinetics during the stance phase. *Results*. No significant differences were observed between novice and experienced runners for both ankle and knee joint kinetics except that the ankle joint plantar flexion torque was significantly greater in the novice runners. However, the plantar flexion, dorsiflexion, range of motion (ROM), plantar flexion torque, and max angular velocity of ankle joint significantly increased in novice runners than inexperienced runners. Additionally, the flexion angle and range of motion of the hip joint were observed to be larger in the novice runners. Moreover, the maximum extension torque and the maximum extension power in the hip joint were significantly increased in the experienced runners. There were no significant differences in the first peak, contact time, and average vertical loading rate. Novice runners showed a larger vertical instantaneous loading rate than experienced runners. *Conclusion*. These preliminary findings indicate that novice runners are prone to running injuries in comparison to experienced runners. Novice runners showed larger kinematics and kinetic parameters in the joint of the ankle and hip. Novice runners should enhance muscle strength in the hip and choose scientific training methods.

1. Introduction

Running is one of the most popular recreational physical activities in the world. Regular running helps prevent the incidence of chronic diseases, such as cardiovascular disease and obesity [1, 2]. Because of easy accessibility, many people prefer participating in long-distance running which can increase cardiopulmonary function and relieve psychological stress [3]. Unfortunately, excessive running can trigger running-related injuries and musculoskeletal injuries to develop [4, 5]. Running injuries are mainly lower limb injuries, primarily knee joint injuries, especially in the front of the knee (such as patellofemoral joint pain) [6–8]. Other common injuries include strains of the tibia, Achilles tendon, gastrocnemius, foot, and thigh muscles [4].

A previous study has shown that the risks of overuse running injury were increased from 20% to 70% in recreational and competitive distance runners [9]. Videbæk et al. have demonstrated the incidence of injury per 1000 hours of running, in which the rate of injury was 17.8% of novice runners compared to recreational runners (7.7%) and ultramarathon runners (7.2%) [10]. Of all populations, novice runners experience a high rate of injury. Novice runner's injury rate was higher compared to recreational, competitive, or marathon runners [11]. It is important to focus on injury prevention among novice runners. Nevertheless, there are few research recommendations for novice runners who desire to begin running training. There are many reasons which can cause running injuries, such as error training, the difference in running surface, different running habits, and running shoes [12, 13]. Although scientific researchers and clinical staff have been working hard to help runners reduce running-related injuries, the incidence of injuries has remained high for many years [14].

Epidemiological studies have found that overuse injuries were associated with kinematic variables of lower limb joints: the increased hip interrotation and hip adduction [13, 15]. Novacheck also found that the increased eversion angle velocity and ankle eversion angle might trigger the development of overuse injuries (Sallis et al., 1992). Running-related injuries were associated with ground reaction force, specifically increased vertical loading rate and vertical instantaneous loading rate, and the first peak caused the tibial stress fractures [16].

Running-related injuries especially in the knee joint have the characteristics of the frequent occurrence in people without running experience [17, 18]. Psychological fear of running-related injuries makes it difficult for nonrunning habit groups to form running habits [19], which hinders the widespread development of running.

Thus, several studies show a biomechanical difference between novice and experienced runners. Schmitz et al. found that there were no significant differences in impact peak, loading rate, peak nonsagittal hip kinematics, or strength among the novice runners and competitive runners. However, novice runners showed larger peak hip internal rotation and a decrease in trunk side-plank endurance [20]. When novice runners and competitive runners ran in a state of fatigue, novice runners showed larger hip abduction and peak trunk lean during midswing [21]. Van Mechelen proposed that about 50% to 75% of sports injuries may be due to overuse injuries caused by the repeated repetition of the same action. Factors related to running injuries include a history of previous sports injuries, a lack of running experience, participation in running competitions, and running long distances per week [22]. Moreover, the effect of running experience on the kinematics and kinetic energy of the lower limb remains unclear. Thus, the purpose of this study was to determine the effect of running experienced on lower limb biomechanical changes during the stance phase at 3.3 m/s among both experienced runners and novice runners. The hypotheses were that the novice runners' group would show higher changes in kinematics and kinetics when compared with experienced runners.

2. Methods

2.1. Participants. Two populations were recruited using flyers around the society and university: experienced runners and novice runners. The experienced runners consisted of 12 males that had been running at least 20 miles per week and the running experience was more than 5 years. The novice runner consisted of 12 males who ran 2 or 5 miles per week. A novice runner was defined as an individual having no former experience in running and never taken part in a running competition. All information about the 24 endurance runners is given in Table 1. Only subjects having the target foot length of US size 9 (±0.5) and self-reported as right leg dominant (defined as the preferred kicking leg) were included. Exclusion criteria consisted of any spinal or lower extremity surgery or any knee ligament or cartilage pathology in the past year. For this test, all the participants were rearfoot strikers (RFS) [23]. Written informed consent was obtained from the subjects, and the testing procedures were approved by Ningbo University.

Table 1: The basic demographics of subjects ($n = 24$).

Characteristic	Experienced	Novice
Age (years)	26.20 ± 4.10	25.60 ± 4.70
Weight (kg)	63.40 ± 7.50	67.50 ± 6.80
Height (cm)	170.00 ± 8.28	173.00 ± 7.28
BMI (kg/m^2)	21.75 ± 2.60	22.89 ± 3.20
Running experience (years)	5.20 ± 3.00	2.10 ± 1.60

2.2. Biomechanical Modeling and Collecting. An eight-camera motion analysis system (Vicon, Metrics Ltd., Oxford, UK) was used to capture the sagittal plane kinematics of the dominant lower extremity at a frequency of 200 Hz [24]. Participants were required to wear tight-fitting pants and T-shirts. All subjects ran with the right foot stepping on a single embedded force plate (Kistler Type, 9281B, Kistler Instrument AG, Winterthur, Switzerland) with dimensions of 600 × 900 mm, which was fixed in the middle of the 15 m walkway and was utilized to collect the ground reaction force (GRF) at a frequency of 1000 Hz. The heel strike and toe-off were determined when the vertical GRF crossed a 30 N threshold level [21]. Kinematic data were collected including angle changes of the lower limb joints (hip, knee, and ankle) in sagittal planes during the stance phase. Kinetic parameters were ground reaction force, joint moment, and joint power.

Retroreflective markers were placed on the subjects according to previous research which included thigh, shank, and ankle [25] (Figure 1). Twenty-five retroreflective markers (diameter: 14.0 mm) were used to define the knee, ankle, and hip segments. The marker locations included right and left anterior superior iliac spine, left and right posterior superior iliac spine, right and left greater trochanter, first and fifth metatarsal heads, distal interphalangeal joint of the second toe, medial and lateral malleoli, and medial and lateral epicondyle of the femur; tracking clusters were placed on the lateral thigh, shank, and right heel (Table 2).

2.3. Running Protocol. All participants wore the same type of running shoe, Anta (Flashedge, China). Participants were instructed to warm up with light jogging and

Figure 1: Placement of combined marker set consisting of retroreflective cluster markers and single 14 mm retroreflective markers.

Table 2: Anthropometric data.

Segment	Definition	Center of mass (%)	Radius gyration (%)
Foot	Lateral malleolus/head metatarsal II	1.37	4.415
Shank	Femoral condyles/medial malleolus	4.33	4.395
Thigh	Greater trochanter/femoral condyles	14.16	40.95

stretching in the common shoes. They then ran for at least 5 minutes in the laboratory at a self-generated comfortable speed. Runners performed each trial by running through a laboratory that was 15 m long and exiting into a hallway. On both sides of the force platform was a speed-measuring instrument (smart speed, Fusion Sport Inc., Burbank, CA, USA) to control the speed of the subjects. The distance between the speed-measuring instrument was 3.3 m. All the subjects ran at a speed of 3.3 m/s. Each test collected six successful trials (Figure 2).

2.4. Data Analysis. This study paid more attention to the variation of the sagittal plane as the main data [26]. Visual 3D (C-motion, Germantown, MD, USA) was used to process the data. First, a fourth-order low-pass zero-lag Butterworth filter was used to filter the marker trajectories at 15 Hz and force plate data at 100 Hz [26].

Sagittal plane hip, knee, and ankle angles were calculated using Cardan angles with the distal segment expressed relative to the proximal segment in Visual 3D. The net internal joint moments and joint powers were calculated using a standard inverse dynamics approach. Segment masses, the center of mass locations, and inertial properties were calculated for the thigh, shank, and foot using anthropometric data [27]. The joint kinetic and the GRF variables were normalized by the subject's body mass. Joint angles, joint moments, and powers were normalized to the stance phase over 101 data points. Max angles were defined as the maximum joint angle during the stance phase, while participants ran the 15 m distance. Min angles were defined as the minimum joint angle during the stance phase. The range of motion was defined as the maximum angle minus the minimum angle. The average vertical loading rate (VALR) and vertical instantaneous loading rate (VILR) were calculated over the portion of the vertical GRF (vGRF) vs. time curve between 20 and 80% of the time to peak impact according to Equations (1) and (2) (Milner et al., 2008).

$$\mathrm{VALR} = \frac{F_{80}\% - F_{20}\%}{t_{80}\% - t_{20}\%}, \tag{1}$$

$$\mathrm{VILR} = \frac{\Delta F_{\max}}{\Delta t} \text{ where } (t_{20}\% < t < t_{80}), \tag{2}$$

Kinematic variables of two groups of runners included eversion and dorsiflexion angles (ankle, knee, and hip), as well as joint (ankle, knee, and hip) angle velocity in the sagittal plane (Figure 3). Kinetic variables included contact time, average vertical loading rate (VALR), vertical instantaneous loading rate (VILR), first peak, joint moment, and joint power.

2.5. Statistical Analysis. All data are given as mean ± standard deviation. Normal distribution and homogeneity were assessed using the Shapiro-Wilk test and Levene's test, respectively. Using SPSS (SPSS Inc., Chicago, IL), an independent samples t-test was used to assess group differences for kinematic and kinetic parameters. The level of significance was set to $p = 0.05$.

3. Results

3.1. Kinematics of Ankle, Knee, and Hip Joints. The Shapiro-Wilk tests revealed that all parameters were normally distributed. There was no significant difference in the max knee

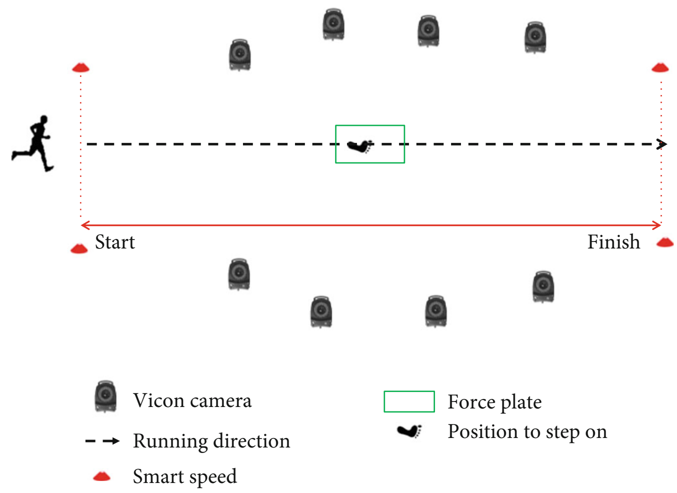

Figure 2: Participant motion capture setup.

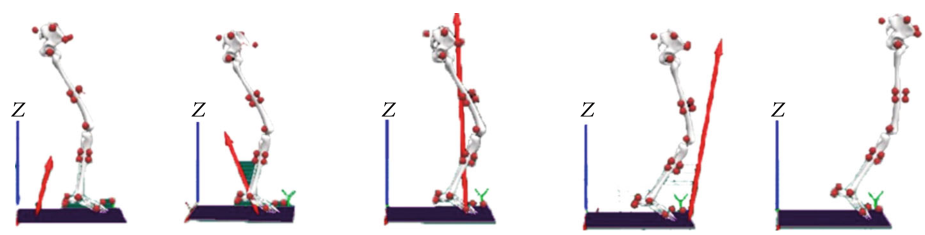

Figure 3: Pictorial illustration of the running gait cycle during the stance phase at 3.3 m/s.

Table 3: Ankle, knee, and hip joint kinematics during the stance phase ($n = 24$).

Joint	Variables	Experienced	Novice	p value
Ankle	Max angle (°)	8.20 ± 1.60	23.70 ± 1.11	$p < 0.01$ *
	Min angle (°)	−14.51 ± 2.66	−8.20 ± 3.47	$p < 0.01$ *
	ROM (°)	22.72 ± 2.53	31.90 ± 3.89	$p < 0.01$ *
	Max angular velocity (°/s)	180.98 ± 29.20	205.19 ± 15.19	0.026*
Knee	Max angle (°)	−7.70 ± 3.00	−20.50 ± 2.56	$p < 0.01$ *
	Min angle (°)	−32.82 ± 3.01	−49.06 ± 2.09	$p < 0.01$ *
	ROM (°)	25.11 ± 2.98	28.56 ± 4.31	0.041*
	Max angular velocity (°/s)	181.71 ± 33.05	205.86 ± 63.13	0.277
Hip	Max angle (°)	26.07 ± 2.89	32.69 ± 2.15	$p < 0.01$ *
	Min angle (°)	−1.63 ± 4.98	−2.58 ± 2.90	0.059
	ROM (°)	27.71 ± 4.10	35.27 ± 2.57	$p < 0.01$ *
	Max angular velocity (°/s)	19.11 ± 18.64	21.91 ± 45.06	0.851

Note: *significant difference between experienced runners and novice runners ($p < 0.05$).

angular velocity, min hip angle, and max hip angular velocity. When analyzing the changes in the joint angles of novice runners during the stance phase (Table 3), maximum ankle angle ($p < 0.01$), minimum ankle angle ($p < 0.05$), ROM of ankle joint ($p < 0.01$), maximum hip angle ($p < 0.01$), ROM of the hip joint ($p < 0.01$), and ROM of knee joint ($p < 0.01$) were increased. In addition, decreased changes were observed in the maximum knee angle ($p < 0.01$) and minimum knee angles ($p < 0.01$) in the novice runners (Figure 4).

3.2. Kinetics of Ankle, Knee, and Hip Joints. Minimum moment of the hip joint ($p < 0.05$) and the maximum power of the hip joint ($p < 0.05$) were significantly smaller in the novice runners than in the experienced runners (Table 4). The minimum ankle moment was significantly greater in the novice runners than in the experienced runners (Table 4) and (Figure 5). However, there were no significant differences in the maximum moment, maximum power, minimum power of ankle joint, maximum moment, and

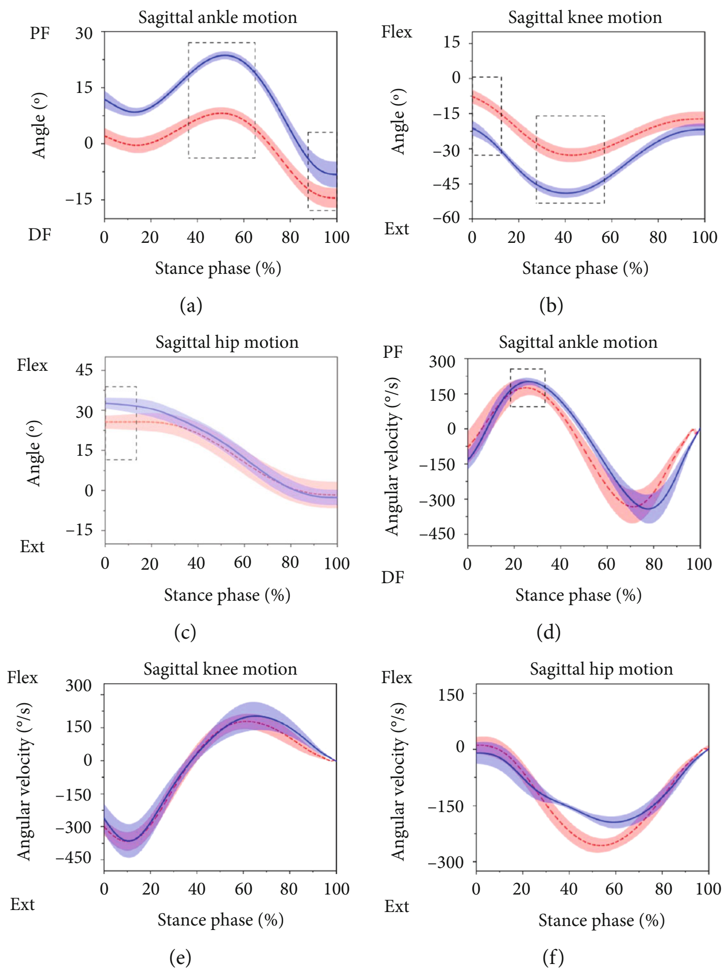

Figure 4: Sagittal ankle, knee, and hip joint kinematics for novice runners (the solid blue line is the mean and the shaded area is the standard deviation) and experienced runners (dashed red line is the mean). Note: the dotted box indicates a significant difference between the two groups of runners, $p < 0.05$.

Table 4: Ankle, knee, and hip joint kinetics during the stance phase ($n = 24$).

Joint	Variables	Experienced	Novice	p value
Ankle	Max moment (Nm)	0.50 ± 0.24	0.43 ± 0.14	0.356
	Min moment (Nm)	-1.85 ± 0.32	-2.22 ± 0.11	0.002*
	Max power (W/kg)	10.43 ± 2.87	11.44 ± 2.35	0.379
	Min power (W/kg)	-3.15 ± 0.85	-4.03 ± 0.77	0.018
Knee	Max moment (Nm)	0.93 ± 0.23	1.08 ± 0.44	0.379
	Min moment (Nm)	-2.71 ± 0.22	-2.63 ± 0.27	0.295
	Max power (W/kg)	1.02 ± 0.43	1.15 ± 0.58	0.530
	Min power (W/kg)	-9.52 ± 2.36	-9.16 ± 1.87	0.681
Hip	Max moment (Nm)	0.82 ± 0.24	0.80 ± 0.12	0.747
	Min moment (Nm)	-4.37 ± 0.45	-3.67 ± 0.48	0.002*
	Max power (W/kg)	17.09 ± 2.81	12.45 ± 3.30	0.002*
	Min power (W/kg)	-1.24 ± 0.60	-1.15 ± 3.26	0.748

Note: *significant difference between experienced runners and novice runners ($p < 0.05$).

minimum power of hip joint. No significant differences existed in the kinematic parameters of the knee joint (Table 4).

3.3. Kinetics of Ground Reaction Force. Contact time increased significantly among novice runners compared to the experienced runners ($p < 0.01$) (Table 5). No significant difference was observed in the vertical average loading rate, contact time, and first peak. Besides, the vertical instantaneous loading rate was lower in the novice runners in comparison to the inexperienced runners ($p < 0.01$).

4. Discussion

Future research directions may also be highlighted [28, 29]. Compared with the ankle variables, the plantar flexion, dorsiflexion, ROM, plantar flexion torque, and maximum angular velocity were significantly increased in novice runners when compared to inexperienced runners. Long-distance running may cause plantar fasciitis and metatarsal stress fracture-related running injuries.

A previous epidemiological investigation found that the knee joint of novice runners is the most prone to

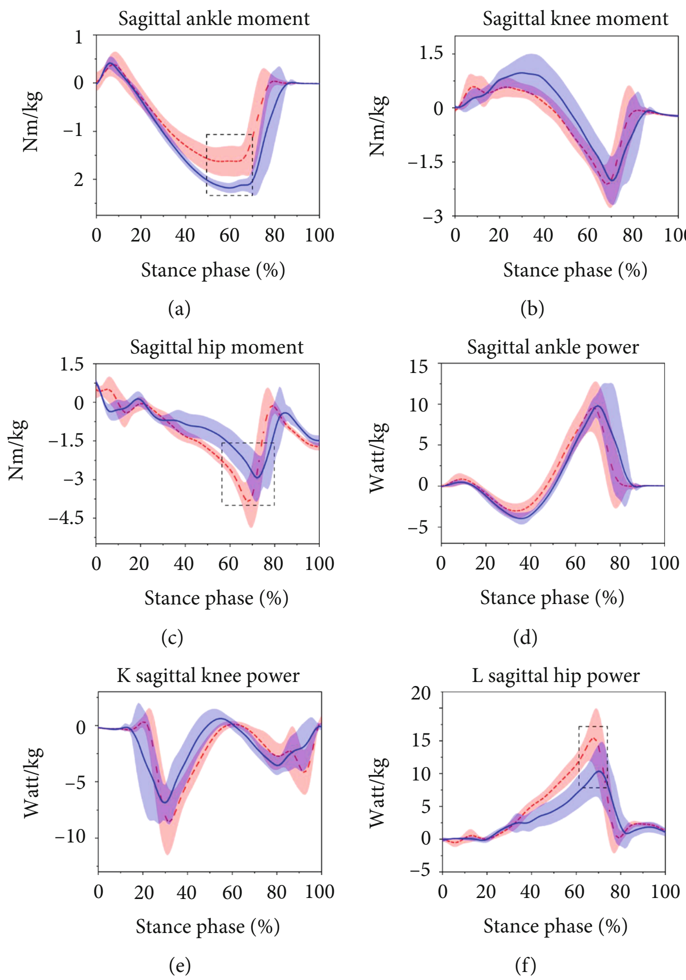

FIGURE 5: Sagittal ankle, knee, and hip joint power and moment for novice runners (the solid blue line is the mean and the shaded area is the standard deviation) and experienced runners (dashed red line is mean). Note: the dotted box indicates a significant difference between the two groups of runners, $p < 0.05$.

TABLE 5: Ground reaction force parameter during the stance phase ($n = 24$).

Parameter	Experienced	Novice	p value
Contact time (ms)	231.00 ± 11.97	230.00 ± 11.97	0.67
Vertical average loading rate (BW/S)	52.58 ± 15.78	48.13 ± 3.60	0.405
Vertical instantaneous loading rate (BW/S)	106.13 ± 41.53	89.00 ± 9.96	0.001*
First peak (BW)	2.15 ± 0.20	2.45 ± 0.18	0.852

Note: *significant difference between experienced runners and novice runners ($p < 0.05$).

injury [30]. Our study results showed that the maximum knee angle and minimum knee angles were smaller than experienced runners. Dierks et al. found that increased knee flexion helps reduce the risk of a knee injury. However, in this study, novice runners observed larger knee flexion than experienced runners [31]. The ROM of the knee joint was larger in the novice runners than experienced runners, and this finding is in agreement with Agresta et al. [32]. This could be attributed to novice runners having poor running mechanics, which results in higher loads on musculoskeletal tissue, especially at the tibia and the knee.

The novice did show greater hip joint flexion angle and ROM of the hip joint in this study. The hip joint plays a very important role in the movement of the lower limbs, and the instability of the hip joint is considered to be an important mechanism of lower limb injuries [9]. In the sagittal plane, the novice runners produced a larger ROM in comparison to the experienced runners. This may suggest poor hip stability among novice runners. The maximum extension torque and the maximum extension power in the hip joint significantly increased in the experienced runners. This phenomenon might have been caused by running miles and running speed. In addition, insufficient hip abductor muscle strength and abnormal anatomical force lines may also affect it [9, 33]. Also, larger extension torque and extension power in the hip joint might lead to the development of iliotibial bundle friction syndrome.

The curve of the ground reaction force during running is a typical double-peak curve. Studies suggest that the increase in the ground reaction force peak and its loading rate will cause higher risks to lower limb injuries [34–36]. In this study, the peak ground reaction force and the corresponding average load rate of the ground reaction forces were consistent with the results of Schmitz et al., who used the same test speed in their experiments [20]. However, the vertical instantaneous loading rate was lower in the novice runners. Many factors may influence the ground reaction force parameters.

Although the ground reaction force parameters were associated with running injuries, our results do not provide more details into novice runners who have a higher rate of running injuries than experienced runners. For novice runners, the risk of running injury was higher than experienced runners. Novice runners should enhance muscle strength in the hip and choose scientific training methods. During the training sessions, novice runners should increase the amount of running and control the running speed on a step-by-step basis and reasonably.

There are some potential limitations to this study. In this study, the anteroposterior ground reaction force was not calculated. The data of anteroposterior ground reaction force might provide a helpful understanding of overuse running injuries for both novice and experienced runners. Moreover, the different running speeds should be considered when compared to the biomechanics parameters. Finally, a further study should focus on the effect of different gender and different BMI.

Data Availability

The data that support the findings of this study are available from the corresponding author upon reasonable request.

Conflicts of Interest

There is no conflict of interests.

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Review Article

How to Improve the Standing Long Jump Performance? A Mininarrative Review

Huiyu Zhou,[1] Peimin Yu,[1,2] Anand Thirupathi ,[1] and Minjun Liang [1,2]

[1]*Faculty of Sports Science, Ningbo University, Ningbo, China*
[2]*Research Academy of Grand Health, Ningbo University, Ningbo, China*

Correspondence should be addressed to Anand Thirupathi; ananthzeal@gmail.com and Minjun Liang; liangminjun@nbu.edu.cn

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Standing long jump (SLJ) is complicated by the challenge of motor coordination in both the upper and lower segments. This movement is also considered to be a fundamental skill in a variety of sports. In particular, SLJ is an important test index for middle school students for assessing their physical fitness levels. This assessment takes the form of a physical fitness test high school entrance examination in some countries such as China. This minireview summarizes recent studies that have investigated how to improve the standing long jump performance from different aspects which include arm motion, takeoff angle, standing posture, warming-up exercise, and handheld weight. The common study limitations, controversial knowledge, and future research direction are also discussed in detail.

1. Introduction

Jumping is a fundamental movement skill in a variety of sports that needs the complex motor coordination of upper and lower limbs to obtain a good performance, such as volleyball, basketball, ski jumping, and some ball sports in which the high velocity of muscle contractions is required. Standing long jump (SLJ) is considered a good predictor of sprint and jump performance, which presents correlations highly with isokinetic measures of lower extremity force [1].

The SLJ is an important physical fitness test index for middle school students in China. This assessment metric takes the form of a physical fitness test high school entrance examination in China. More often than not the outcome of this physical fitness test determines the eligibility of the student to be admitted to their high school of choice. Therefore, the SLJ score is of particular importance, and examining methods to effectively improve SLJ performance could be beneficial to middle school students. Previous researchers have also investigated various aspects of the SLJ ([2–4]. These studies have investigated body configuration and joint function of both upper and lower segments between adults and preschool-age children [3] and explored the significant correlations between a variety of isometric, kinetic, kinematic, and SLJ performance parameters [2, 4]. The category of jumps can be divided according to the arm motion. i.e., jumps with restricted arm motion (JRA) and jumps with free arm movement (JFA). In terms of comparing the JRA and JFA jump categories, the influence of arm movement on SLJ performance was explored with respect to understanding whether the jumping distance could be increased by arm swing [5]. To detect more biomechanical mechanisms for improving the movement performance, the optimum takeoff (TO) angle 29°-38° of SLJ in adult males was found, although the biomechanical evidence for this result is unclear [2, 3, 5]. Meanwhile, the different TO positions for SLJ have also been studied by Mackala and associates where the different results of parallel or straddle foot starting placement for the quality of SLJ were determined [6]. Moreover, Koch et al. explored the potential effects of stretching and warm-up activities on the SLJ in moderate and well-trained subjects. The results presented show that warm-up activities had a slight effect on jumping performance, while maximum muscles strength showed a significant correlation with jumping ability [7]. On the other hand, according to the research by Minetti and Ardigo, the hand with halteres increased the TO speed in the SLJ mainly because the muscle exerted more force in the moderately loaded subjects in comparison to the non-loaded scenario [8]. The researchers compared the effects of hand-holding different weights on the kinematic and dynamic characteristics of SLJ subjects, and the relationship between the hand-held weight and jumping performance was clarified (Fukashiro et al. 2005; [8]).

Besides, the biomechanical characteristics of SLJ from a computer modeling perspective have been investigated by Hickox et al. Hickox and colleagues verified the effectiveness of SLJ modeling based on the two-dimensional sagittal plane evaluation, and the results showed that plane analysis was sufficient to detect lower limb movement [9]. In addition, Ashby and Delp documented that arm activity can improve the SLJ performance by using the optimal control simulation method [10]. These have provided us with an insightful understanding of the sports coordination mechanism of SLJ.

To date, a narrative review on the effect of motor activity on the standing long jump performance remains unavailable in the literature. Therefore, the purpose of this article was to summarize the methods for coaches and trainers to improve the SLJ performance from the perspective of several aspects based on previous studies.

2. The Different Analytical Aspects of the Standing Long Jump

2.1. The Role of Arm Motion in the Standing Long Jump. Many previous studies have elucidated the role of arm movements in various jump activities [11]. There are several benefits to arm motion, such as arm swing increases the velocity of the body's center of gravity (CG) at TO, acquires the larger peak magnitude of the vertical ground reaction force, and creates an additional downward force on the body which allows for greater muscle force development [11–13]. To be more specific, Ashby and Heegaard have revealed the biomechanical mechanism of the role of an arm in the SLJ [5]. They conducted a comparative study between JFA and JRA subjects; the results showed that the average distance improved by 21.2% in the JFA group compared to JRA, the average velocity of the CG increased by 12.7% at TO, and the horizontal displacement of CG before TO significantly increased among the JFA subjects when compared to JRA subjects. In terms of kinetics, the peak value of horizontal ground reaction force (HGRF) in the JFA group was also significantly increased when compared to the JRA group. Additionally, it was considered that majority of the improvements observed in the SLJ were attributed to the increased CG velocity at TO during arm movements.

Three different theories have been proposed to explain the principle of how CG velocity was increased by arm movement at TO. The theory of "hold back" indicated that the lower limb extensor was activated by arm motion during the propulsive phase to limit excessive forward rotation, which would achieve an optimal landing. On the contrary, if arm motion was restricted, the jumper must "hole back" to limit the lower limb extensor thereby avoiding excessive forward rotation of the trunk and legs that would limit proper landing [5]. The theory of "joint torque augmentation" suggests that the arm swing creates a downward force on the shoulder, which slows down the shortening velocities of the lower extremity joint extensors thus resulting in a greater muscle torque [11, 12]. The "energy transfer" theory is that muscles in the shoulder and elbow joints transfer energy to the rest of the body before takeoff, increasing the speed and displacement of the CG in both horizontal and vertical directions [12]. Ashby assessed the reliability of all three theories in jumping movement by using the optimal control simulations; it was found that the "energy transfer" theory is the primary mechanism for increasing the velocity of the CG in JFA at TO, because the large work of the upper limb joint muscles is produced by free arm movement which can be effectively transferred to the lower limb [14]. Above all, jumping with a free arm movement can significantly improve SLJ performance.

2.2. The Optimum Takeoff Angle of CG. The trajectory of CG movement can be likened to a projectile in the flying phase of the SLJ. Therefore, an appropriate projection angle is identified as a crucial factor to develop an ideal performance. Previous studies have suggested that the projection angle between 29° and 38° has been considered as an optimum TO angle for jumpers, but the biomechanical reasons for this projection angle option are not well explained [2, 3, 5]. However, the results showed that the TO angle was not the main factor contributing to a successful SLJ performance, especially that the distance affecting by TO speed was more important than TO angle [15]. In order to obtain maximum TO speed, the optimum TO angle in SLJ was suggested to be less than 45° [15]. Additionally, the authors also suggested that spiked shoes should be used at very low takeoff angles to increase traction at TO phase so as to reach a greater jump distance. There are few researches that have been done on the optimum takeoff angle in SLJ movement. Further studies should be conducted in the future to verify the role of appropriate TO angle in SLJ.

2.3. The Standing Posture of SLJ. The coordination strategies of a jumper can be affected by different standing postures. Despite the conclusions by previous researches suggesting that jump distance is insensitive to the initial position, which is determined by angle of knee flexion and posterior angle of the trunk at TO phase, the initial postures play an important role in SLJ movement for attaining a good performance [16]. Actually, the effects of various foot positions on the quality of SLJ have been studied extensively specifically from parallel and straddle position perspectives. The parallel SLJ setup involves placement of the feet at shoulder width apart or more and parallel to the starting line. The straddle SLJ setup contrastingly involves placement of the feet in self-selected straddle position ranging from 30 cm to 40 cm with one of the feet in front. Mackala and colleagues have investigated the effect of the differences in kinematics and kinetics between parallel and straddle placement in SLJ movement. In their study, three related muscle group activities were evaluated by electromyography (EMG) in different foot placement groups. The results showed that the average distance can be improved by 5.18% in the straddle position when compared to the parallel position. More specifically, larger flexion angles at the trunk, hip, and knee joints were observed in the straddle position. Larger peak joint moments were also found at straddle feet placement in comparison to the parallel position. The subject's whole body was more likely to tilt forward in the straddle position and produce a lower center of mass that can generate a larger momentum in the forward and upward movements, thus contributing to a better performance. In contrast to parallel posture, the greatest muscle activation was observed in the gluteus maximus and biceps femoris during the push-off phase in the straddle starting foot position, and the lower limb extensor muscles such as gluteus maximus and biceps femoris could exert more force in the straddle position compared to parallel position [6].

Besides the experimental measurement of SLJ, long jump simulation researches have also been conducted. The numerical simulation could take the advantage of decreasing the biased effects. In order to effectively study the influence of starting posture on SLJ, a planar 4-segment human model has been established by Cheng and Chen [18] to detect the joint torque activation level and TO time in SLJ movement. Three different starting postures included the squat, low squat, and high squat were tested; the height of the squat was determined by the initial center of mass heights at 78 cm, 88.4 cm, and 62.9 cm, respectively. However, the results showed that the jump distance was slightly dependent on the initial posture [18]. It is a little difficult to draw a conclusion based on the current researches regarding whether SLJ performances can be influenced by starting posture, since different strategies of starting posture in selected articles have been used. One of the articles focused on the feet placement [19], and another was concerned about the height of squat [20]. We may be able to get some information from limited studies including preliminary studies to provide clarity in understanding the effects the straddle starting feet placement may have on jump distance during the SLJ movement. To fully understand the complexities associated with improving performance in the SLJ, further investigations into feet placement, different squatting heights, and other postures associated with the SLJ need to be investigated and understood so that future projects can provide more valuable information to the jumper and coaches.

2.4. The Effect of Warm-Up on the SLJ. The warm-up exercises are considered as an important factor for injury prevention and a prerequisite for good athletic performance. Stretching movements have been widely applied in warm-up exercises for training and competition purposes in a variety of sports [7]. Researchers have shown that after warming up, the muscle's stiffness is reduced and relevant muscles have more compliance before the sporting activity is started [21]. Furthermore, some studies have found that stretching contributes to a negative effect on muscle strength, performance, and strength endurance. [22–25]. Similarly, Koch et al. also detected the negative effect of different warm-up exercises which included stretching, high force, and high power in trained and untrained men and women. According to this research, the results revealed that no significant differences were found in any warm-up exercise routines [7]. It was demonstrated that the effect of warm-up exercises on SLJ performance was not obvious, and the muscle strength was strongly associated with jump ability. This finding is consistent with the conclusion drawn by Koch and colleagues who found that no effect on sports performance was observed during their investigation of a static stretch involving a standing/seated toe touch and standing/seated quadriceps stretch [7]. Even in the vertical jump movement, previous researches have indicated that a small (3%) reduction in height of the vertical jump was found after the performance of proprioceptive neuromuscular facilitation stretching [8]. Above all, the adverse effects of warm-up exercises on SLJ sports performance have been consistently confirmed by previous researches; therefore, the warm-up exercises are not recommended for SLJ movement.

2.5. The Function of Handheld Weight on SLJ Performance. The effect of handheld weights on jumping performance has been conducted by a few studies [8, 26]. Papadopoulos and associates demonstrated that each hand carrying a 3 kg load would contribute to a 6% increment in the jump distance performed at the same TO speed. In addition, the computer simulation presented when subjects were jumping with 2 kg to 9 kg weights in each hand showed that the velocity of TO can be increased by 5-7% [27]. The loading effect during jumping allowed muscles to exert larger strength which led to a reasonable muscle contraction [27]. Researchers have compared the different effects of various handheld masses on the kinematic and dynamic features of SLJ [8]. They suggested that better SLJ performance could be achievable with extra weights between 3 kg and 6 kg due to the larger horizontal translation of the COM and the greater GRF that was yielded. This conclusion is consistent with Lenoir and associates who showed that a jump distance of 13.88 ± 0.70 cm was achieved without loads while the distance was significantly increased with extra weights (14.64 ± 0.76 cm) [28]. Ashby also indicated that jumpers who carried a 4.6 kg loading increased their jump distance by 0.39 cm [26]. Furthermore, using a simulation analysis, Minetti and Ardigo noted that a 5 kg to 6 kg load is the optimal weight for increasing jump distance [27]. Subsequently, Huang et al. tested the optimal weights for SLJ jumpers and found it to be 5.6 kg (Huang et al. 2005). According to the analysis presented, the improvement in SLJ performance by extra weight is mainly attributed to greater GRF force and greater takeoff velocity of COM in the horizontal direction. Therefore, the method of holding extra loading to improve SLJ performance can be applied in a training program for different sports purposes.

3. Conclusion

Many studies reveal the effect of the object on standing long jump from a different perspective. The five methods that could influence the SLJ performance were included in this mininarrative review, in which the arm motion, takeoff angle less than 45°, and 5 kg-6 kg handled weight play a positive effect on SLJ performance. All these biomechanical variables identified as the main factors to achieve an ideal SLJ performance generally improved takeoff velocity of COM and increased the power of the lower extremity. On the other hand, warm-up exercises have presented a negative influence on SLJ movement since it reduces the muscle's stiffness and increases muscular compliance. There was a contradictory view in the starting posture of SLJ movement as indicated by the different strategies of starting posture in selected articles. Further studies on muscle activities in the lower extremities during the SLJ movement are needed since muscle strength is a determining factor to achieve better performance. On the other hand, the application of specialist jumping shoes in SLJ movement is also an important research topic since running shoes have been extensively investigated; however, to date, no research has focused on jumping shoes.

Data Availability

The data that support the findings of this study are available from the corresponding author upon reasonable request.

Conflicts of Interest

The authors declare that they have no conflicts of interest.

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