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Abstract

Exploring the internal mechanical mechanisms and injury mechanisms of table tennis topspin forehand and footwork from a biomechanical perspective could provide useful information for sports medicine and motor control, and could also provide a reference for table tennis footwear development and therapeutic equipment. Table tennis is generally considered a low-injury risk sport, which has led to a large amount of past research focusing on performance improvements. However, serious sports injuries are widespread among top table tennis players. The prevention of sports injuries and the improvement of movement control are also factors that cannot be ignored by athletes and coaches. This study uses a series of different biomechanical experiments and computer simulations to import experimental data into OpenSim to implement musculoskeletal simulation to explore the mechanical characteristics and possible injury risks of the lower limb trunk and joints in table tennis topspin forehand and footwork. There are three research objectives of this study which are as follows:

The first research objective is based on subject-specific musculoskeletal modeling and simulations to explore the injury risk, and motor control strategy of table tennis footwork, and guide the footwear design.

The second research objective is to explore the cryotherapy effect on balance recovery after fatigue to guide the cryotherapy equipment design and applications.

The third research objective is to reveal the intrinsic biomechanical mechanism of table tennis topspin forehand and provide guidance for optimization.

Use Matlab to perform two-factor repeated measures analysis of variance and one-dimensional statistical parametric mapping, and use SPSS to perform statistical analysis on the dispersion index. And the main results of this study were as follows:

First, the joint reaction force and joint stiffness of the subtalar are large, which indicates that the subtalar joint bears a large impact force during landing, which may cause injury

to the foot. The medial and lateral of the rearfoot showed high plantar pressure during the landing stage, and the big toe area and medial forefoot showed high plantar pressure during the forward phase. Strengthening the posterior muscles of the lower limbs can help improve stability during the landing phase.

Second, cryotherapy was not recommended for balance recovery if the competition was on the same day or within 24 hours but it was recommended if the competition was on the next day or after the next day.

Third, compared with the long-line topspin forehand, the cross-court topspin forehand shows a significant violent movement on lumbar left bending and flexion.

In conclusion, the present study innovates a multidisciplinary approach combining biomechanics, sports medicine, and rehabilitation sciences. Combine experimental and computational workflow to model joint loading and muscular contribution to reveal the mechanism during topspin forehand and footwork. The coach and athlete could acquire valuable information to optimize training strategy and enhance motor control. Relevant researchers could quickly establish a basic understanding and knowledge through this study.

Abbreviations

ANOVA: analysis of variance

AP: antero-posterior axis

BP: backswing phase

BW: body weight

CC: cross-court

CI: cryotherapy intervention group

CMC: computed muscle control

CON: control group

COP: center of pressure

CS: chasse-step

3D: 3 Dimensions

EA: elite athletes

EB: end of the backswing

EF: end of forward-swing

FP: forward-swing phase

FTI: force time integral

GRF: ground reaction force

HZ: hertz

ID: inverse dynamics

IK: inverse kinematics

JRF: joint reaction force

KG: kilogram

LAR: lumbar axial rotation

LBP: lower back pain

LF: lateral forefoot

LF: lumber flexion

LL: long-line

LLB: lumbar left lateral bending

LR: lateral rearfoot

M: midfoot

MA: medium athletes

MAXD: maximized reach distance

MF: medial forefoot

ML: medium-lateral axis

MR: medial rearfoot

MVC: maximal voluntary contraction

N: newton

NP: natural position

OS: one-step

PAR: pelvis axial rotation

PTI: pressure time integral

RM: repetition maximum

RMS: root mean square

ROM: range of motion

sEMG: surface electromyography

SPM1d: one-dimensional statistical parametric mapping

SSC: stretching-shortening cycle

T: toe

YBT: Y-balance test

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The Motivation

As one of the most popular racket sports, table tennis is not only loved by people all over the world in the field of public health but also receives widespread attention in the Olympic Games and competitive competitions. With the redesign of the size and material of table tennis balls and the adjustment of competition rules, the competitiveness of this sport has been further improved. After the size of the table tennis ball was adjusted to 40+, due to the increase in volume, the rotation speed of the ball decreased, which directly affected the development and application of the skills, tactics, and technology of this sport. Affected by the current rules, this sport is becoming more and more suitable for players who are good at forehand attack style or have more comprehensive skills. This makes table tennis players pay more attention to the development of physical function and muscle strength than ever before, and expect to enhance their aggression by increasing their hitting power to help score points. Therefore, in recent years, the biomechanical mechanism of forehand topspin has received a lot of attention and research from a large number of scholars.

Due to the enhancement of competitiveness and the improvement of physical function requirements, sports fatigue occurs more frequently during competition. Frequent exercise fatigue leads to the occurrence of acute injuries. In addition to acute injuries, overuse of the body due to extensive training and competition tasks among professional athletes has been proven to be a major injury factor in racket sports. Acute injuries in table tennis and injuries caused by overuse often occur in the lower limb joints, lower limb trunk, and shoulders, which greatly limits the sports life and performance of elite athletes.

The execution of tactics and the stable performance of technology are key factors for athletes to win in competitions. In table tennis, a perfect stroke play is the basic condition for scoring, which requires athletes to strengthen the movement control of the body and enhance the coordination and stability of the body. The ability to maintain body balance is an external manifestation of an athlete's body control and stability. The main thing that table tennis players need is to reach the designated area in a very short time to complete the hitting task. Due to the continuous improvement of competition level and changes in competition rules, the flight speed of the ball continues to increase, making players need to complete the hitting task in a shorter time. This further results in athletes often needing to maintain body balance and stability under extreme circumstances. However, this situation aggravates the occurrence of acute injuries.

In conclusion, the motivation of this study is to explore the inherent mechanical mechanism and possible injury risks of lower extremity joints and trunk in table tennis players during the topspin forehand from a biomechanical perspective. Help athletes and coaches formulate reasonable and scientific plans to improve sports performance during training and competition, avoid the occurrence of sports injuries, extend the sports life of athletes, provide useful information to footwear design, and promote the further development of table tennis around the world.

Research Objective

Based on the work motivation, the main research objectives of the dissertation are as follows.

The first research objective: Based on subject-specific musculoskeletal modeling and simulations to explore the injury risk and motor control strategy of table tennis footwork and guide the footwear design.

The second research objective: Explore the cryotherapy effect on balance recovery after fatigue to guide the cryotherapy equipment design and applications.

The third research objective: Reveal the intrinsic biomechanical mechanism of topspin forehand stroke, provide guidance for optimization.

1 Introduction

1.1 Lower Extremity Injury in Racket Sports

Table Tennis, badminton, Tennis, and squash are considered racket sports, and even tennis is the most popular sport in the world. Before discussing the possibilities and mechanisms that limit injury to athletes in these sports, we must describe the characteristics that these sports impose on athletes. Hughes and Bartlett [1] describe racquet sports as net and wall games that rely on points rather than time. Because of the different rules for the number of serves allowed, the number of rebounds allowed, and the number of volleys allowed, the player's racket sport will have unique metrics. However, common factors associated with all net and wall games were identified as serve, stroke, winner, and length of rallies [1].

In recent years, with the development of sports science and the commercialization of racket sports, attention has been focused on improving the performance of racket sports, which has led to a more detailed study and understanding of all aspects of racket sports. In addition, these sports also have many differences in scientific disciplines such as exercise physiology, nutrition, performance analysis, biomechanics, medicine, engineering, psychology, motor skills, and injury [2]. The pattern of injuries reflects changes in training, a more aggressive style of play, changes in grip, the open stance of the forehand, and the improved physical ability of the players. [3]

Sports injuries account for 10-19% of all serious injuries treated in emergency rooms, and the most common type of injury is knee and ankle injuries [4]. In the context of common injuries in racket sports, studies have shown that in tennis [5], badminton [3], squash [3], and table tennis, sports injuries often occur in knees, feet, wrists, ankles, and feet.

Racquet sports have many possibilities for occurrence and overuse injuries. In tennis, two-thirds of injuries are due to overuse and one-third are due to traumatic injuries or acute events [6]. In squash, most squash injuries occur in the lower extremities, accounting for 32% to 58% of all injuries, with the ankle being the most common injury site, accounting for 13% to 16% of all squash injuries, followed by the knee, accounting for 7% to 9% [7]. In badminton, Goh et al. [8] reported that one-third of injuries occurred in lower limbs, they suggested that badminton injuries could occur in many musculoskeletal areas, but were most likely to occur in especially the knee joint, the lower back, and the ankle joint injuries. In addition, tendinitis occurs primarily in the knee or ankle, possibly due to repeated stress. Previous studies [9, 10, 11, 12] have reported that tennis injuries can occur in many musculoskeletal areas, with the most likely areas occurring in the ankles, lower back, and shoulders. Ankle sprains are the most common injury in tennis and other racquet sports. In squash, sports injuries can occur in many musculoskeletal areas, but they occur most intensively in the knees, neck, and shoulders. This evidence reaffirms previous speculation that most injuries in racquet sports are caused by chronic overuse.

To sum up, in badminton, tennis, squash, table tennis, and other racket sports, the occurrence of knee sprain/strain has a common character and is one of the most common injuries. Tendinitis, on the other hand, occurs primarily in the knee or ankle, possibly due to repeated stress, confirming previous speculation that most injuries in racquet sports are caused by chronic overuse. From the above data distributed by anatomical region in racquet sports, it can be seen that the most common injuries in racquet sports are relatively related to the lower extremities.

1.2 Biomechanics of Topspin Forehand

Topspin forehand is known as one of the most basic and aggressive strokes in table tennis. Especially for an elite offensive player, excellent forehand topspin skill is necessary to maintain a strong attacking posture [13]. Some clinical experimental studies have collected kinematic and dynamic information about players hitting the forehand topspin through 3D (3 dimensions, 3D) motion capture systems, such as infrared cameras and high-speed cameras. The whole-body coordination mechanism is very important in table tennis, and the performance level of the upper limbs is largely determined by the lower limbs [14, 15]. In recent years, the important role of lower limb function in table tennis has been widely studied and reported [16 - 22]. As the origin of the kinetics chain, perfect lower limb movement performance will benefit the velocity of the racket and ball [14, 23, 24]. Although several studies investigated biomechanical information or highlighted the lower limb during topspin forehand, their experimental design, protocol process, and methods were generally inconsistent. Meanwhile, the common characteristics of elite athletes with the same skill and playing mode can reflect the internal mechanisms of sports at different levels and the technical characteristics. Therefore, to optimize topspin training items and provide guidance information, it is necessary to explore the common lower limb biomechanical characteristics of high-level athletes during topspin forehand strokes.

The original characteristics of each included study can be seen in Figure 1. The parameters of joint kinematics were the most focused which included 17 studies. The racket and plantar information were also focused which included 5 and 4 studies, respectively. The percentage of included studies from China and Poland was 53% and 26%, as well as Japan, Italian, and France were 11%, 5%, and 5%, respectively. There are 47% of the studies' sample sizes in the 10 to 15, and the total sample size of included studies was 263, a total of 111 players' performance levels belonging to the national I. The stroke task, footwork, and performance level were the most concerned maneuver settings which included a total of 7, 6, and 5 studies, respectively.

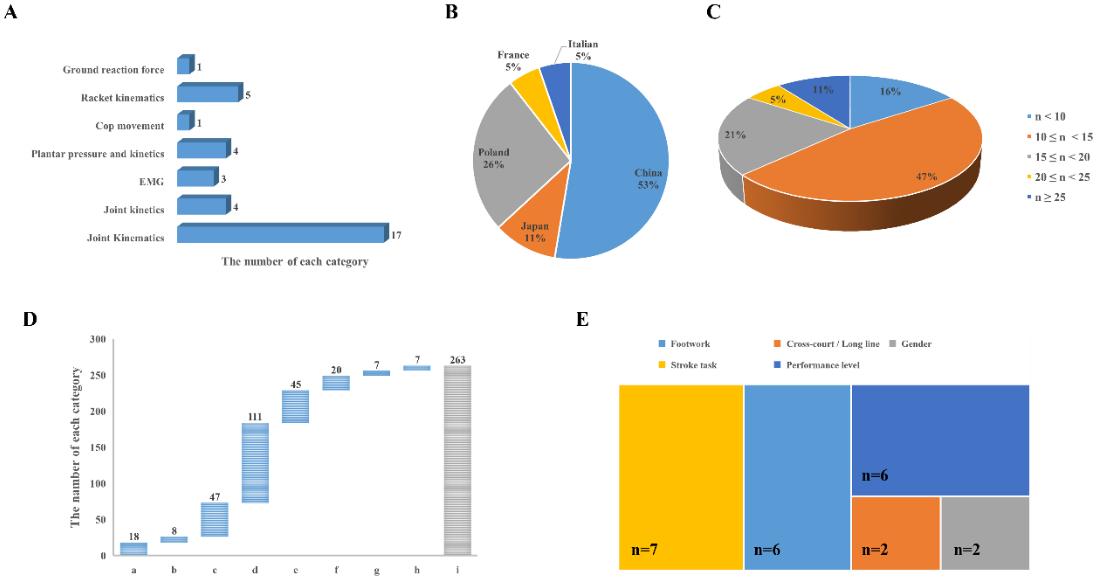

Figure 1. Characteristic information of the research about topspin forehand.

Note: (A) parameters; (B) country of included studies; (C) sample size; (D) performance level of player, “a” refers to university level, “b” refers to national III, “c” refers to national II, “d” refers to national I, “e” refers to the national team level, “f” refers to national top 16, “g” refers to national top 10, “h” refers to national top 200, “I” refers to total; (E) maneuvers type.

1.2.1 Gender in the Topspin Forehand

Gender differences are reflected in many sports. McLean et al. found significant sex effects in knee, hip, and ankle joint kinematics by comparing male and female basketball athletes, and female basketball athletes were at higher risk of anterior cruciate ligament injury [25]. Other researchers suggested that there may be different biomechanical loading mechanisms between males and females by finding that the trunk and pelvic kinematics of young rowers were different during rowing [26], which was also found in young runners.

Gender differences exist in the morphological structure of table tennis players [27]. Zagatto et al. (2016) [28] revealed that fat-free mass, fat mass, and body fat percentage values of male table tennis players were higher than female players. Male players are superior to female players in dynamic posture control during multi-ball table tennis training [29]. According to [30], the male topspin stroke pattern allows for greater use of large muscle groups and joints than for females. Females tend to attack topspin strokes from both sides of the forehand and backhand, while males tend to look for opportunities to hit more powerful topspin strokes from the forehand. These differences may be due to morphological gender variations. By outlining the movement patterns of topspin stroke, differences in male and female contributions to thoracic rotation, external shoulder rotation, dorsal flexion, and supination in the wrist were revealed during the stroke stage.

Two articles reported on the gender differences in forehand topspin, one of which explored the effect of gender on lower limb joints' biomechanical characteristics during the forehand topspin stroke with chasse step footwork. Compared with female athletes, male athletes performed significantly greater movements in the lower limb joints, such as the extension and flexion of the hip, trunk, and knee joints [18, 30]. However, In the backward phase, female athletes showed a significantly greater hip abduction than male athletes. The maximal acceleration of the playing hand of the male athletes was significantly greater than the female athletes during the topspin forehand [30].

1.2.2 Performance Level in the Topspin Forehand

The topspin forehand loop was used frequently in table tennis winners compared with other types of strokes, and it probably shows that mastery of this shot would have a critical effect on winning matches [31]. With the competition of table tennis becoming more and more intense, performing a better topspin forehand loop may influence the result of competitions. Mastering topspin forehand strokes could distinguish the performance of different level players even though it is difficult [13].

As one of the most offensive table tennis strokes, the topspin forehand is extremely important for an aggressive player to master this shot properly [13]. Iino et al. (2009) reported the biomechanical analysis during the topspin forehand loop against backspin balls between different level table tennis players. They reported values for the kinematic analysis of lower trunk flexion, rotation, and extension. Two group male table tennis players including nine EA (elite athletes, EA) and eight MA (medium athletes, MA) hit topspin forehands against heavy and light backspins. However, the ankle, hip, and knee kinematics in the frontal, sagittal, and transverse planes were not investigated in their research.

Lower trunk axial rotation of EA contributed more to racket speed during the topspin forehand [13]. Lower limb joint angles, joint velocity, and range of motion were greater in the sagittal and horizontal planes in EA, such as the hip [17, 27], knee [27], and ankle [16] joints. In addition, the movement characteristics of COP (center of pressure, COP) were also significantly different among athletes of different levels. Compared with MA, the EA showed greater medial-lateral COP displacement during the backward phase, but less anterior-posterior displacement throughout the stroke process [32]. In addition, the EA has a larger plantar contact area than the MA [17]. Results indicated that EA possessed better foot drive skills and the ability of foot movement control during the topspin forehand.

1.2.3 Muscle Force and sEMG (surface electromyography, sEMG) in the Topspin Forehand

Three articles addressed sEMG information during the topspin forehand, two were based on table tennis footwork, and two were based on performance level differences. Lower limb muscle activity levels were significantly higher during forehand topspin compared with other types of strokes [21]. Hip, knee, and ankle flexion muscle groups such as the biceps femoris, gluteus maximus, rectus femoris, gastrocnemius, and soleus were thoroughly activated during high-intensity topspin forehand strokes [21, 27].

1.2.4 Footwork in the Topspin Forehand

In table tennis competitions, athletes have to play good strategies to win the match [33]. Footwork is one of the core skills that table tennis players need to master. Athletes have to return to the ready position for the next movement during the match. Good footwork plays an important role in balancing dynamic stability and agility [23]. Players perform large amounts of active running to ensure that they can reach the most suitable hitting position prior to playing the next stroke [34]; this positive behavior can provide sufficient preparation time for playing the next stroke. There is a strong link between stroke, type of footwork, and different types of strokes that may be combined with specific types of footwork [34]. Therefore, footwork is not only the basis but also one of the key points of table tennis training. The chasse step and one step are the basic footwork patterns that combine with forehand and backhand strokes in table tennis [15, 35, 36, 37]. In addition, proficient mastery of footwork can bring advantages to energy transfer in the power chain of the lower extremities. Therefore, the study of biomechanics in table tennis footwork is an interesting field for athletes and scientists. The chasse step is a footwork movement that is used in combination with racket play to perform a set of defensive and offensive strokes by making easy side movements. The one step is a footwork movement which allows the player to move for relatively long distances in the shortest time possible.

Biomechanical research on footwork in table tennis has received a lot of attention in recent years. A total of 6 articles explored the biomechanical characteristics of footwork during the topspin forehand. The lower limb biomechanics of cross-step, chasse-step, and one-step footwork seem to have received more attention and research. The chasse step footwork is a side movement that could combine with racket movement to perform offensive and defensive strokes in table tennis [22, 38]. Comparing the long-distance chasse step footwork with the short-distance chasse step footwork, the ankle joint ROM (range of motion, ROM) and angular velocity in the coronal and transverse planes of the long-distance chasse step footwork were significantly faster than the short-distance chasse step footwork during the topspin forehand [38]. The maximal knee flexion and ankle inversion angular velocity of the cross-step footwork were significantly greater than the chasse step footwork during the topspin forehand [15]. The joint angles and ROM of the hip, knee, and ankle joints of the one-step footwork were significantly smaller than those of the cross-step and chasse-step footwork [15]. Gender and level factors were also important in relation to research content in footwork biomechanics during the topspin forehand. In the foreword phase, hip angular velocity and ROM in the male athletes were significantly greater than the female athletes [18]. Compared with MA, the EA showed significantly greater flexion velocity in the hip and knee during cross-step footwork, as well as significant hip and knee moments during a fast topspin forehand using the cross-step footwork [27].

1.2.5 Plantar Biomechanics in the Topspin Forehand

Several studies have investigated the biomechanical characteristics of plantar during stroke play in table tennis. Lam et al. (2018) [15] investigated the biomechanical information of ground reaction forces, plantar pressure, and joint kinetics distribution during topspin forehand under three typical footwork conditions. Qian et al. (2016) [17] have identified significant differences in in-shoe plantar pressure between different level table tennis players. One possible explanation for the differences observed is the synergy that exists between the torso and lower extremities during the entire stroke motion [97]. Shao et al. (2020) [19] investigated the kinetics and kinematics differences between professional and novice athletes during one-step footwork based on the Oxford foot model. In addition, the effect of foot performance during stroke-play has been demonstrated in previous studies.

A total of 4 articles explored the mechanical characteristics of the plantar during the topspin forehand. Plantar mechanics are related to lower limb drivability. In addition to the movement of COP [32], indexes such as pressure in various plantar regions [15, 22, 39], plantar force [22], contact area [17], force-time integral [22], and pressure-time integral [15, 22] have been successively studied and reported. Overall, the differences in plantar mechanical characteristics were concentrated in the first metatarsal, the medial-lateral of the forefoot, and the medial-lateral of the rearfoot [15, 17, 22, 39]. The peak pressure in the total foot and toe regions of the cross-step and chasse-step footwork was significantly greater than that in the one-step footwork [15]. The peak pressure in the total foot and first metatarsal regions was significantly greater in the cross-step than in the chasse step footwork [15, 22]. Chasse step footwork showed significantly greater plantar force, force-time integral, and pressure-time integral than one-step footwork in both backward and forward phases. In the foreword phase, the peak pressure in the toe region of the chasse step footwork was significantly greater than that of the one-step footwork [15, 22]. Differences in performance levels also led to differences in plantar mechanics, with EA exhibiting higher peak pressures in the medial-lateral forefoot region and the medial-lateral rearfoot region when performing chasse step footwork [39]. In addition, EA has significantly larger plantar contact areas during topspin forehand [17].

1.2.6 Relationship between Lower Limb Joints and Racket in Topspin Forehand

Five articles reported the relationship between lower limb joints' biomechanical characteristics and racket movement during the topspin forehand. The influence of the human joint movement on racket movement has always been the main content of biomechanical research on the topspin forehand. The maximum speed of the racket is increased through the human kinetic chain effect, which brings benefits to enhancing the rotation and aggression of the ball [15, 17, 20, 27]. In general, racket velocity is related to the angular velocity of axial motion of the hip, pelvis, and ankle joints. Specifically, the flexion angular velocity of the hip joint on the playing side the extension angular velocity of the other side [40, 41], and the plantar flexion angular velocity of the ankle joint during the topspin forehand [16]. The peak velocity of pelvic axial rotation and the work carried out by the pelvic axial rotation torque on the playing side have a positive correlation with the horizontal velocity of the racket at impact during the topspin forehand [41].

1.3 Biomechanics Exploration in Balance

Balance refers to the ability of the body to keep the center of gravity and body posture stable without the help of external forces, automatically maintain movement and resist external interference, and maintain the body's center of gravity on the support surface. The classification of balance ability includes three main aspects: standing balance, sitting balance, and moving balance. There are many kinds of classification methods for balance, and the more common methods are dichotomy and trichotomy. According to the classification of the binary method, balance is divided into two categories: one is to divide balance into predictive balance control and active or compensatory balance control; the other is to divide balance into static balance and dynamic balance, and dynamic balance is divided into self-dynamic balance and other dynamic balance. Self-homeostasis refers to the ability of the human body to regain a stable state when it performs autonomous movement, and homeostasis refers to the ability of the human body to regain a stable state when it is subjected to external interference. The method of thirds divides the equilibrium into three categories: stability equilibrium ability, symmetry equilibrium ability, and dynamic stability. In the aspect of sports biomechanics, balance ability is usually divided into static balance ability and dynamic balance ability in order to better study the factors that affect balance ability. Static balance ability refers to the ability of the human body to maintain a posture or stable state by adjusting its center of gravity in a relatively static state. The ability of dynamic balance refers to the ability of the human body to automatically adjust and maintain the body posture and control the body balance when it is in motion or under the action of external forces.

The human balance system is an extremely complex closed-loop regulation system, whose physiological structure includes the sensory system, nervous system, motor system, etc. The human body orientates the spatial information and proprioception of the human body through the sensory system, and the information is transmitted to the nervous system for analysis and then issues regulatory instructions, and the motor system executes regulatory actions accordingly to maintain human balance. In competitive sports and daily life, balance ability plays an important role and is one of the basic elements of human sports. In complex sports with high technical requirements, such as figure skating, gymnastics, cross-country skiing, snowboarding, and other sports, athletes need to have good posture control ability to complete a series of difficult movements in an unfavorable environment. Therefore, athletes in figure skating, cross-country skiing, gymnastics, and other events generally need to undergo special professional and technical movement training to improve their balance ability [42]. Balance ability training, as a classic core stability training method, can enhance the control of nerves over muscles and raise energy. Thus, the purpose of maintaining and regulating the balance of the human body is achieved by activating and controlling the muscles that can maintain the stability of the human body

1.3.1 Factors Affecting Balance

In daily life, due to the complex diversity of human activities, it is a very complicated process to maintain the balance of the human body in different states. In addition, according to the physiological basis of balance, the balance system is composed of the spiritual system, the sensory system, and the motor system, and the physiological conditions of each component will affect the balance ability, so there are many physiological factors that can lead to the decline of balance ability. For example, decreased visual ability [43], vestibular organ damage [44], nervous system fatigue, and muscle strength decline [45] will all lead to decreased balance ability. A large number of studies have shown that muscle fatigue, as an important inducing factor leading to sports injury, is also one of the indirect factors leading to the reduction of balance ability. Suscod [46] et al. pointed out that the balance problem often troubled the injured athletes, and the injury caused by muscle fatigue indirectly led to the imbalance of the athletes. The study of Rose [47] et al. reported that during the gradual recovery of injured athletes, the balance ability was improved along with the recovery of muscle strength and endurance. In recent years, scholars at home and abroad have paid extensive attention to and studied whether muscle fatigue around joints will reduce the body's balance ability and proprioception.

1.3.2 Evaluation Method of Static Balance

Balance is the ability to keep the body's center of gravity above the supporting base plane. The body's center of gravity changes with posture and body movement. When the human body is standing still, the body essentially has been around its own equilibrium point in a constant state of shaking, and the human subjective consciousness cannot control this shaking, in physiology, this phenomenon is called physiological posture wavering. Since 1980, the international began to use the pressure plate method to record the continuous change of the human center of gravity in the horizontal plane trajectory chart, and then gradually developed a computerized balance tester. By standing the subject on a stationary biomechanical platform or plate, the platform's highly sensitive force sensor can be used to record the subject's body sway, and after a series of analysis software processing, can calculate the static balance evaluation parameters to evaluate the human balance. Generally, the evaluation parameters of such static balance include the position of the center of gravity of the subject's body, the center of gravity moving path or the area of the region, the total length of the center of gravity moving path, and the ratio of the center of gravity parameters of the subject during the measurement with eyes closed and the measurement with eyes open respectively [48]. Winter [49] pointed out that COP can be approximately equal to the center of gravity of the body when the human body is in a static or slow-moving state. Therefore, in this study, COP-related parameters were used as indicators to assess patients' balance ability.

1.3.3 Evaluation Method of Dynamic Balance

There are many kinds of evaluation methods for dynamic balance ability, including observation method, scale evaluation method, and balance test instrument evaluation method.

The observation method is mainly used for rough screening of patients with balance dysfunction to assess whether the subjects can maintain balance under active conditions, such as the Mann test and one-foot upright test.

Due to its high reliability and good validity, the scale evaluation method has become a more common method in clinical applications. Generally, more scales are used, including the Berg balance scale, the "stand up - walk" timing test, and Tinetti scale, and functional extension scales, such as the Y balance scale and star translation balance ability test scale.

The dynamic Balance test instruments commonly used in the world mainly include Balance Master, Equitest, and Balance Performance. However, due to the high price, it has limited its development and use in China to a certain extent. The dynamic balance of the human body mainly depends on the ankle joint, and the rapid balance response ability of the human body needs to rely on the participation of various sensory systems of the human body. The test content of dynamic balance ability through the instrument mainly includes stability test, motion control test, strain ability test, and sensory integration test.

1.4 Summary

Based on the introduction, the research about rackets sports injury, topspin forehand, and balance was wildly focused. However there are some unanswered questions that need to be raised in this field of research. Firstly, in the existing biomechanical studies on topspin forehand skills in table tennis, there is little research and motion capture of the ball movement. Combining human motion with ball motion to explore the inner connection between human movement and ball movement can further elaborate and develop the depth and breadth of research in this field. Secondly, the current research also lacks reports on the collision effect between the ball and the racket, which is an important section of research. Thirdly, from the perspective of sports injuries, it is also interesting and valuable to use musculoskeletal modeling to further explore the biomechanical characteristics of joints, such as calculating joint reaction force and loads through OpenSim. Therefore, the biomechanical exploration of lower extremity injury mechanism during table tennis topspin forehand and the implication of skills optimization and motor control is necessary.

2 Material and Methods

2.1 Ethics Statement

This dissertation was performed in compliance with the declaration of Helsinki and was approved by the Institutional Ethics Committee of Ningbo University. All subjects were informed of the corresponding experiment and simulation content as well as the potential risks before taking part in this study, and then they gave their consent to participate. The informed consent form is shown in Figure 2.

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The application that you submitted to this office in regard to the use of human subjects in the proposal referenced above has been reviewed by the Secretary, the Chair, and five members of the **Ethics Committee of Ningbo University.**

In addition: the research has been approved for the inclusion of human subjects, for waiver of the consent process (for the one being assessed), for a waiver of the assent process (for the one being assessed), for a waiver of the parental permission process (for the one being assessed) and for a waiver of the **Health Insurance Portability and Accountability Act (HIPAA) Research Authorization** (entire research study for the one being assessed).

Date of IRB Approval: **August, 22, 2021**
Date of IRB Approval Expiration: **August, 21, 2023**
Approval Number: **RAGH202108223005.7**

If applicable, informed consent (and **HIPAA Research Authorization**) must be obtained from subjects or their legally authorized representatives and documented prior to research involvement. The **IRB**-approved consent form and process must be used. Changes in the research (e.g., recruitment procedures, advertisements, enrollment numbers, etc.) or informed consent process must be approved by the **IRB** before they are implemented (except where necessary to eliminate apparent immediate hazards to subjects).

This approval is valid for **two years** from the date of **IRB** review when approval is granted or modifications are required. The approval will no longer be in effect on the date listed above as the **IRB** expiration date. A continuing review application must be approved within this interval to avoid expiration of **IRB** approval and cessation of activities. A final report must be provided to the **IRB** and all records relating to the research (including signed consent forms) must to retained and available for audit for at least 3 years after the research has ended.

It is the responsibility of all investigators and research staff to promptly report to the **IRB** any serous, unexpected and related adverse events and potential unanticipated problems involving risks to subjects or others.

This approval is issued under

Jianshe Li
Chairman of the Ethics Committee of Ningbo University

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您提交给本办公室的提案申请"**The characteristics of Lower Limb Muscle Forces in Table Tennis Footwork During Stroke: A pilot study Based on The OpenSim Musculoskeletal Model**"已经由**宁波大学伦理委员会**的秘书、主席和两名成员审查。

此外：该研究已被批准纳入人类受试者，豁免同意过程（对于被评估的受试者），豁免父母许可程序（对于被评估者）和豁免健康保险便携性和责任法案（HIPAA）研究授权（对于被评估者的整个研究）。

伦理审查委员会批准日期： **2021 年 8 月 22 日**
伦理审批截止日期： **2022 年 8 月 22 日**
批准文号： **RAGH202108223005.7**

如果适用，在参与研究前必须获得受试者或其合法授权代表的知情同意（以及 **HIPAA 研究授权**）并记录在案。必须使用经**伦理委员会**批准的同意书和程序。研究的变更（例如，招募程序、广告、登记号码等）或知情同意程序在实施前必须得到**伦理委员会**的批准（除非有必要消除对受试者明显的直接危害）。

该批准有效期为两年，自**伦理委员会**审查后批准或进行必要修改的日期开始。在超过上述**伦理审批截止日期后**，该批准将不再有效。持续审查申请必须在此期间内获得批准，以避免**伦理委员会**审批到期和活动停止。必须向**伦理委员会**提交最终报告，所有与研究有关的记录（包括签署的同意书）必须保留，并在研究结束后至少 3 年内供审核。

所有研究者和研究人员有责任及时向**伦理委员会**报告任何严重的、意外的和相关的不良事件和潜在的涉及受试者或其他人风险的未预料到的问题。

这项申请批准于：2021年8月22日.

李建设
宁波大学伦理委员会主席

Research Academy of Grand Health, Ningbo University
Human Informed Consent Form

Instructions to the Student Researcher(s): An informed consent/assent/permission form should be developed in consultation with the Adult Sponsor, Designated Supervisor or Qualified Scientist.
This form is used to provide information to the research participant (or parent/guardian) and to document written informed consent, minor assent, and/or parental permission.

- When written documentation is required, the researcher keeps the original, signed form.
- Students may use this sample form or may copy ALL elements of it into a new document.

If the form is serving to document parental permission, a copy of any survey or questionnaire must be attached.

Student Researcher(s): Yuqi He
Title of Research: The characteristics of lower limb muscle force in table tennis footwork during stroke: a pilot study based on the opensim musculoskeletal model
I am asking for your voluntary participation in my science research. Please read the following information about the research.
If you would like to participate, please sign in the appropriate area below.
Purpose of the research:

If you participate, you will be asked to:

Time required for participation:

Potential Risks of Study:

Benefits:

How confidentiality will be maintained:

If you have any questions about this study, feel free to contact:

Adult Sponsor/QS/DS: ______________________ Phone/email: ______________

Voluntary Participation:
Participation in this study is completely voluntary. If you decide not to participate there will not be any negative consequences. Please be aware that if you decide to participate, you may stop participating at any time and you may decide not to answer any specific question.
By signing this form I am attesting that I have read and understand the information above and I freely give my consent/assent to participate or permission for my child to participate.

Adult Informed Consent or Minor Assent Date Reviewed & Signed: ______________

Research Participant Printed Name: Signature:

Parental/Guardian Permission (if applicable) Date Reviewed & Signed: ______________

Parent/Guardian Printed Name: Signature:

Figure 2. Human informed consent form.

2.2 Participants

2.2.1 Explore the Kinematics of the Lower Extremity during Topspin Forehand

Ten professional male table tennis athletes were allowed to participate in this study. All of the participants were attached to the table tennis team at Ningbo University, Ningbo China. The group of participants were divided into two groups: Five players (Height: 173 ± 4.2 cm, Weight: 70 ± 7.9 kg, Experience: 10 ± 3 years) belonged to EA which also plays in the China National Level I. Another five players (Height: 172 ± 2.7 cm, Weight: 69 ± 8.5 kg, Experience: 9 ± 3 years) belong to MA who play in China National Level II. All participants were right-handed and nobody with previous diseases or deformities of the lower limb for 3 months before this study. The handedness of the athletes was identified and confirmed based on the preferential hand used to hold the racket. Caffeine was forbidden to ingestion of all participants for 4 hours before this study. Before the commencement of this research, participants were provided to write informed consent. This research was approved by the Ethics Committee of the Research Academy of Grand Health at Ningbo University (RAGH20191121).

2.2.2 Explore the Plantar Force and Pressure during Topspin Forehand

Twelve national level 1 table tennis players (Height: 172 ± 3.80 cm, Weight: 69 ± 6.22 kg, Age: 22 ± 1.66 years, Experience: 11 ± 1.71 years) from Ningbo University volunteered to participate in the study. All participants were free of any form of lower extremity injury or disease within 6 months before data collection. All participants were right-handed, had a dominant right leg, and were in good physical health. The Human Ethics Committee of Ningbo University approved the study (RAGH20200901). All participants received and signed written informed consent after being informed of the objectives, details, requirements, and procedures of the table tennis experiment.

2.2.3 Explore the Balance Recovery

Twelve table tennis players (Height: 175.17 ± 4.99 cm, Weight: 66.96 ± 4.44 kg, Age: 23 ± 1.65 years, Leg length: 90.79 ± 1.86 cm, Foot length: 267.78 ± 5.04 mm, Foot width: 102.07 ± 5.07 mm) volunteered to participate in this study, all participants belonged to the national level-one. Participants with no lower limb muscle and joint sports injuries within 3 months before the experiment. In addition, the experimenter will fully inform the participants of the possible risks and requirements of the experiment to ensure that the participants are physically and mentally able to withstand the cryotherapy experiment. All participants were asked to avoid any moderate to vigorous physical activity and to follow a routine (no alcohol, caffeine, and insomnia) two days before the study began.

2.2.4 Explore the Muscle Force of the Lower Extremity during Topspin Forehand

Six male national-level table tennis athletes (height: 171.98 ± 4.97 cm; weight: 68.77 ± 7.86 kg; experience: 10.67 ± 1.86 years; age: 22.50 ± 1.64 years) from Ningbo University were recruited to participate in this study. All subjects were free from any neuromuscular injury within 6 months, while all subjects were right-handed. Before the start of the formal experiment, all subjects were fully informed of the purpose, process, and requirements of the study, and all subjects provided informed consent. The Ethics Committee of Ningbo University approved this study (RAGH202108223005.7).

2.3 Biomechanical Experiments

2.3.1 Experimental Protocol, Procedures, and Experimental Instrument

(1) Explore the Kinematics of the Lower Extremity during Topspin Forehand

As outlined in Figure 3. Data were collected as previously described [17]. The kinematic information of the participant was captured by an 8-camera Vicon motion analysis system (Oxford Metrics Ltd., Oxford, UK) at a 100 Hz frequency. A total of 16 reflective markers (diameter: 14 mm) were attached with adhesive tape on the bilateral lower limbs respectively. Marker locations included: posterior-superior iliac spine, anterior superior iliac spine, lateral mid-thigh, lateral knee, lateral malleolus, lateral mid-shank, second metatarsal head, and calcaneus.

Players were asked to perform only the forehand topspin with maximal power to return the topspin ball played by a professional table tennis coach to the target area (A: L1 = 45 cm, L2 = 38 cm). The size set of the target area, is based on own research. was considered to limit and standardize the quality of the stroke play [20]. If the ball is missed or outside of the target area (A), the action will not be measured, and participants need to do it again. The coach was asked to keep a stable ball track and drop point, the speed and frequency of the ball were also controlled by the coach. The action of the players was recorded until three full motions were successfully captured. And participant was allowed adjustment or rest 1 min between actions. In addition, all players were given at least 15 min to warm up and adaption to the experimental environment before the commencement of the official experiment and data collection.

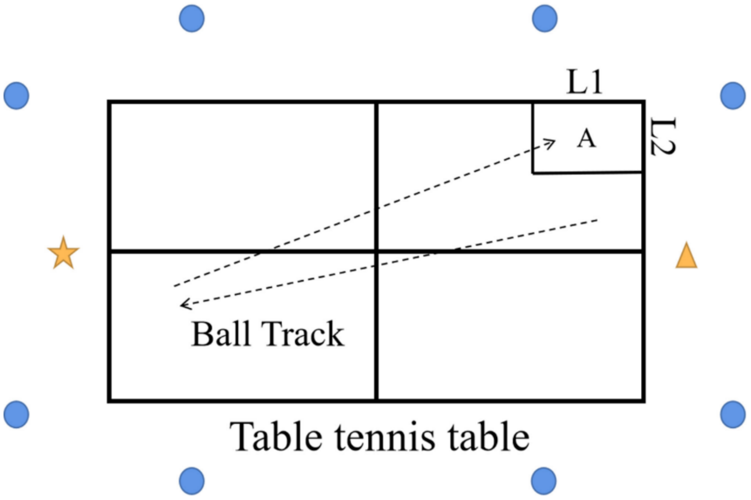

Figure 3. Experiment setup.

Note: L1: 45 cm, L2: 38 cm, A: Target area, ☆: Participant, △: Coach, ○: Camera.

Segmental Coordinate Systems

The segmental coordinate systems were created as previously described [13].

Lower trunk: The vector from the middle point on both sides of the hip joint centers to the middle point on both sides of the shoulder joint centers was defined as the z-axis of the lower trunk coordinate system. The x-axis was defined as the cross product of the vector from the center of the right hip to the center of the left hip with the z-axis. The cross-product of the z-axis and x-axis was defined as the y-axis.

Driving leg: The z-axis of the driving leg (right leg) was the same as the lower trunk z-axis. The y-axis was the cross product of the z-axis of the system and a vector from the ankle joint center to the knee joint center. The cross product of the y- and z-axis was defined as the x-axis.

Ankle: The z-axis of the driving leg (right leg) was the same as the lower trunk z-axis. The y-axis was the cross product of the z-axis of the system and a vector from the ankle joint center to the knee joint center. The cross product of the y- and z-axis was defined as the x-axis.

Experimental Material

The location of the experiment was Ningbo University Table Tennis Training Centre, which is a professional competition and training facility. During the experiment, players were asked to wear professional table tennis match shoes. Besides, all players were asked to use the same table tennis racket (Timoboll-zlc; Butterfly Technical Center, Tokyo, Japan) with the Butterfly Tenergy 05 Max (Butterfly Technical Center, Tokyo, Japan) and DHC Hurricane 3 (Double Happiness Sports Company, Shanghai, China) rubber sheets. The DHC Hurricane 3 rubber was the forehand side one. The playing table used for data capture (Rainbow; Double Happiness Sports Company, Shanghai, China) was a professional game table.

Definition of Motion Phase

As outlined in Figure 4. Motion phase A was defined as an NP (natural position, NP). Figures 2A–2C were defined as the BP (backswing phase, BP), Figs. 2D–2F was defined as the FP (forward-swing phase, FP). Besides, this research focuses on the key event of the entire motion, so we defined position C as the key event which meant the EB (end of backswing, EB). The position F was defined as the key event which means the EF (end of forward-swing, EF).

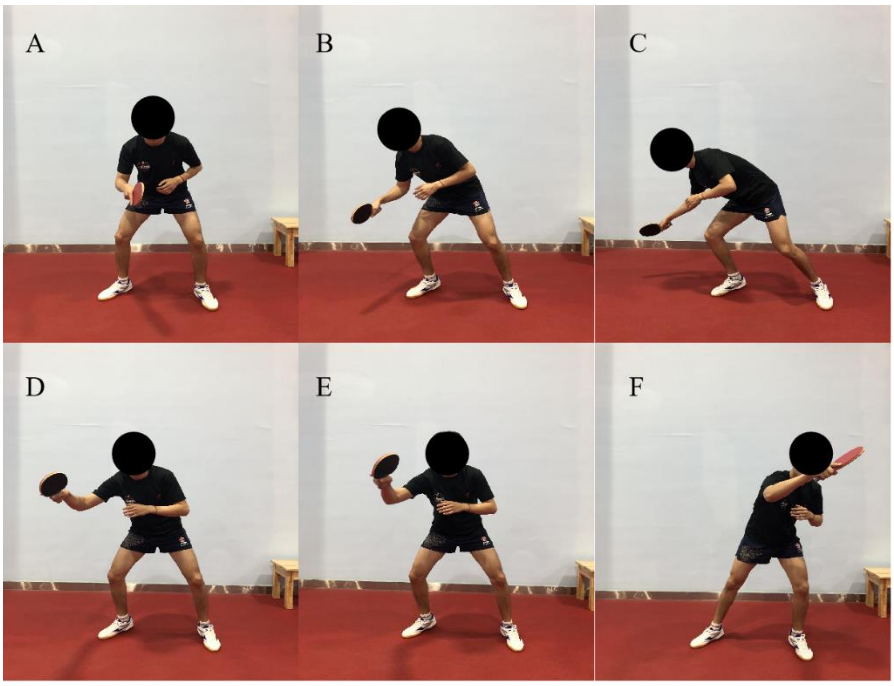

Figure 4. The divide and definition of motion phase.

Note: Motion phase A was defined as a natural position (NP), (A–C) was defined as the backswing phase (BP), and (D–F) was defined as the forward swing phase (FP). Position C was defined as the key event which meant the end of the backswing (EB). Position F was defined as the key event which meant the end of the forward-swing (EF).

(2) Explore the Plantar Force and Pressure during Topspin Forehand

Experimental Design

The experiment was performed at the Ningbo University table tennis training gymnasium. As outlined in Figure 5, the kinetic data of the right leg was recorded using a Novel Pedar insole plantar pressure measurement system (Novel GmbH, Munich, Germany, sampling frequency of 100 Hz) and a force platform (AMTI, Watertown, United States, sampling frequency of 1,000 Hz). The table tennis table, balls, and rackets used complied with international standards. Before the start of the formal experiment, subjects were provided with time to warm up and familiarize themselves with experimental procedures. The warmup details included jogging on a treadmill at a comfortable speed and stretching. In the formal experiment, participants were asked to return the coach’s shot to the target area using chasse steps and one step, respectively. The hitting methodology for this experiment was as follows: the coach was asked to serve to the impact zone, which was in the centerline, and then serve to the impact zone which in the right side of the table tennis table. The player then needed to use the chasse step and one-step footwork to return the ball to the target area. Participants were asked to complete four successful strokes using chasse step footwork in the first instance, then complete four further successful strokes using one-step footwork. The smoothness of the movement was judged by the players themselves, and the quality and effect of the ball play were supervised by a qualified table tennis coach.

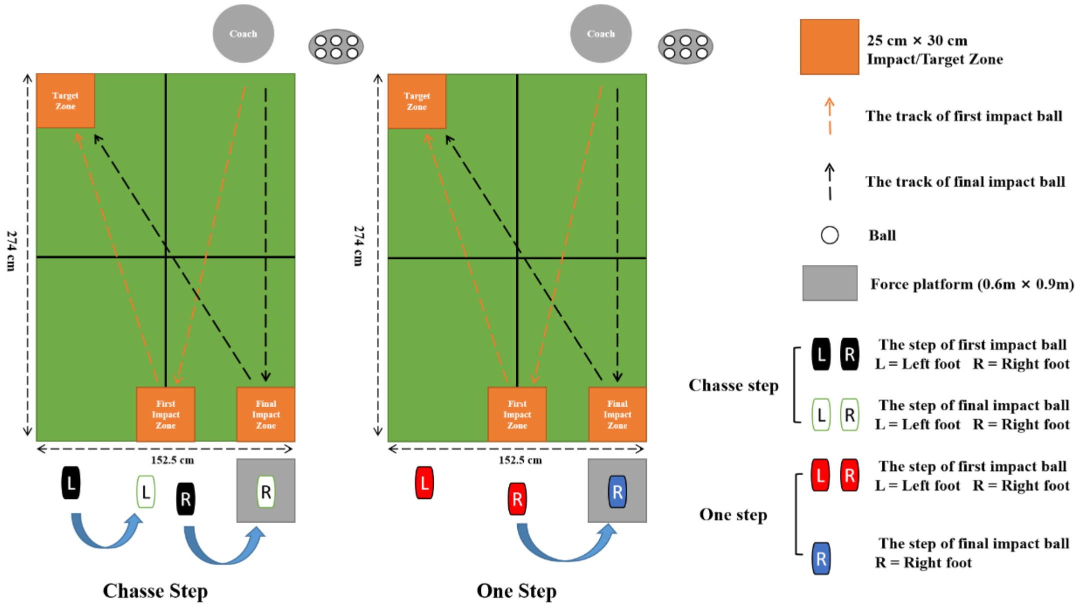

Figure 5. The experimental setting.

Note: The left side shows the chasse step, and the right side shows the one step.

Data Collection and Processing

Information for plantar force, maximum plantar force, peak pressure of each plantar region, FTI (force-time integral, FTI), and PTI (pressure time integral, PTI) were recorded by the Novel Pedar insole plantar pressure measurement system (Novel GmbH, Munich, Germany, sampling frequency of 100 Hz). The plantar was divided into six areas: T (Toe, T), MF (Medial forefoot, MF), LF (Lateral forefoot, LF), M (Midfoot, M), MR (Medial rearfoot, MR), and LR (Lateral rearfoot, LR). The data was then exported into MATLAB R2019a (The MathWorks, United States), and a written script was produced to process the data. The participants remained in the ready position on the left side of the table, and the data collection started 1s before the coach served. The coach served after hearing the “start” command, the participant was asked to hit the ball with maximum force to the target zone. And data collection stopped after the participants completed the stroke action. As outlined in Figure 6, to collect data closer to the real situation, in a data collection task, the coach will execute two serves, and the subjects are asked to complete two consecutive strokes. After completing the first stroke, the subjects were asked to complete the second stroke in combination with footwork.

The footwork of the second stroke was asked to fully step on the force platform. And only the footwork of the second stroke was considered and analyzed. The contact period of the right leg of the selected footwork, from initial contact to take-off, is determined from the data provided by the force platform. As right leg movements were responsible for forgiving the greatest contribution to the forehand stroke [15]. When the value of the ground reaction force reaches 10N for the first time, it is defined as the contact moment, and when the value of the ground reaction force decreases to 10N for the first time, it is defined as the airborne moment [15]. By collecting the ground reaction of the right leg during forehand stroke motion through the force platform, two peaks can be observed. The first peak is the peak time of the right leg landing phase, and the second peak is the peak time of the right leg kicking phase. Based on the kinetics information of the force platform, the movement stages of the footwork movement were divided. The phase of ground reaction force from the 10N to the trough was defined as the BP (As shown in Figure 6E–6F). The phase of ground reaction force from the trough to below 10 N was defined as the FP (As shown in Figure 6G–6H).

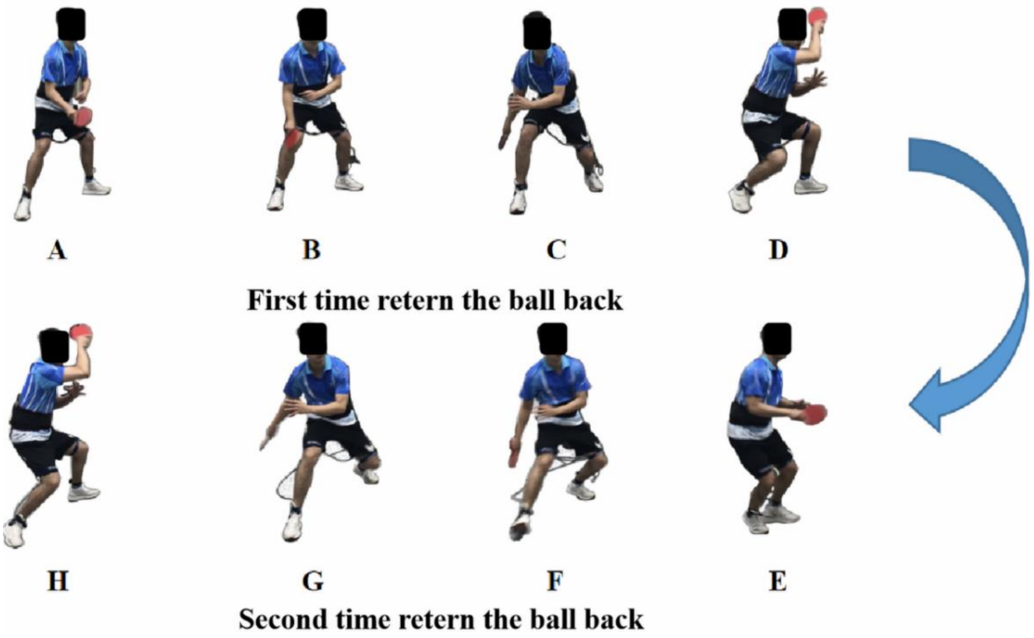

Figure 6. The technical performance of a participant during the test.

Note: (A–C) The backward phase of the first hit process. (C–D) The forward phase of the first hit process. (E–F) The backward phase of the second hit process. (G–H) The forward phase of the second hit process.

(3) Explore the Balance Ability Recovery

Experimental Design

All participants were required to participate in two experiments: CI (cryotherapy intervention group, CI) and CON (control group, CON). The first experiment was the CI, and to avoid possible influencing factors, the CON was three weeks later. In every single experiment, the balance ability of participants was measured at the six-time points: post-warm-up, post-fatigue, post-cryotherapy, 24 h post-cryotherapy, 48 h post-cryotherapy, and 72 h post-cryotherapy. The temperature of the laboratory is uniformly controlled at 26°C through air conditioning. As shown in Figure 7. The cryotherapy equipment (Chenhui Medical, Suzhou, China) in this study was cooled by a compressor, and R134A tetrafluoromethane and an antifreeze fluid in the bladder were in contact with the skin. The lowest temperature of the cryotherapy equipment was –5°C, and the maximum working time of cryotherapy was 30 min. Therefore, the cryotherapy equipment has met the requirements of this experiment. The participants first need to warm up at an adaptive speed for 4 minutes in the playground. After the warm-up, the participants will have 2 minutes to fully familiarize themselves with the experimental environment and instruments. And then measured the balance ability of the dominant legs of the participants. After completing the pretest of the experimental indicators, the participants were subjected to exercise muscle fatigue modeling. The experimental indicators were measured again after fatigue. The experimental indexes were measured in a uniform order, the static balance ability index was collected first, and then the dynamic balance ability index was collected. In the CI, subjects were required to sit on the laboratory chair in a quiet state after fatigue modeling. Meanwhile, the experimenter wrapped the cryotherapy device on the thigh and lower leg of the subject’s right leg. The temperature of CI in this study was set at 0°C. All subjects were wrapped in the same position to ensure full coverage of the thigh and lower leg area of the subject. The cryotherapy device was immediately attached to the subjects’ limbs, and the experimenters recorded the time through a stopwatch. The intervention time was controlled for 10 minutes, during which the subjects were not allowed to drink or eat.

When the stopwatch shows that the time is 10 minutes, the experimenter will remove the cryotherapy device from the subject's body, and the subject will measure the experimental indicators immediately. In the CON, after fatigue modeling, subjects were asked to sit on a chair in the laboratory in a quiet state. The intervention time was controlled for 10 minutes, and the experimenter recorded the time through a stopwatch. Subjects were not allowed to drink or eat during the intervention. At the end of the intervention, the subjects were asked to take measurements of the experimental indicators immediately.

Figure 7. The instrument of cryotherapy.

Muscle Fatigue Model

As shown in Figure 8. After the participants have fully warmed up, they will be tested for maximum squat load and the participants will perform repeated squats with a weight of 50 KG (kilogram, KG) by the Keiser (Fresno, United States, 002521PP). The barbell is required to be positioned in the back deltoid muscle of the neck. The downward movement of the squat ends when the thigh is below the horizontal plane. During the whole process, participants' movements are supervised and protected by a professional physical fitness coach.

Figure 8. Illustration of squat test and training.

The maximum strength of the participants was calculated by using Brzycki's (1990) 1 RM (repetition maximum, RM) formula:

$$1RM \approx \frac{\omega}{1.0278 - (0.0278 \times r)} \tag{1}$$

The ω represents the weight of the barbell during squats, and r represents the total number of squats completed under the weight of the barbell. Referring to the motility muscle fatigue modeling method of Pearcey [50] and MacDonald [51] et al, 60% of 1 RM was uniformly selected as the exercise load of the experiment. In a formal experiment, the participants performed 10 times *10 groups of weight-bearing squat training, and each group had 2–3 minutes of rest after the completion of the training. In addition, the time of each squat was strictly controlled in this experiment. During the squat process (centrifugal movement), the time was controlled at 4 seconds to control the centrifugal contraction process of the lower limb muscles of the participants. At the end of the squat process, the thighs should be below the horizontal level and paused for 1 second to control the peak contraction process of the lower limb muscles. In the process of squatting, the time is strictly controlled at about 3 seconds to control the centripetal contraction process of the lower limb muscles of the participants. The time control of the whole process is carried out by the experimenter using a stopwatch.

Index and Calculation Method

The measurement of dynamic balance: as shown in Figure 9. The YBT (Y-balance test, YBT) was used to measure the maximum extension reaching distance of participants in three directions anterior, posteromedial, and posterolateral based on single-leg standing.

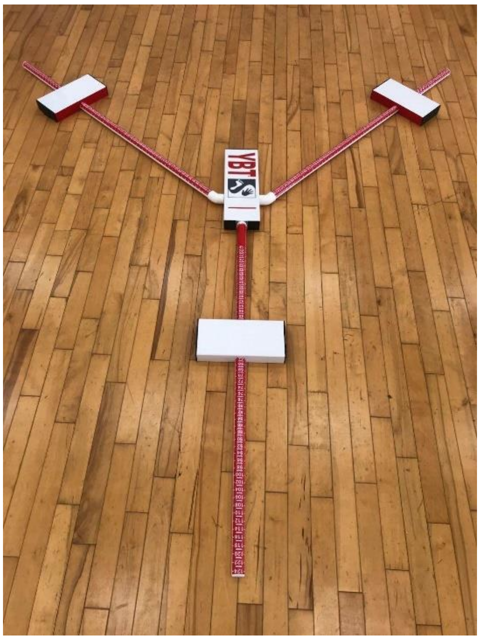

Figure 9. The Y-Balance Test system.

The right leg of all participants was selected as the standing leg. In the formal tests, repeat the test three times in each direction. The results are accurate to 0.5 cm. Retest if the following conditions occur during the test: 1) The standing leg deviates from the central footplate of the YBT system; 2) The unstable center of gravity causes the reaching leg to touch the ground; 3) The reaching leg is unable to back the starting position smoothly. The length of the reaching leg during the supine position was measured and recorded by experienced experimental (anterosuperior iliac spine to the center of the ipsilateral medial malleolus). Standardization of data: the reach distance in each direction was normalized to the leg length by calculating the %MAXD (maximized reach distance, MAXD). %MAXD was used to evaluate the dynamic balance ability of the participant. The specific calculation formula [52] was:

%MAXD = (anterior distance + posteromedial distance + posterolateral distance) / (3×leg length) ×100%. (2)

The measurement of static balance: the static balance ability was evaluated by COP area, the maximum displacement of COP on AP (Antero-Posterior axis, AP) and ML (Medium-Lateral axis, ML), and the displacement velocity of COP on AP and ML. The data of COP track during the 30-second eye-opening single-leg standing was collected by the Kistler force platform (AMTI, Watertown, United States, sampling frequency of 1000 Hz). In order to avoid the influence of visual factors on the static balance ability of subjects with single-leg support, all subjects were required to focus their eyes on the two-meter mark in front of them during the single-leg static balance test with eye-opening. The coordinates of each frame of COP were recorded by the force platform. As shown in Figure 10. The foot length and foot width were measured by a 3D foot scanner (Easy-Foot-Scan, OrthoBaltic, Kaunas, Lithuania) with an accuracy of 0.3 mm and the scanner volume of 400 (length)*200(width)*200(height) mm^3. Foot length and foot width were used to standardize the data.

Figure 10. Illustration of foot morphology measurement.

The experimental indexes of static balance capacity are calculated as follows [Cavalheiro 2009]: 95% static COP area:

$$S_{APML} = \frac{1}{N}\sum_{I-1}^{N} AP(i) \times ML(i) \tag{3}$$

$$D = \sqrt{(S_{AP}^2 + S_{ML}^2) - 4 \times (S_{AP}^2 \times S_{ML}^2 - S_{APML}^2)} \tag{4}$$

$$Major_{axis} = \sqrt{2 \times (S_{AP}^2 + S_{ML}^2 + D)} \tag{5}$$

$$Minor_{axis} = \sqrt{2 \times (S_{AP}^2 + S_{ML}^2 - D)} \tag{6}$$

$$Aera = \pi \times Major_{axis} Minor_{axis} \tag{7}$$

S_{AP} and S_{ML} are the standard deviations of the distance between COP and AP and ML directions, and S_{APML} is the covariance of the COP distance in the AP and ML.

The displacement velocity of COP in AP and ML:

$$V(n) = \frac{|d_{cop}(n+1) - d_{cop}(n)|}{T} \tag{8}$$

$$MV = \frac{1}{N-1} \sum_{i=1}^{N-1} V(i) \tag{9}$$

The displacement velocity of COP is obtained by calculating the average of all instantaneous velocities and finally all instantaneous velocities. The maximum displacement of COP is calculated by calculating the difference between the maximum and minimum values in AP or ML. The formula is as follows:

The maximum displacement in the ML = the maximum value of the X-axis – the minimum value of the X-axis. (10)

The maximum displacement in the AP = the maximum value of the Y-axis – the minimum value of the Y-axis. (11)

Experimental Intervention Method

The intervention of this experiment mainly included CON and CI. The temperature of the laboratory was stabilized at 26℃ by air conditioning, and the intervention temperature in CI of this study was set at 0℃. In CON, after completing fatigue modeling, subjects were required to sit in a laboratory chair in the same position in a quiet state for 10 minutes. The intervention time was controlled by the experimenter through a stopwatch. During the intervention, subjects were not allowed to drink or eat. When the stopwatch shows the time as 10 minutes, the intervention is over, and the subjects are required to measure the experimental indicators immediately. In CI, after completing fatigue modeling, the subjects were asked to sit in the laboratory chair in the same position in a quiet state. Meanwhile, the experimenter wrapped the cryotherapy instrument on the subjects' right thigh and calf, and all subjects were wrapped in the same position to ensure that the subjects' thigh and calf areas were all covered.

After the cryotherapy device was attached to the subjects' limbs, the experimenter recorded the time through a stopwatch immediately, and the intervention time was controlled for 10 minutes. During the intervention, the subjects were not allowed to drink or eat. When the stopwatch shows that the time is 10 minutes, the experimenter will remove the cryotherapy device from the subject's limb and the subject has to complete the experimental indicator measurement immediately.

(4) Explore the Muscle Force, Joint Reaction Force, and Joint Stiffness of the Lower Extremity during Topspin Forehand

Experimental Protocol and Equipment

This experiment was carried out in the biomechanics laboratory of the Ningbo University Research Academy of Grand Health. A Vicon motion capture system was used with a force platform to record the kinematics and kinetics data of subjects during movement. According to the sampling frequency setting of previous study [13–19], the kinematics and kinetics information were captured at 200 and 1000 Hz, respectively. As shown in Figure 11, an sEMG system (Delsys, Boston, United States) was used to record the muscle activity at 1000 Hz. Muscle force and activation of five muscles of the right leg were recorded in this study (Medial gastrocnemius, Lateral gastrocnemius, Semitendinosus, Rectus femoris, and Tibialis anterior). The mid-point of each muscle was selected for attaching the sEMG electrodes shown in Figure 12. The Gait2392 model was selected to complete the musculoskeletal modeling in the OpenSim (Stanford University, Stanford, United States), with the 39 reflective markers (12.5 mm in diameter) placement shown in Figure 12. The Gait2392 primarily lower extremity model with two legs and a lumped torso segment. Includes 23 degrees of freedom and 92 muscle-tendon actuators. Simulating and analyzing human movement that is dominated by lower extremity muscles. Results may be inaccurate during motions with high degrees of knee flexion. I choose this model because the model can be used for both kinematics and dynamics analyses [53].

Subjects were asked to wear tights and match table tennis shoes during the experiment and to use uniform rackets with the DHC Hurricane 3 and Butterfly Tenergy 05 Max rubber sheets, as well as uniform table tennis balls to complete the test on a professional table tennis table.

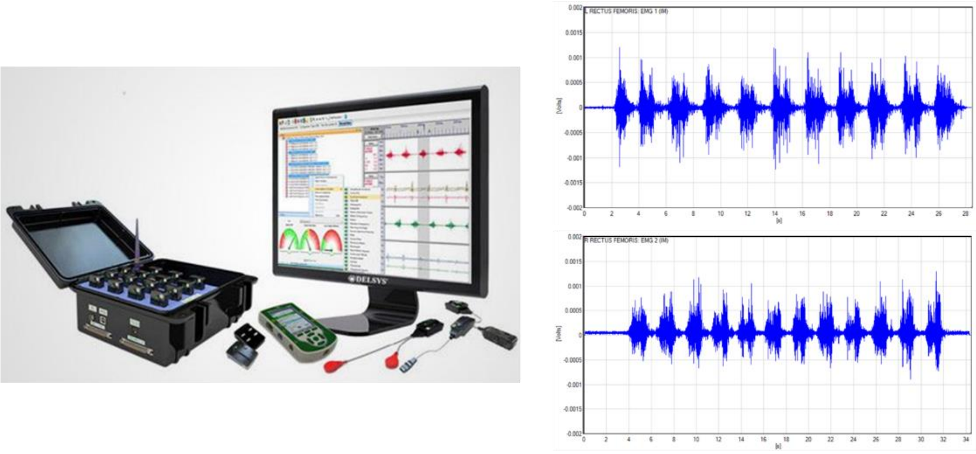

Figure 11. Illustration of the sEMG system.

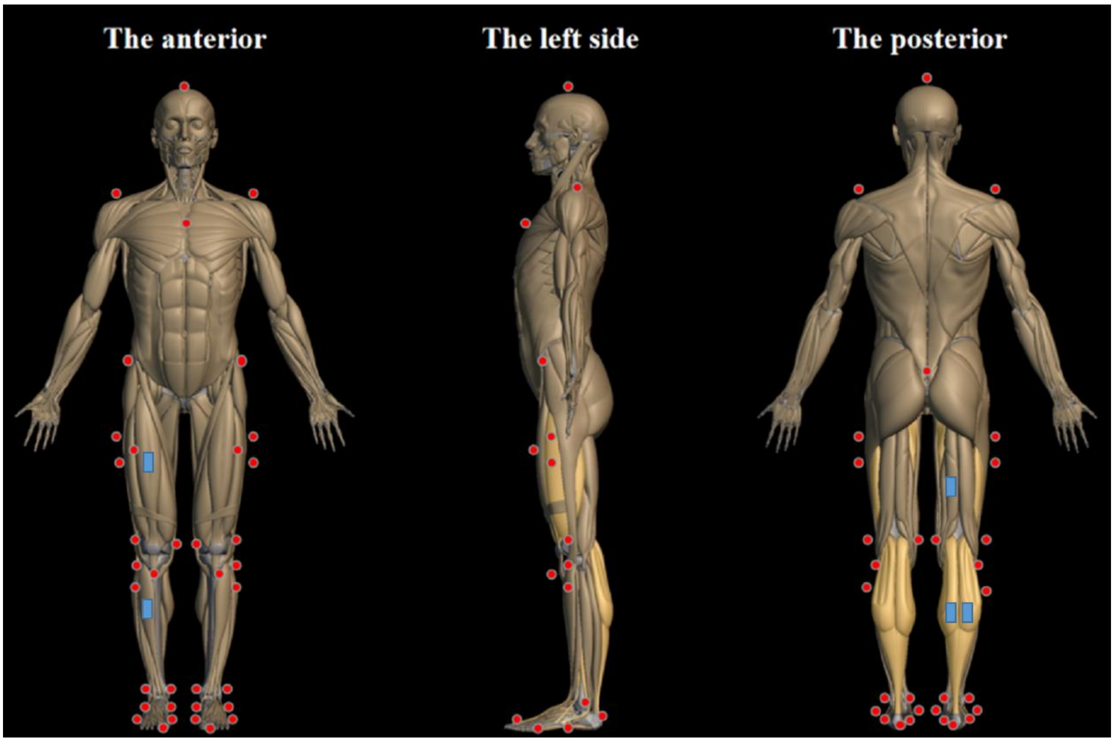

Figure 12. Illustration of the placement of the reflective markers and sEMG electrodes on three sides.

Note: The reflective markers were shown as the red point, and the sEMG electrodes were shown as the blue rectangle.

Procedure

At first, subjects completed 10 min of running and 5 min of static stretching at an adaptive speed in the playground to ensure the body was completed activation based on own consideration. Secondly, the selected muscle surface skin was cleaned to avoid affecting the accuracy of sEMG data (the skin surface of the selected muscle is first shaved with a razor, and then the skin is cleaned with an alcohol swab), and then the MVC (maximal voluntary contraction, MVC) collection of all muscles was completed. Subjects were required to stand on the force platform to finish the static coordinates collection. Before the formal test, subjects were allowed to complete five test tasks as a way of helping them quickly familiarize themselves with the laboratory environment. As shown in Figure 13A, the process of “a–i” was performed in the chasse step footwork, and the process of “j–n” was performed in the one-step footwork. In the chasse step footwork, the “a” shows the ready position. The “a–b” and “e–g” show the backward phase of the first stroke process and second stroke process, respectively. The “b–d” and “g–i” show the forward phase of the first stroke process and second stroke process, respectively. In the one-step footwork, the “j” shows the ready position, the “j–l” shows the backward phase, as well as the “l–n” shows the forward phase. In the formal test, the coach was asked to shoot the ball with normal served to the first impact zone (25 cm * 30 cm) and final impact zone (25 cm * 30 cm), respectively. Subjects were asked to perform topspin forehand by the one-step and chasse-step footwork to stroke the ball from the coach to the target area (25 cm * 30 cm), as shown in Figure 13B. The length and width of the impact zone and target area are set concerning previous studies [33]. Subjects were asked to complete three successful strokes by chasse step footwork, then complete three successful strokes by one-step footwork. The motion capture system, sEMG system, and force platform were connected by electrical signals to achieve the multi-parameter synchronous acquisition of the data of sEMG, kinetic, and kinematic. The motion smoothness was judged by the subjects themselves, as well as the quality and effect of the ball play were supervised by a qualified table tennis coach. Experimental operators can also evaluate the validity of data collection based on data performance.

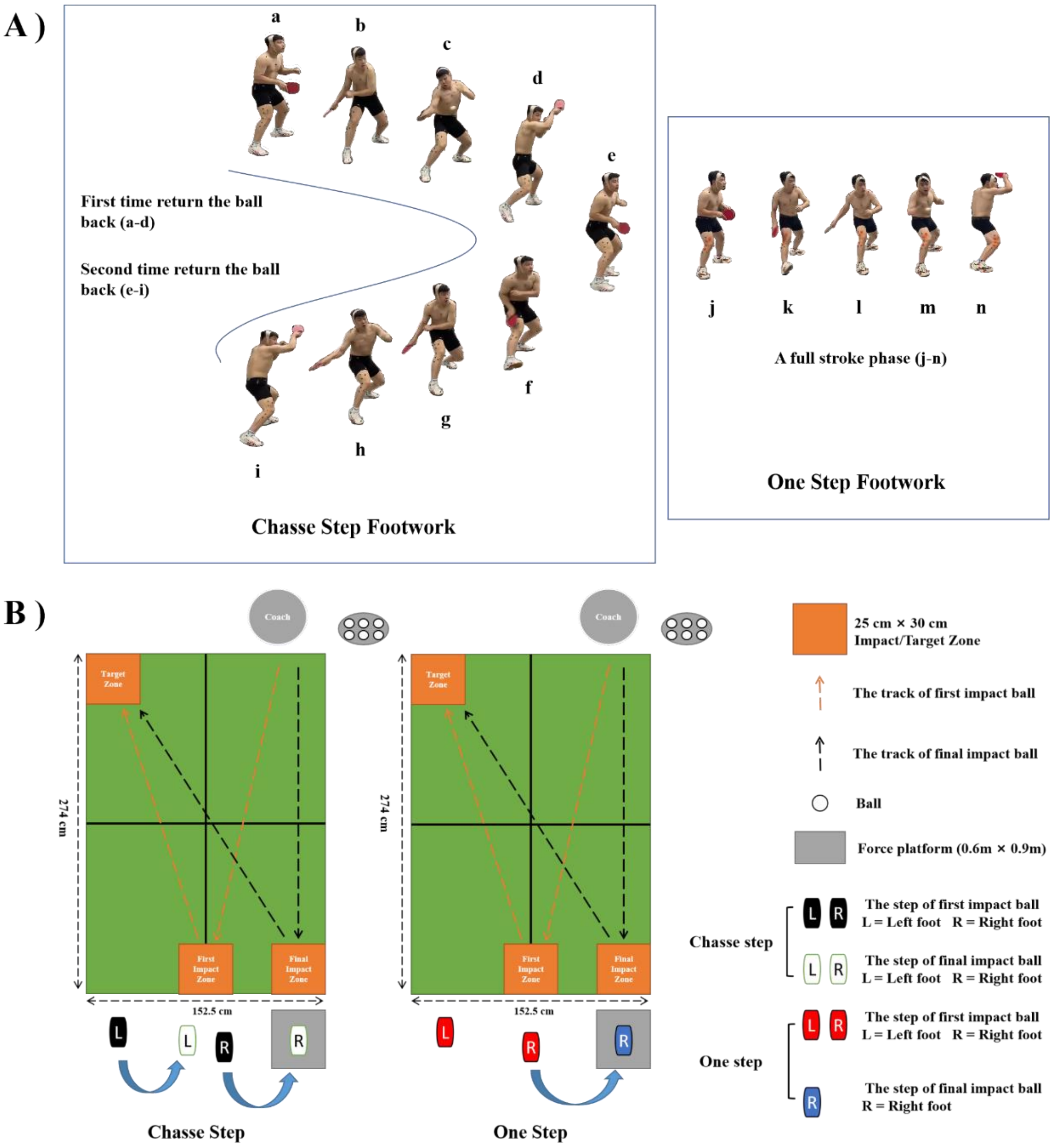

Figure 13. (A) Illustration of chasse step and one-step footwork. (B) Illustration of experiment process of a table tennis stroke.

(5) Explore the Kinematics and Dynamics of Lumbar and Pelvis during Topspin Forehand

Experimental Protocol and Equipment

The experiment was performed in the biomechanics laboratory of the Ningbo University Research Academy of Grand Health. As shown in Figure 14, the kinematics of participants were captured by eight-camera Vicon motion capture system which was set at the sampling frequency of 200 Hz. The kinetics of participants were recorded by force platform using a sampling frequency of 1,000 Hz. All devices used for data acquisition were electronically connected to achieve the multi-parameter synchronous acquisition of the test data. The Gait2392 model was selected to simulate the movement of the participant in the OpenSim. Participants used uniform rackets (Butterfly Tenergy 05 Max and DHC Hurricane 3 rubber sheets), balls (D40+, Double Happiness Sports Company, Shanghai, China), and playing table (Rainbow, Double Happiness Sports Company, Shanghai, China), as well as match table tennis shoes and tights during the experiment.

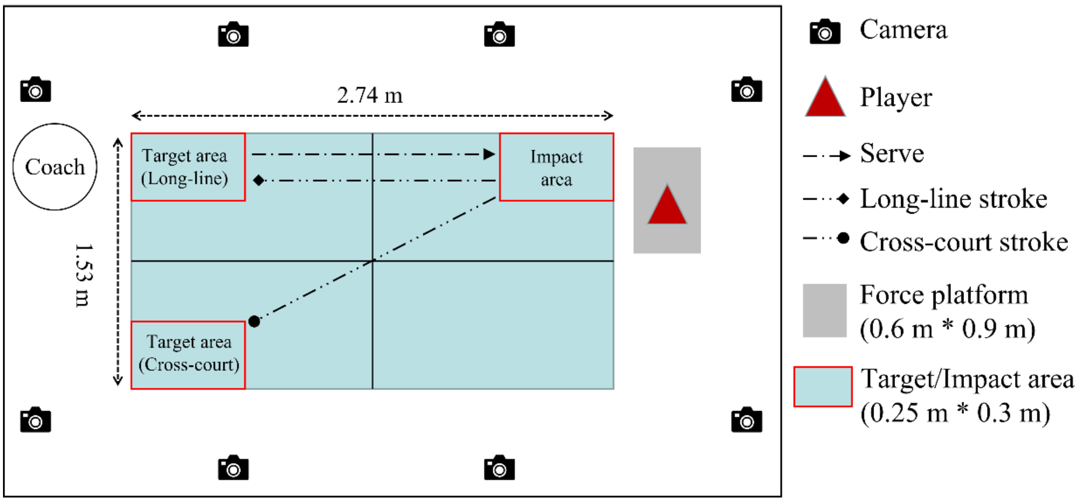

Figure 14. Experimental environment and set-up.

Procedure

Prior to the commencement of the formal test, participants were allowed to complete 5 min of static stretching and 10 min of running in a spacious area to warm up. Subjects were required to stand on the force platform to complete the static coordinates collection process after putting the reflective markers on the subjects' bodies. To check the operation of all the equipment and help the subjects quickly familiarize themselves with the laboratory environment, subjects were asked to perform five topspin forehand stroke tasks before the formal data collection session. As shown in Figure 14, in the formal test, the coach was shooting the ball with normal service to the impact area (0.25 m * 0.3 m). The subject stood on the right side of the playing table and was required to perform the topspin forehand stroke to return the ball to the LL (long-line, LL) target area (0.25 m * 0.3 m) and CC (cross-court, CC) target area (0.25 m * 0.3 m), respectively. The CC topspin forehand started first, then performed the LL topspin forehand. There is no rest time during the formal test until the successfully recorded 5 trial data of the CC and the LL topspin forehand for each participant, respectively. The subject and a qualified coach judged the quality of motion during the test. The test data were excluded if the drop point of the ball was out of the target area and the motion quality was questioned. Meanwhile, the data performance was also used to evaluate the validity of data collection. The size set of the impact and target area was as same as in previous studies [54].

Definition

In this study, only the data in the forward swing phase during the stroke were collected and analyzed. The pelvis movement in the transverse plane was defined as PAR (pelvis axial rotation, PAR), as well as the lumbar movement in the sagittal, frontal, and transverse plane, was defined as LF (lumber flexion, LF), LLB (lumbar left lateral bending, LLB), and LAR (lumbar axial rotation, LAR) in this study. As shown in Figure 15, “A”, “B”, and “C” are the CC and LL topspin forehand stroke process in the full body, lumbar, and pelvis view, respectively. Besides, the “a-c”, and “m-o” in CC and “d-f’, and “p-r” in LL indicate the “EB” and “EF”, respectively. The definition of EB and EF was completed in the Vicon Nexus 1.8.6 software (Oxford Metrics, Ltd., Oxford, United Kingdom). When the GRF (ground reaction force, GRF) wave reached the first peak value was defined as EB. And the second peak value in the GRF wave was defined as EF.

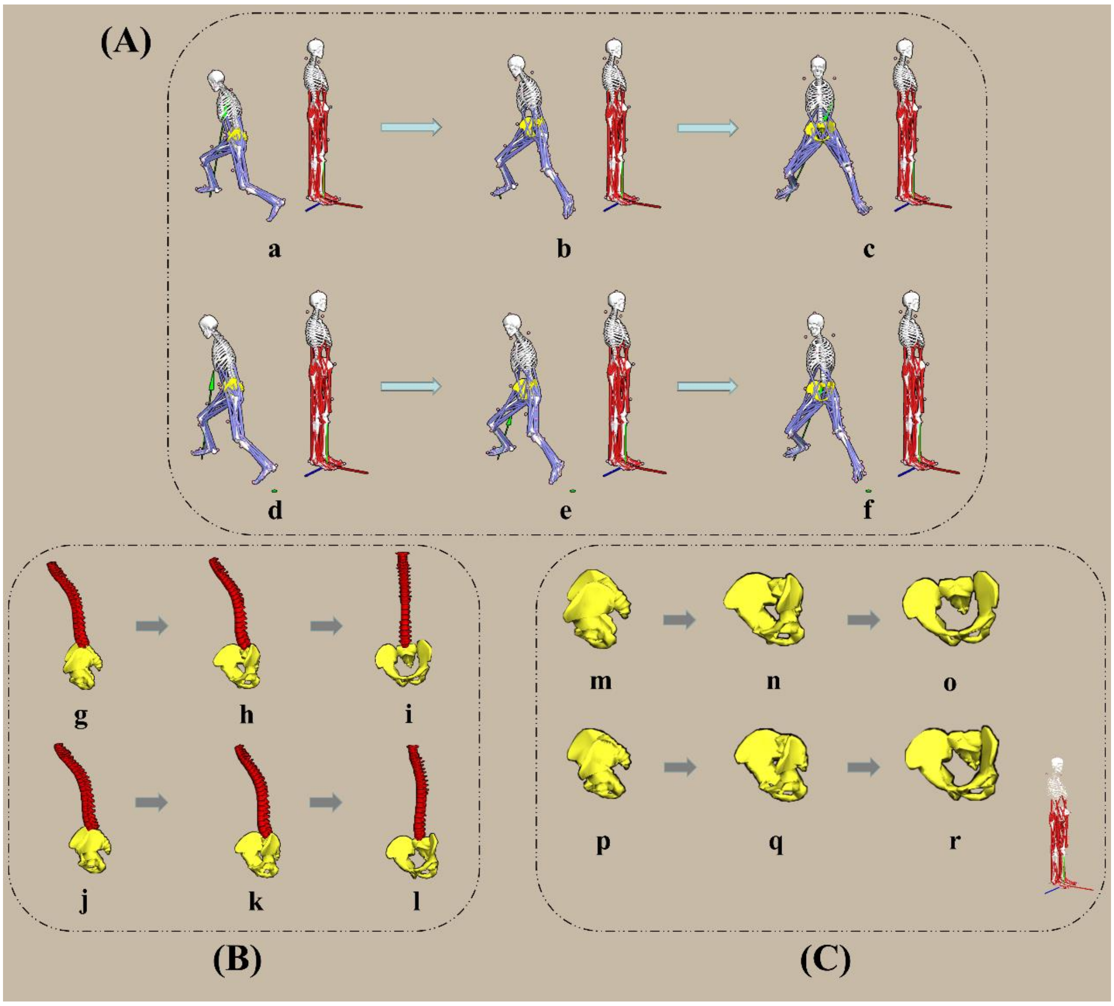

Figure 15. Diagram of the human musculoskeletal model of the CC and LL topspin forehand stroke.

Note: (A) indicate the topspin forehand stroke process. (a–c) and (d–f) indicate the CC and LL, respectively. (B) shows the lumbar and pelvis movement during the topspin forehand stroke. (g–i) and (j–l) indicate the CC and LL, respectively. (C) shows the pelvis movement during the topspin forehand stroke. (m–o) and (p–r) indicate the CC and LL, respectively.

2.3.2 Data Analysis Process and Statistical Analysis

(1) Explore the Kinematics of the Lower Extremity during Topspin Forehand

Statistical analysis and calculation were used by SPSS 19.0 version software (SPSS

Inc., Chicago, IL, United States). The normal distribution of variables was verified using the Shapiro-Wilks normality test. As the driven leg, the right leg kinematic differences in the topspin forehand loop between the two levels of players were examined by independent T-tests. The Analysis included joint angles, motion time, angular changing rate, and ROM of the ankle, knee, and hip joints. The significance level was set at $P < 0.05$.

(2) Explore the Plantar Force and Pressure during Topspin Forehand

During SPM1d (one-dimensional statistical parametric mapping, SPM1d) analysis processing, the generation of a separate integration curve was completed for each task before performing the SPM1d analysis. All kinetics data during the chasse step and one-step footwork were extracted. The next step was to generate a custom Matlab script and proceed with the interpolation process. The data points were expanded into a time series curve of 101 data points (representing 0–100% of the BP and FP phase). For the traditional discrete variable analysis, a script was written, and analysis was performed using Matlab R2019a (The MathWorks, Natick, United States) to extract and calculate the data for maximum plantar force, peak pressure of each plantar region, FTI, and PTI of the chasse step and one-step footwork during stroke play.

Before statistical analysis, all data were tested using the Shapiro–Wilk normality test ($W = 0.9361$, $P = 0.863$). All traditional discrete variable analyses were carried out by SPSS 19.0. Paired sample T-tests were used to analyze the maximum plantar force, peak pressure of each plantar region, FTI, and PTI. In the SPM1d analysis, the plantar force time series curve was marked as a 100% process. In addition, a paired-sample T-test in Matlab was used to analyze plantar force between the chasse step and one step during BP and FP, respectively. An alpha level of 0.05 ($\alpha = 0.05$) was set as being statistically significant.

(3) Explore the Balance Ability Recovery

SPSS 19.0 statistical software was used for statistical analysis of the collected data, which were expressed in the form of Mean ± SD. All data in this study are programmed and calculated by Matlab. Determine whether the data of each group is normal based on the boxplot, and the Shapiro-Wilk test was used to determine whether the data of each group followed an approximate normal distribution. The Two-way Repeated Measures ANOVA (analysis of variance, ANOVA) evaluates the subjects in different means of intervention and at different points in time the change of dynamic balance and static balance ability. Mauchly's spherical hypothesis test was used to determine whether the data of each group met the spherical hypothesis. When Mauchly's spherical hypothesis is satisfied, the influence of interaction terms on the dependent variable is judged to be statistically significant. If Mauchly's spherical test was not satisfied, the greenhouse-Geisser method was used to correct it and to judge again whether the influence of interaction terms on dependent variables was statistically significant. When the influence is statistically significant, the individual effects of factors within the study object should be analyzed one by one, and Bonferroni pairwise comparison of Post-hoc Analysis should be used for subsequent Analysis. If there is no statistical significance, the main effect of factors within the study object should be analyzed. When the main effect exists, pairwise comparisons are made. The significance level of this study was set as $P < 0.05$.

(4) Explore the Muscle Force, Joint Reaction Force, and Joint Stiffness of the Lower Extremity during Topspin Forehand

The sEMG signals recorded in the experiment were converted into activation through RMS (room mean square, RMS) processing (0 indicates no activation and 1 indicates full activation), and then it was compared with the activation obtained by the OpenSim optimization algorithm. A 4th-order band-pass filter between 10 and 500 Hz was applied to the sEMG data before it was full-wave-rectified, as well as a 10 Hz low-pass filter was used to smooth data.

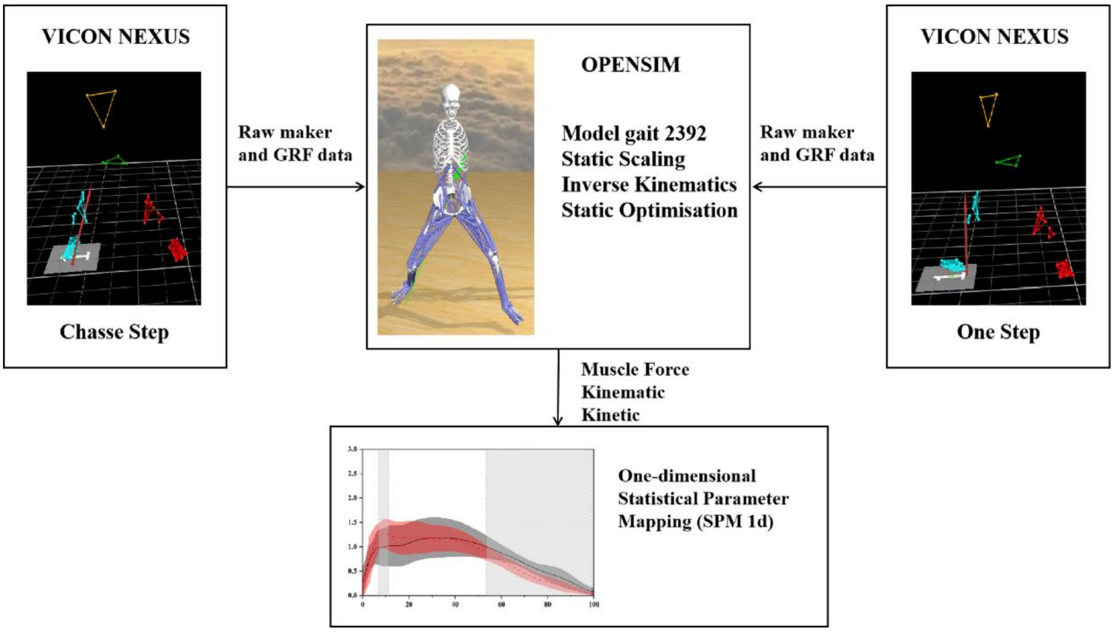

Figure 16. Flowchart of data processing in table tennis footwork.

Figure 16 shows the data process. Kinematics and GRF data of chasse step and one-step footwork during stroke were acquired and identified by the Vicon Nexus 1.8.6 software. The data was exported into a C3D. format file by the Vicon Nexus software then performs coordinate system conversion, low-pass filtering, data extraction, and format conversion for kinematics and ground reaction force data by Matlab.

As shown in Figure 17. In Matlab, perform the following steps: (1) convert the coordinate system of the kinematics and GRF data to the subsequent simulations coordinate system. (2) Use the 6 and 30 Hz fourth-order zero-phase lag Butterworth low-pass filters to filter the marker trajectory and the GRF. (3) The kinematics and GRF data of chasse step and one-step footwork during stroke were extracted and converted to the trc. and mot. formats required by the OpenSim simulation software.

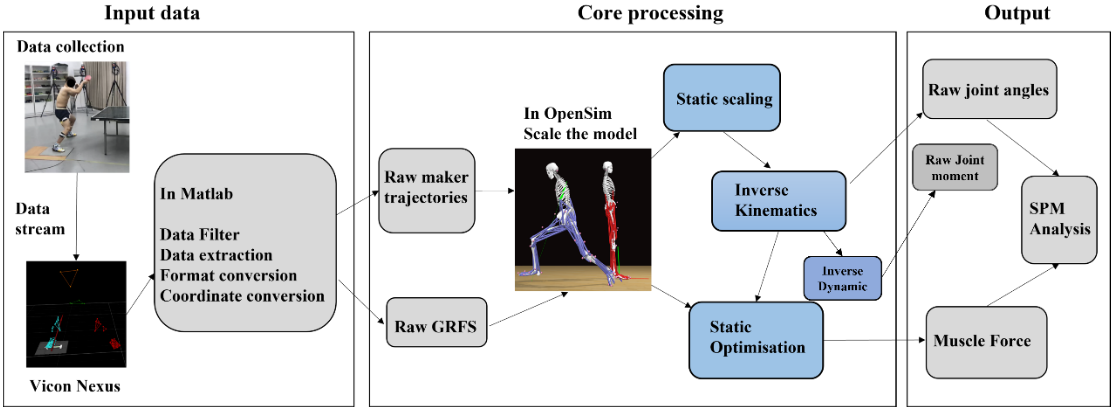

Figure 17. The research and data process workflow.

In OpenSim, perform the following steps: (1) Import the statics model and obtain the anthropometric model of subjects by the scale tool. Identify the starting and ending point of muscle, as well as ensure the moment arms consistent with the subjects' limb length [55]. (2) Calculate the kinematics data of the one-step and chasse step footwork during a stroke by the IK (inverse kinematics, IK) tool and create a motion file (mot). Then, import the markers and GRF files by the inverse dynamics tool and calculate the joint moment of the subjects. (3) Smoothing the kinematics data by the residual reduction algorithm to improve the preciseness of the dynamic data to be consistent with the kinematics and kinetics data measured in the experiment. (4) Calculate the muscle activation and the muscle force by the CMC (computed muscle control, CMC) with the smoothed kinematics data calculated in the last step [56]. (5) Running the Joint Reaction Force tool to calculate the joint reaction force of lower extremity joints.

The joint stiffness was calculated as follows: the ratio of the joint moment changes to the joint angle changes from the initial stage to the maximum ankle dorsiflexion.

For SPM1d, lower limb joint angle and moment, as well as the muscle force of the chasse step and one-step footwork during stroke were extracted. All data of the stroke phase was expended into a time series curve of 101 data points by a Matlab custom script. The open-source SPM1d paired samples t-test script was used to analyze the difference in joint angle, joint moment, and muscle force between the chasse step and one-step footwork during stroke [44]. The significance level was set at $p < 0.05$ in this study.

(5) Explore the Kinematics and Dynamics of Lumbar and Pelvis during Topspin Forehand

As shown in Figures 14 and 15, the GRF and kinematic data during CC and LL topspin forehand were identified and acquired using Vicon Nexus 1.8.6 software. The data was exported from the Vicon Nexus with a c3d. format file, and use Matlab to perform coordinate system conversion, lower pass filtering, data extraction, and format conversation for all data. The detailed process in Matlab has been outlined in previous studies [33, 54] as follows: convert the coordinate to the subsequent simulation coordinate system, filter the marker trajectory and the GRF, and convert the formats of data to the trc. and mot. formats that are required by OpenSim. The statics model of the subjects was imported into OpenSim and the anthropometric model was obtained. Then we identified the muscle's starting and ending points and ensured the moment arms were consistent with the length of the subject's limb [55]. We used the IK to calculate the kinematics data of the subject during CC and LL topspin forehand and created a motion file using mot format. We then imported the GRF and markers files using the ID (inverse dynamics, ID) and calculated the joint moment. In OpenSim, the weighted least square problem was solved by the IK function to minimize the distance of markers' placements between the experimental and virtual; the generalized positions, velocities, and accelerations defined the motion of the model, which resulted in the unknown generalized forces were calculated by those known motion variables.

Kinematics and the moment of the pelvis and lumbar were analyzed by SPM1d analysis in Matlab. The Rom and peak moment of the pelvis and lumbar were analyzed by independent samples t-test in SPSS. In the SPM1d analysis, we performed the custom script in Matlab to expend all data into a time series curve of 101 data points. The significance level in this study was set as $p < 0.05$.

2.4 Musculoskeletal Simulation

2.4.1 Model Select

As shown in Figure 12 and Figure 18. The Gait2392 model was selected to simulate the movement of the participant in the OpenSim (Stanford University, Stanford, United States), and the thirty-nine reflective markers (12.5 mm in diameter) placement was replicated according to the previous studies [55].

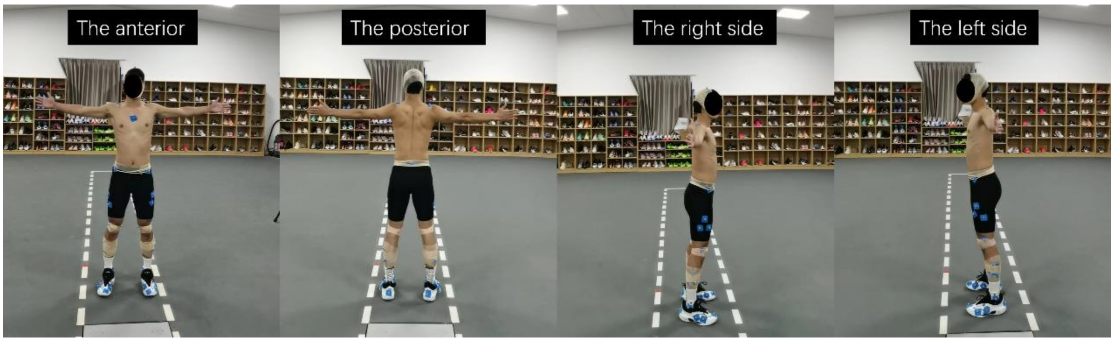

Figure 18. Illustration of the maker placement of the Gait 2392 model.

2.4.2 Preparing Motion Data

(1) Coordinate Systems and Data Format Transform

The data was exported into c3d. format file by the Vicon Nexus software. Convert this file into Marker point trajectory experimental data .trc, .mot, and force plate data during static calibration. The conversion method is to use the c3dExport.m file provided by OpenSim for conversion.

(2) Scale Tool and Scale Factors

Open the Scale model tool, enter the subject's weight, import the static standing action trajectory file .trc, then select Scale Factors, and adjust the Scale Factors scaling factor according to personalized needs. There are two main ways, one is the Use Measurement, one is the Use Manual Scales. In this study, the Use Manual was selected. Click the Static Pose Weights tab to set the Markers mark point weight value. The purpose of setting the weight value is to allow the scaling tool to have different tracking effects on the errors of different markers during the scaling process. For markers with heavy weights, the scaling tool will make the errors smaller. Usually for markers with bony landmarks, we will set a larger weight value. For other markers that do not require high accuracy, we set a smaller weight value. The setting of the weight value must be adjusted to varying degrees based on different data. Here they are all set to 100. Click the run button and save the scaled model after scaling is completed for the IK, ID, and Joint Reaction.

3 Results

3.1 Results of Biomechanical Experiments

3.1.1 Explore the Kinematics of the Lower Extremity during Topspin Forehand

(1) Motion Time

As outlined in Table 1. The time taken to perform a topspin loop was 0.96 ± 0.09 s and 0.97 ± 0.09 s for EA and MA, respectively. EA demonstrated significantly less time than MA in the BP (t-value = −3.097, P = 0.004), however, EA showed a significantly larger time in the FP (t-value = 2.180, P = 0.038). Moreover, there were no significant differences in the time during the entire stage between EA and MA (t-value = −0.277, P = 0.784).

Table 1. Comparison of time at the phase of BP and FP between EA and MA (unit: second).

Variables	**EA (Mean±SD)**	**MA (Mean±SD)**	**P-Value**
BP	0.39±0.06	0.45±0.05	0.004*
FP	0.57±0.07	0.52±0.06	0.038*
Entire phase	0.96±0.09	0.97±0.09	0.784

Note: * indicates a significant difference between the EA and MA. BP, backward-swing phase; FP, forward-swing phase; EA, elite athlete; MA, medium athlete.

(2) Joint Angle

Table 2 and Figure 19 show the angles of the joint at BE as well as FE in the transverse, frontal, as well as sagittal planes for both EA with MA. In the frontal plane as well as the transverse plane, EA displays significant differences in joint angles for the entire stage compared with the MA. In the sagittal plane, EA showed significantly less knee (t-value = −7.496, P < 0.001) and hip (t-value = −25.397, P < 0.001) flexion in the BE phase compared with MA. In the frontal plane, EA showed a significantly larger ankle varus (t-value = 3.282, P = 0.003) and eversion (t-value = 8.799, P < 0.001) than MA in the BE and FE phases, respectively. Moreover, in the transverse plane, EA displayed a significantly larger ankle internal rotation (t-value = −3.320, P = 0.003) and external rotation (t-value = −7.428, P < 0.001) than MA in the BE and FE phase, respectively. EA showed a significantly larger knee external rotation (t-value = 5.027, P < 0.001) and internal rotation (t-value = 19.219, P < 0.001) in the BE and FE phases respectively compared with MA. However, MA showed a significantly larger hip external rotation (t-value = −6.299, P < 0.001) and internal rotation (t-value = −10.590, P < 0.001) in the BE and FE phases respectively compared with EA.

Table 2. Comparison of joint angles at key events between EA and MA (unit: degrees).

Variables	ANKLE BE (Mean±SD)	ANKLE FE (Mean±SD)	KNEE BE (Mean±SD)	KNEE FE (Mean±SD)	HIP BE (Mean±SD)	HIP FE (Mean±SD)
X(EA)	15.77±5.12*	15.96±13.52	41.15±8.83*	42.50±22.52	48.31±2.13*	2.00±11.29*
X(MA)	12.34±3.27*	22.13±2.37	58.80±2.29*	47.86±5.00	66.48±1.77*	23.35±2.28*
Y(EA)	4.85±3.78*	17.42±3.59*	14.74±2.86*	20.73±3.28*	-10.64±3.12	-26.05±7.36*
Y(MA)	1.55±0.97*	8.74±1.31*	30.18±2.12*	36.01±2.14*	-11.88±2.18	-34.07±2.45*
Z(EA)	-25.25±16.53*	-50.60±8.36*	18.86±5.94*	13.61±1.96*	23.60±6.41*	11.37±1.41*
Z(MA)	-10.74±3.70*	-33.67±2.82*	10.96±1.33*	-0.10±2.20*	34.18±1.13*	16.40±1.18*

Note: x–the sagittal plane; y–the frontal plane; z–the transverse plane. BE, backward-end; FE, forward-end; EA, elite athlete; MA, medium athlete. * indicates a significant difference at the hip, knee, and ankle (respectively) (P < 0.05).

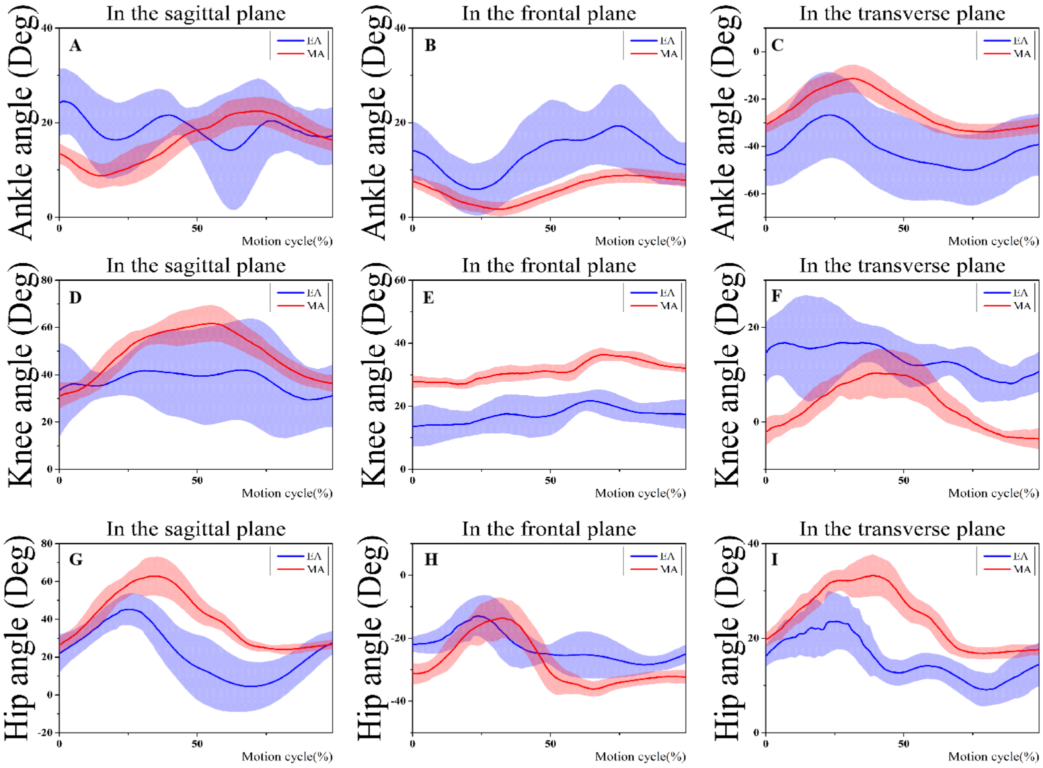

Figure 19. Changes of the lower limb joints angle during the entire phase in three planes.

Note: (A–C) Ankle angle changes during the entire phase in three planes; (D–F) knee angle changes during the entire phase in three planes; (G–I) Hip angle changes during the entire phase in three planes.

(3) Range of Motion

ROM at BP and FP between EA and MA in all planes are displayed in Table 3 and Figure 20. Lower-limb ROM showed significant differences during the BP as well as FP phases between EA and MA. Compared with MA, EA showed significantly larger ankle dorsiflexion (t-value = 3.838, P = 0.001) and plantarflexion (t-value = 4.792, P < 0.001) ROM in the BE and FE phase respectively. Moreover, EA showed a significantly larger ankle varus (t-value = 3.788, P = 0.001) and external rotation (t-value = 2.251, P = 0.032) ROM in the BE and FE phase respectively. However, EA showed significantly less hip flexion (t-value = −5.836, P < 0.001) and external rotation (t-value = −4.211, P < 0.001) ROM in the BE phase contrast with MA.

Table 3. Comparison of ROM at the phase of BP and FP between EA and MA (unit: degrees).

	ANKLE		KNEE		HIP	
Variables	**BP (Mean±SD)**	**FP (Mean±SD)**	**BP (Mean±SD)**	**FP (Mean±SD)**	**BP (Mean±SD)**	**FP (Mean±SD)**
X(EA)	11.50±3.76*	19.66±6.31*	16.02±5.62*	14.48±4.25	25.01±9.10*	44.93±10.52
X(MA)	7.23±2.11*	10.95±3.13*	28.16±5.92*	16.31±4.79	39.94±3.92*	42.90±2.32
Y(EA)	9.78±3.08*	15.36±3.47*	5.90±1.28*	8.44±1.80	12.00±2.47*	20.44±4.45*
Y(MA)	6.60±1.06*	7.22±1.03*	4.65±1.28*	15.36±3.47	20.43±2.40*	24.90±2.51*
Z(EA)	20.52±5.77	27.05±6.66*	12.54±3.37	11.67±6.25	8.11±6.23*	13.74±7.51*
Z(MA)	21.73±3.31	22.75±3.20*	13.34±2.91	13.37±2.40	15.16±1.80*	19.45±1.14*

Note: x–the sagittal plane; y–the frontal plane; z–the transverse plane. BP, backward-swing phase; FP, forward-swing phase; EA, elite athlete; MA, medium athlete. * indicates a significant difference at the hip, knee, and ankle (respectively) (P < 0.05).

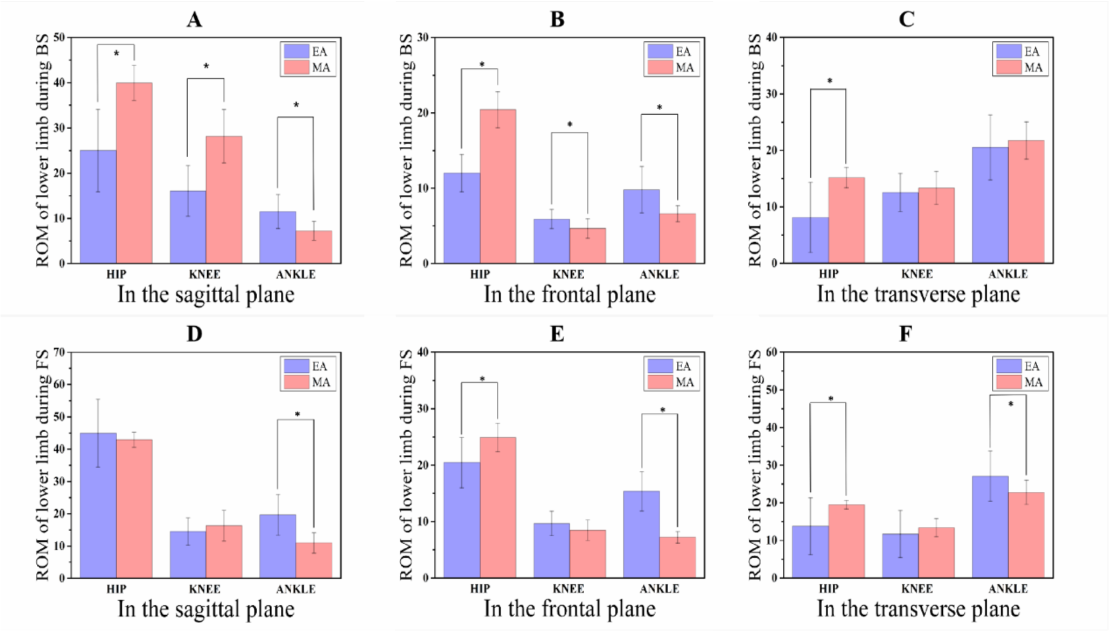

Figure 20. Changes of lower limb ROM during BS and FS phase in three planes.

Note: (A–C) Lower limb ROM changes during BP in three planes; (D–F) Lower limb ROM changes during FP in three planes.

(4) Angular Changing Rate

Angular changing rates at BP and FP phases between EA and MA in all planes are shown in Table 4 and Figure 21. In the stages of BP and FP, both in the sagittal and frontal plane, the angular changing rate of the ankle joint for the EA was significantly larger than MA. However, EA showed a significantly smaller angular changing rate in the hip (t-value = −4.572, $P < 0.001$) and knee (t-value = −5.592, $P < 0.001$) joint during BP in the sagittal plane.

Table 4. Comparison of the angular changing rate at the phase of BP and FP between EA and MA (unit: degrees/second).

Variables	ANKLE BP (Mean±SD)	ANKLE FP (Mean±SD)	KNEE BP (Mean±SD)	KNEE FP (Mean±SD)	HIP BP (Mean±SD)	HIP FP (Mean±SD)
X(EA)	34.01±11.80*	30.98±8.48*	37.56±10.15*	25.06±5.68*	55.02±19.10*	72.26±14.85
X(MA)	15.09±4.98*	21.81±8.05*	59.60±11.30*	31.44±6.71*	89.62±8.79*	77.66±12.90
Y(EA)	23.74±9.93*	26.22±4.61*	15.89±4.64*	16.22±5.23	33.84±7.70*	34.07±9.74*
Y(MA)	14.86±2.08*	13.70±2.21*	10.99±3.33*	16.50±4.47	46.23±8.57*	46.70±9.04*
Z(EA)	55.99±16.37	47.63±13.89	31.03±5.63	18.66±8.91*	10.46±15.21*	15.97±11.52*
Z(MA)	46.39±7.50	43.07±8.23	28.81±4.93	25.73±4.15*	34.19±3.32*	35.53±5.44*

Note: x–the sagittal plane; y–the frontal plane; z–the transverse plane. BE, backward-swing phase; FE, forward-swing phase; EA, elite athlete; MA, medium athlete. * indicates a significant difference at the hip, knee, and ankle (respectively) ($P < 0.05$).

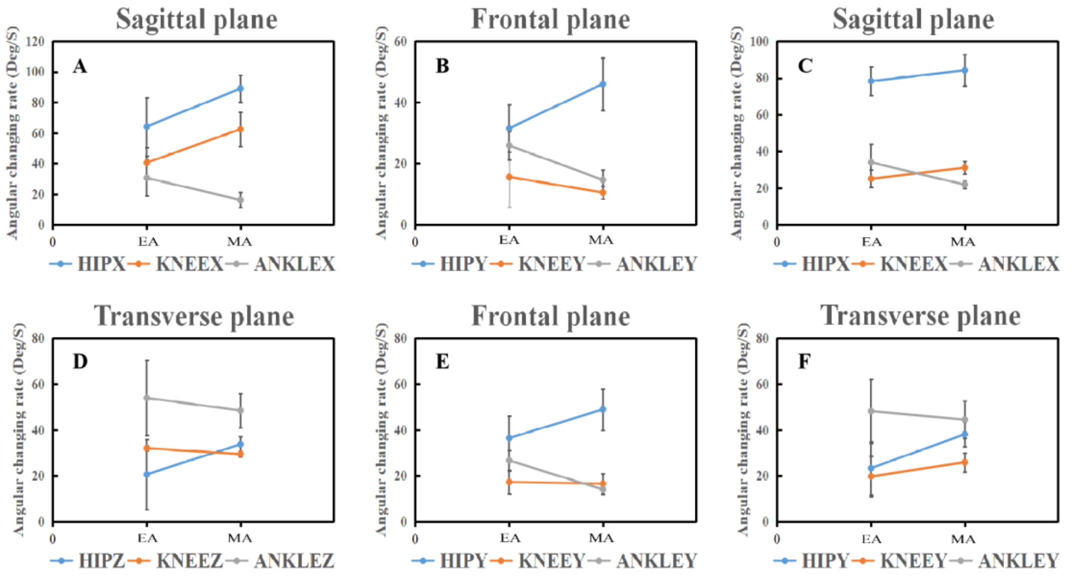

Figure 21. Angular changing rate of lower limb joints during BP and FP phase in three planes.

Note: The top is the BP, Bottom is the FP. (A–C) Angular changing rate of the lower limb during BP in three planes; (D–F) angular changing rate of the lower limb during FP in three planes.

3.1.2 Explore the Plantar Force and Pressure during Topspin Forehand

(1) Plantar Force

As shown in Figure 22, the one step produced a greater plantar force than the chasse step during 6.92–11.22% BP ($P = 0.039$). The chasse step produced a greater plantar force than the one step during 53.47–99.01% BP ($P < 0.001$). During the FP, the chasse step showed a greater plantar force than one step in 21.06–84.06% ($P < 0.001$).

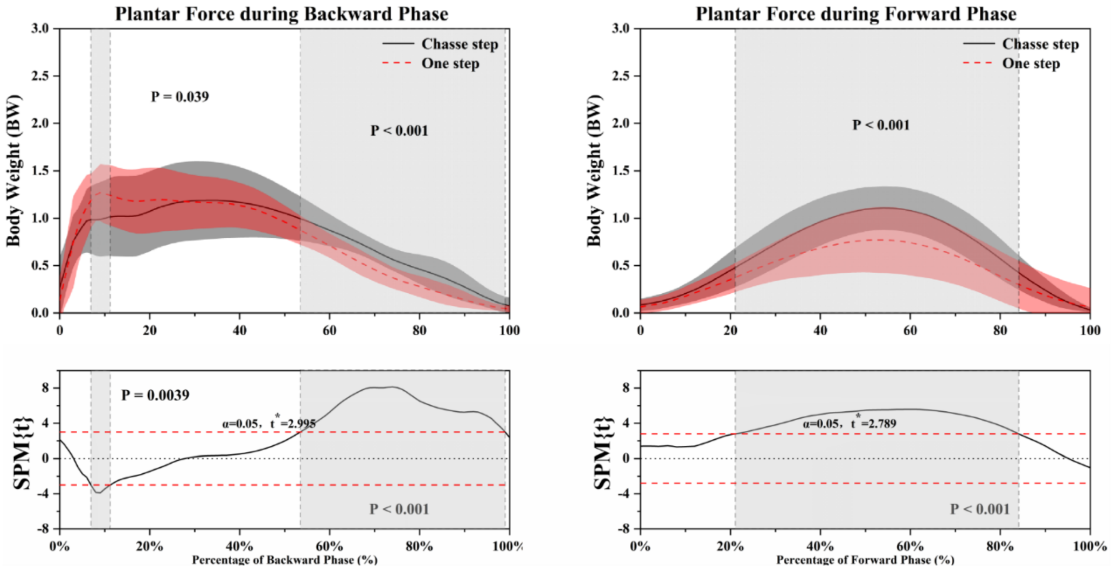

Figure 22. The SPM1d results of plantar force between the chasse step and one step during the BP and FP.

Note: Grey-shaded areas indicate that there are significant differences ($p < 0.05$) between the chasse step and one step. The top figures refer to the comparison of plantar force between the chasse step and one step. The bottom figures refer to the details of the SPM1d results. BW means body weight. "$\alpha = 0.05$" means set the 0.05 as being statistically significant. "*" refers to significance with $p < 0.05$

(2) Maximum Plantar Force

As shown in Table 5, the one step produced a greater maximum plantar force than the chasse step in the BP ($P = 0.032$). In addition, the chasse step produced a greater maximum plantar force in the FP ($P = 0$). The P-value shown at 0 means that the comparison result shows that the chasse step and one-step is significant different.

Table 5. The comparison of maximum plantar force during BP and FP between the chasse step and one step. (Unit: BW).

	Phase	**Chasse Step (Mean±SD)**	**One step (Mean±SD)**	**P-value**
maximum plantar force	BP	1.27 ± 0.38	1. 41 ± 0.24	0.032*
	FP	1.12 ± 0.23	0.82 ± 0.33	0*

Note: "*" refers to significance with $p < 0.05$.

(3) Plantar Pressure

As shown in Table 6 and Figure 23, for the Toe, the chasse step produced a greater peak pressure than the one step in the FP (P = 0). In the LF, the one step produced a greater peak pressure than the chasse step during the BP (P = 0.042). In addition, the one step produced a greater peak pressure than the chasse step in the LR (P = 0) and MR (P = 0) during BP.

Table 6. The peak pressure comparison of each plantar region between the chasse step and one step at BP and FP. (Unit: kpa).

Partition	**Phase**	**Chasse Step (Mean±SD)**	**One-step (Mean±SD)**	**P value**
T	BP	174.97 ±88.64	178.13 ± 89.03	0.742
	FP	388.85 ± 165.38	277.14 ± 59.61	0*
LF	BP	100.52 ± 20.74	116.04 ± 42.58	0.042*
	FP	129.44 ± 45.84	132.60 ± 83.07	0.764
MF	BP	243.75 ± 91.12	262.45 ± 114.63	0.069
	FP	379.43 ± 83.39	348.65 ± 145.31	0.078
M	BP	119.01 ± 23.56	119.01 ± 41.84	1.000
	FP	55.82 ± 24.29	47.71 ± 19.72	0.104
LR	BP	395.11 ± 64.81	563.72 ± 83.89	0*
	FP	90.43 ± 74.95	70.74 ± 61.52	0.206
MR	BP	404.27 ± 146.27	517.96 ± 119.44	0*
	FP	85.58 ± 57.32	85.17 ± 60.46	0.976

Note: “*” refers to significance with $p < 0.05$, the value which higher than 0.05 refers to there has no significant difference between chasse step and one-step.

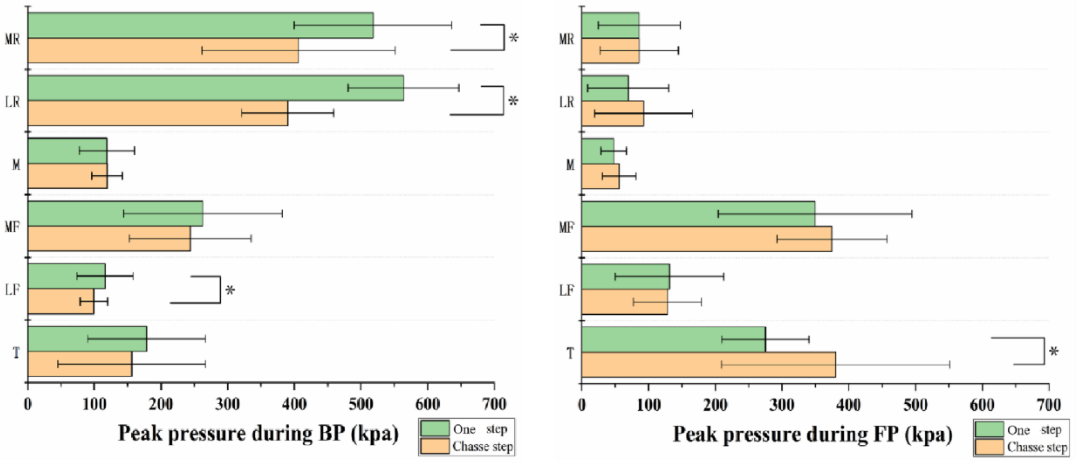

Figure 23. Comparison of peak pressure of each plantar region during BP and FP.

Note: The asterisk (*) refers to significance with $p < 0.05$.

(4) PTI and FTI

As shown in Table 7 and Figure 24, during BP, the chasse step produced a greater FTI (P = 0) and a lower PTI (P = 0) than the one step. During FP, the chasse step produced a greater FTI (P = 0) and PTI (P = 0.001) than the one step.

Table 7. FTI and PTI comparison between the chasse step and one step during BP and FP.

	Phase	**Chasse Step (Mean±SD)**	**One Step (Mean±SD)**	**P value**
FTI (N·s)	BP	161.31 ± 20.73	148.13 ± 13.49	0*
	FP	102.29 ± 31.87	72.17 ± 31.04	0*
PTI (Ns/cm^2)	BP	69.70 ± 7.98	77.91 ± 11.65	0*
	FP	83.49 ± 16.69	67.85 ± 26.14	0.001*

Note: “*” refers to significance with $p < 0.05$.

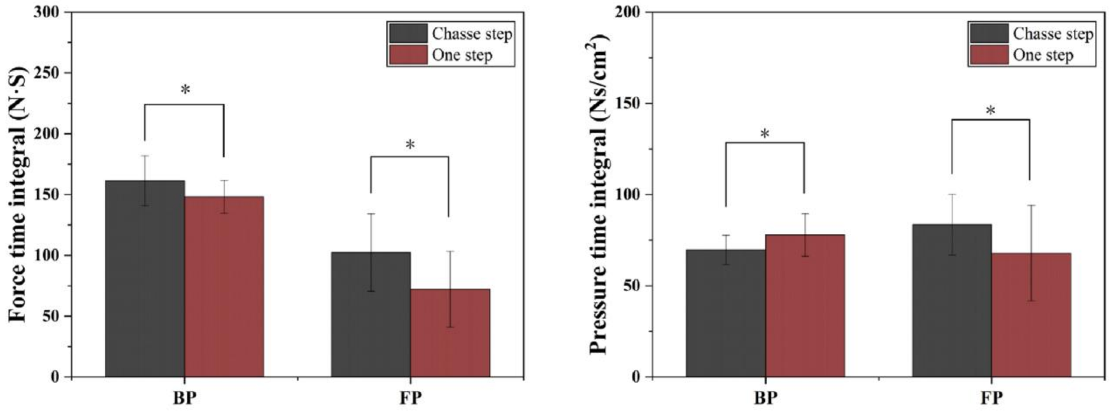

Figure 24. Comparison of PTI and FTI of plantar on driving foot between chasse step and one step during BP and FP.

Note: The left shows the force-time integral, right shows the pressure-time integral. “*” refers to significance with $p < 0.05$.

3.1.3 Explore the Balance Ability Recovery

(1) Static Balance Recovery

COP Area

As shown in Table 8. Repeated measurement ANOVA was conducted for the COP area. Mauchly’s spherical hypothesis test found that interaction term group * time met the spherical test (P = 0.509), and the interaction between the two was not statistically significant, F (5, 35) = 1.557, P = 0.237. Therefore, it is necessary to further interpret the principal effect of group factors and time factors. If the principal effect of factors within the study object is greater than two levels, pairwise comparison should be carried out later. Since there are only two levels of grouping factors, there is no need to test whether the spherical hypothesis is true. The principal effect of group factors on COP area was not statistically significant, F (1, 7) = 0.13, P = 0.912. The principal effect of the time factor on the COP area was not statistically significant, F (5, 35) = 1.992, P = 0.104. The COP area of the CI was 5.788 (95% CI: –125.708 ~ 114.133) mm2 smaller than that of the CON, and the difference was not statistically significant.

Table 8. Table of COP area of two interventions at each moment.

COP area (mm^2)	**①CI**		**②CON**			
	Mean ± SD	**95%CI**	**Mean ± SD**	**95%CI**	**P-value**	**Δ (①-②)**
Post-warm-up	380.43±179.10	[230.694-530.161]	401.68±148.06	[277.897-525.455]	0.912	-5.788
Post fatigue	741.44±409.30	[399.259-1083.623]	635.56±287.52	[395.189-875.936]		
post-intervention	660.15±169.63	[518.337-801.968]	431.70±294.44	[185.542-677.863]		
24h Post-intervention	618.55±501.74	[199.086-1038.02]	518.14±366.19	[211.998-824.286]		
48h Post-intervention	523.66±257.03	[308.779-738.546]	872.63±516.23	[441.049-1304.21]		
72h Post-intervention	415.71±187.73	[258.771-572.658]	514.96±445.14	[142.819-887.111]		
RM ANOVA	Whether the spherical hypothesis is satisfied? Yes (P = 0.509)		F (5, 35) = 1.557		The interaction was not significant (P = 0.237)	

The Maximum Displacement of COP in ML

As shown in Table 9. Repeated measurement ANOVA was performed for the maximum displacement of COP in ML. Mauchly's spherical hypothesis test found that interaction term group * time meets the spherical test (P = 0.313), and the interaction between them was significant, F (5, 35) = 7.485, P < 0.001. Therefore, separate effect tests for group and time factors are needed further. Simple effect analysis of group factors found that the group factor at 72 h post-intervention had a statistically significant effect on the maximum displacement of COP in the ML. The maximum displacement of COP on ML in CI was smaller than that in CON, and there was a significant difference (P = 0.005, F (1, 7) = 16.433). The time factor of CI met the spherical test (P = 0.068). The intrasubjective effect test showed that the influence of the time factor on the maximum displacement of COP in ML was statistically significant in the CI, P = 0.001, F (5, 35) = 5.027, so another pairwise comparison of six time points was needed. After the simple effect analysis of the time factor, it was found that the maximum displacement of COP in the ML at the moment of post-warm-up was less than post-intervention in the CI, and there was a significant difference (P = 0.007). The time factor of the CON met the spherical test (P = 0.195), and the intrasubjective effect test showed that the influence of the time factor on the maximum displacement of COP in the ML was not statistically significant (P = 0.053, F (5, 35) = 2.449).

Table 9. The COP maximum displacement of the two interventions at each moment in ML.

maximum displacement (%)	**①CI**		**②CON**			
	Mean±SD	**95%CI**	**Mean±SD**	**95%CI**	**P-value**	**Δ (①-②)**
Post-warm-up	31.88±5.38	[27.374-36.376]	33.75±3.01	[31.232-36.268]	0.243	-1.875
Post fatigue	42.87±7.15	[36.898-48.849]	42.90±8.16	[36.075-49.723]	0.991	-0.025
post-intervention	41.94±6.58a	[36.439-47.446]	40.76±4.98	[36.596-44.922]	0.653	1.184
24h post-intervention	38.78±8.57	[31.615-45.936]	39.33±8.17	[32.506-46.160]	0.853	-0.557
48h post-intervention	37.11±6.84	[31.396-42.825]	40.01±6.71	[34.401-45.618]	0.106	-2.899
72h post-intervention	36.01±8.88*	[28.585-43.439]	39.39±8.85	[31.993-46.784]	0.005	-3.376*
RM ANOVA	Whether the spherical hypothesis is satisfied? Yes (P = 0.313)		F (5, 35) = 7.485		The interaction was significant (P < 0.001)	

Note: “a” indicates that there is a significant difference between the moment of post-warm-up and other moments (post fatigue, post-intervention, 24h post-intervention, 48h post-intervention, and 72h post-intervention). “*” indicates that there was a significant difference between the CI and the CON.

The Maximum Displacement of COP in the AP

As shown in Table 10. Repeated measurement ANOVA was performed for the maximum displacement of COP in AP. The interaction item group * time met the spherical test (P = 0.053), and the interaction was significant, F (5, 35) = 4.110, P = 0.005. Therefore, separate effect tests for group and time factors are needed further. After a simple effect analysis of group factors, it was found that the influence of group factors on the maximum displacement of COP in AP was not statistically significant (P = 0.407). A simple effect analysis of the time factor showed that the time factor in the CI did not meet the spherical hypothesis (P = 0.016). After greenhouse-geisser correction, the influence of the time factor on the maximum displacement of COP in the AP was statistically significant (P = 0.015, F (5, 35) = 4.966). It was found that the maximum displacement of COP in AP at post-warm-up in the CI was less than post-intervention, and there was a significant difference (P = 0.023). The time factor in the CON met the spherical hypothesis (P = 0.195), and the intrasubjective effect test showed that the time factor in the CON had no statistical significance on the maximum displacement of COP in the AP (P = 0.053, F (5, 35 = 2.449)).

Table 10. The COP maximum displacement of the two interventions at each moment in AP.

maximum displacement (%)	①CI		②CON			
	Mean ± SD	**95%CI**	**Mean ± SD**	**95%CI**	**P-value**	**Δ (①-②)**
Post-warm-up	16.13±1.46	[14.906-17.344]	16.13±2.64	[13.916-18.334]	1	0
Post fatigue	25.25±7.74	[18.780-31.717]	23.72±5.92	[18.770-28.673]	0.457	1.527
Post-intervention	21.70±2.91a	[19.269-24.128]	21.68±4.02	[18.315-25.037]	0.988	0.022
24h post-intervention	20.51±4.39	[16.838-24.178]	20.72±4.34	[17.088-24.349]	0.903	-0.211
48h post-intervention	20.10±4.36	[16.456-23.753]	21.18±7.00	[15.324-27.036]	0.752	-1.076
72h post-intervention	17.34±2.32	[15.395-19.280]	20.06±4.87	[15.995-24.134]	0.211	-2.728
RM ANOVA	Whether the spherical hypothesis is satisfied? Yes (P = 0.053)		F (5, 35) = 4.110		The interaction was significant (P = 0.005)	

Note: “a” indicates that there is a significant difference between the moment of post-warm-up and other moments (post fatigue, post-intervention, 24h post-intervention, 48h post-intervention, and 72h post-intervention).

The Displacement Velocity of COP in the ML

As shown in Table 11. The displacement velocity of COP in ML was analyzed by repeated measurement ANOVA. The interaction group * time met the spherical test (P = 0.067), and the interaction between the two groups was not statistically significant, F (5, 35) = 0.968, P = 0.45. Therefore, separate effect tests for group and time factors are needed further. If the principal effect of factors within the study subjects is greater than two levels, subsequent pairwise comparisons are required. Since the group factor has only two levels, there is no need to test whether the spherical hypothesis is met. The principal effect of group factors on the displacement velocity of COP in the ML was not statistically significant, F (1, 7) = 0.033, P = 0.860. The principal effect of the time factor on the displacement velocity of COP in the ML was not statistically significant, F (5, 35) = 2.227, P = 0.073. The displacement velocity of COP in the CI on the ML was 1.098 (95%CI: –15.325 ~ 13.128) mm/s smaller than the CON, but the difference was not statistically significant.

Table 11. The COP displacement velocity of the two interventions at each moment in ML.

displacement velocity (mm/s)	**①CI**		**②CON**			
	Mean ± SD	**95%CI**	**Mean ± SD**	**95%CI**	**P-value**	**Δ(①-②)**
Post-warm-up	579.72±65.48	[524.979-634.468]	593.50±71.92	[533.378-653.630]	0.86	-1.098
Post fatigue	772.05±127.01	[665.861-878.229]	772.95±116.02	[675.954-869.937]		
post-intervention	774.56±127.33	[668.112-881.007]	793.34±125.11	[688.740-897.934]		
24h Post-intervention	785.67±218.65	[602.879-968.470]	773.72±203.16	[603.875-943.560]		
48h Post-intervention	791.77±298.87	[541.907-1041.626]	791.48±322.49	[521.874-1061.094]		
72hPost intervention	724.57±117.08	[626.687-822.454]	709.94±128.52	[602.491-817.389]		
RM ANOVA	Whether the spherical hypothesis is satisfied? Yes (P = 0.067)		F (5, 35) = 0.968		The interaction was no significant (P = 0.45)	

The Displacement Velocity of COP in the AP

As shown in Table 12. The displacement velocity of COP in AP was analyzed by repeated measurement ANOVA. The interaction group * time met the spherical test (P = 0.704), and the interaction was not statistically significant, F (5, 35) = 1.326, P = 0.276. Therefore, separate effect tests for group and time factors are needed further. If the principal effect of factors within the study subjects is greater than two levels, subsequent pairwise comparisons are required. Since the group factor has only two levels, there is no need to test whether the spherical hypothesis is met. The principal effect of group factors on the displacement velocity of COP in the AP was not statistically significant, F (1, 7) = 0.273, P = 0.618. The principal effect of the time factor on the displacement velocity of COP in AP was not statistically significant, F(5, 35) = 2.106, P = 0.088. The displacement velocity of COP in AP in the CI was 1.395 (95%CI: –4.922 ~ 7.712) mm/s higher than the CON, and the difference was not statistically significant.

Table 12. The COP displacement velocity of the two interventions at each moment in AP.

displacement velocity (mm/s)	**① CI**		**②CON**			
	Mean ± SD	**95%CI**	**Mean ± SD**	**95%CI**	**P-value**	**Δ(①-②)**
Post-warm-up	633.411±27.647	[568.037-698.784]	625.903±34.089	[545.296-706.510]	0.618	1.395
Post fatigue	809.653±46.837	[698.901-920.406]	815.240±43.338	[712.763-917.717]		
post-intervention	821.016±45.384	[713.700-928.332]	814.846±45.505	[707.243-922.449]		
24h Post-intervention	825.317±80.098	[635.916-1014.718]	844.387±87.485	[637.519-1051.255]		
48h Post-intervention	819.955±94.528	[596.432-1043.477]	812.482±91.238	[596.739-1028.225]		
72h Post-intervention	762.436±52.411	[638.505-886.367]	750.560±48.440	[636.019-865.101]		
RM ANOVA	Whether the spherical hypothesis is satisfied? Yes (P=0.704)		F (5, 35)=1.326		The interaction was no significant (P=0.276)	

(2) Dynamic Balance Recovery

As shown in Table 13 and Figure 25. The dynamic balance was analyzed by repeated measurement ANOVA. The interaction item group * time met the spherical test (P = 0.198), and the interaction was significant, F (5, 35) = 15.004, P < 0.001. Therefore, separate effect tests for group and time factors are needed further. After simple effect analysis of group factors, it was found that the group factors 24 h post-intervention had a significant impact on dynamic balance ability, and the score of dynamic balance ability of the CI was higher than CON, with significant differences ($P = 0.004$, $F(1, 7) = 18.142$). The group factors at 48 h post-intervention had a significant influence on dynamic balance ability. The score of dynamic balance ability in the CI was higher than CON, and there were significant differences ($P = 0.002$, $F(1, 7) = 21.284$). At 72 h post-intervention, the group factors had a significant impact on dynamic balance. The score of dynamic balance in the CI was higher than CON, and there were significant differences ($P = 0.001$, $F(1, 7) = 27.354$). After the simple effect analysis of the time factor, it was found that the CI did not meet the spherical hypothesis ($P = 0.001$). After greenhouse-geisser correction, the influence of the time factor on the dynamic balance was statistically significant, $F(5, 35) = 46.508$, $P < 0.001$. Pairwise comparisons are required at six more time points. In the CI, post-fatigue ($P < 0.001$), post-intervention ($P < 0.001$), 24 h post-intervention ($P < 0.001$), 48 h post-intervention ($P = 0.001$), 72 h post-intervention ($P = 0.016$) and the scores of dynamic balance post-warm-up were significantly different. There were significant differences between post-fatigue and post-intervention ($P = 0.046$) and 72 h post-intervention ($P = 0.009$). There were significant differences between post-fatigue and 48 h post-intervention ($P = 0.005$) as well as 72 h post-intervention ($P = 0.001$). The moment of 24 h post-intervention, 48 h post-intervention ($P = 0.001$), and 72 h post-intervention ($P < 0.001$) there were significant differences. Besides, there was a significant difference between 48 h post-intervention and 72 h post-intervention ($P = 0.006$). The simple effect analysis of time factors found that the CON met the spherical hypothesis ($P = 0.171$), so the impact of time factors on dynamic balance in the CON was statistically significant. Pairwise comparisons are required at six more time points. In the CON, post fatigue ($P = 0.007$) and post-intervention ($P < 0.001$), 24 h post-intervention ($P < 0.001$), 48 h post-intervention ($P = 0.001$), 72 h post-intervention ($P = 0.001$), and the scores of dynamic balance at post-warm-up were significantly different. There were significant differences between 24 h post-intervention, 48 h post-intervention ($P = 0.010$), and 72 h post-intervention ($P = 0.004$). There was a significant difference between 48 h post-intervention and 72 h post-intervention (P = 0.044).

Table 13. Table of the dynamic balance of the two interventions at each moment.

dynamic balance (%)	**①CI**		**②CON**			
	Mean ± SD	**95%CI**	**Mean ± SD**	**95%CI**	**P-value**	**Δ(①-②)**
Post-warm-up	98.83±6.69	[93.24-104.42]	99.07±7.38	[92.90-105.24]	0.643	-0.239
Post-fatigue	93.26±6.34a	[87.96-98.56]	92.76±7.13a	[86.80-98.73]	0.232	0.499
post-intervention	90.17±6.87ab	[84.43-95.92]	92.22±6.61a	[86.69-97.75]	0.145	-2.046
24h Post-intervention	93.62±6.92a*	[87.83-99.41]	88.57±6.90a	[82.81-94.34]	0.004	5.049*
48h Post-intervention	95.77±6.94acd*	[89.97-101.57]	91.65±6.55ad	[86.17-97.13]	0.002	4.113*
72h Post intervention	97.46±6.69abcde*	[91.87-103.05]	93.61±6.84ade	[87.89-99.33]	0.001	3.857*
RM ANOVA	Whether the spherical hypothesis is satisfied? Yes (P=0.198)		F (5, 35)=15.004		The interaction was significant (P=0)	

Note: “a” indicates that there is a significant difference between the moment of post-warm-up and other moments (post fatigue, post-intervention, 24h post-intervention, 48h post-intervention, and 72h post-intervention). “b” indicates that there is a significant difference between the moment of post fatigue and other moments (post-intervention, 24h post-intervention, 48h post-intervention, and 72h post-intervention). “c” indicates that there is a significant difference between the moment of post-intervention and other moments (24h post-intervention, 48h post-intervention, and 72h post-intervention). “d” indicates that there is a significant difference between the moment of 24h post-intervention and other moments (48h post-intervention and 72h post-intervention). “e” indicates that there is a significant difference between the moment of 48h post-intervention and the moment of 72h post-intervention. “*” indicates that there was a significant difference between the CI and the CON.

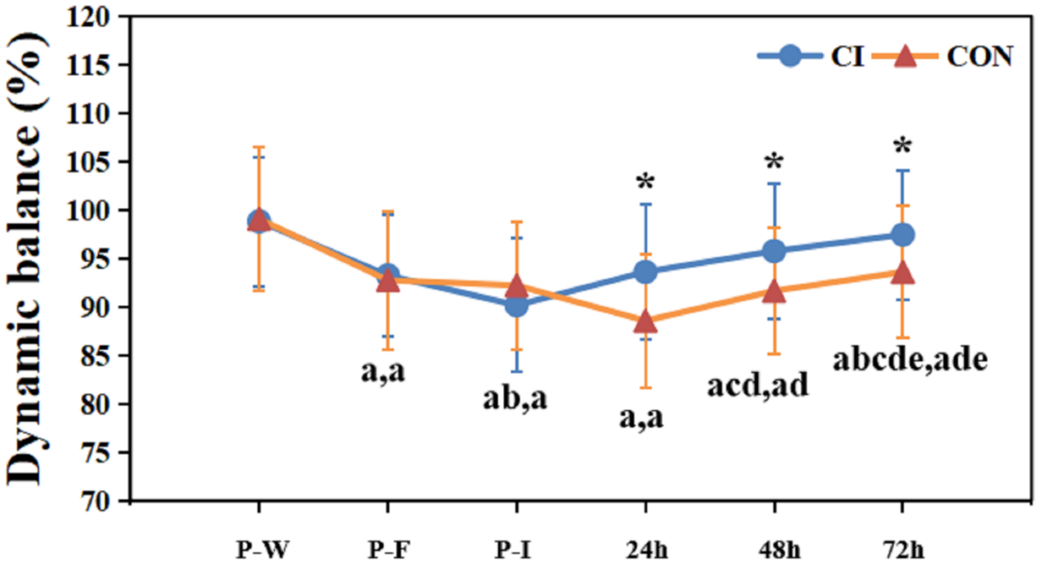

Figure 25. The difference in dynamic balance between CI and CON at each moment.

Note: “a” indicates that there is a significant difference between the moment of post-warm-up and other moments (post fatigue, post-intervention, 24 h post-intervention, 48 h post-intervention, and 72 h post-intervention). “b” indicates that there is a significant difference between the moment of post fatigue and other moments (post-intervention, 24 h post-intervention, 48 h post-intervention, and 72 h post-intervention). “c” indicates that there is a significant difference between the moment of post-intervention and other moments (24 h post-intervention, 48 h post-intervention, and 72 h post-intervention). “d” indicates that there is a significant difference between the moment of 24 h post-intervention and other moments (48 h post-intervention and 72 h post-intervention). “e” indicates that there is a significant difference between the moment of 48 h post-intervention and the moment of 72 h post-intervention. “*” indicates that there was a significant difference between the CI and the CON.

3.2 Results of Musculoskeletal Simulation

3.2.1 Model Validation

As shown in Figure 26, the lower limb muscle sEMG signals and activations from OpenSim optimization at the chasse step and one-step footwork during stroke were compared. Including biceps femoris long head (biceps femoris lh), lateral gastrocnemius, medial gastrocnemius, rectus femoris, semitendinosus, tibialis anterior, vastus lateralis, and vastus medialis. According to Figure 27, The lower lime muscle activation of the chasse step and one-step footwork during stroke calculated by the musculoskeletal model was similar to the surface sEMG signal recorded in the experiment, which indicated that the OpenSim model data in this study was relatively reliable.

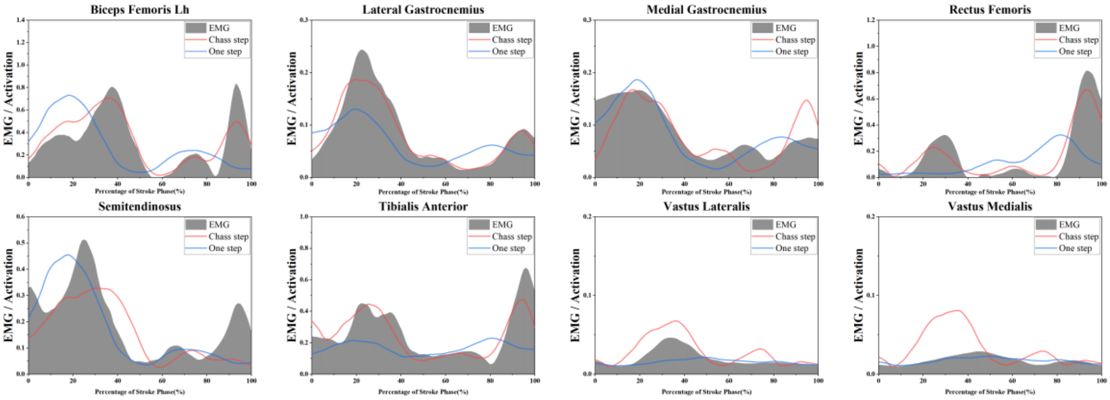

Figure 26. Comparison of lower limb muscle sEMG signals and activations from OpenSim Optimization between the chasse step and one step during stroke in table tennis.

3.2.2 Kinematics and Dynamics of Hip, Knee, and Ankle in Footwork

Figure 27 shows the SPM1d analysis result of the lower limb joint angle of the chasse step and one-step footwork during the stroke. Figure 28 shows the SPM1d analysis result of a joint moment of the chasse step and one-step footwork during the stroke. The ankle plantarflexion joint angle and moment of one-step footwork were significantly higher than chasse step footwork in the 50.51%–87.75% ($p < 0.001$), 55.75%–87.00% ($p < 0.001$) stroke phase, respectively. The subtalar joint valgus angle and moment of in the one-step footwork were significantly higher than the chasse step footwork in the 18.5%–100% ($p < 0.001$) and 56.71%–86.47% stroke phase ($p < 0.001$), respectively. The knee flexion angle and moment of the one-step footwork were significantly higher than the chasse step footwork in the 42.83%–100% ($p < 0.001$), and 48.57%–87.21% stroke phase ($p < 0.001$), respectively. However, the knee flexion angle and moment in 8.61%–30.19% ($p = 0.008$) and 2.82%–40.97% ($p < 0.001$) stroke phase of the chasse step were significantly higher than the one-step footwork. Besides, the hip flexion angle in the 16.55%–49.23% stroke phase ($p = 0.001$) and the hip extension moment in the 32.63%– 58.37% stroke phase ($p < 0.001$) are significantly greater in the chasse step than the one-step footwork. The chasse step footwork shows a significantly higher angle of hip abduction in the 59.77%–69.80% stroke phase ($p = 0.009$). The chasse step footwork hip external rotation angle in 48.20%–73.85%, 93.05%–100% stroke phase and moment in 58.84%–77.88%, 86.87%–100% stroke phase was significantly higher than the one-step footwork, respectively.

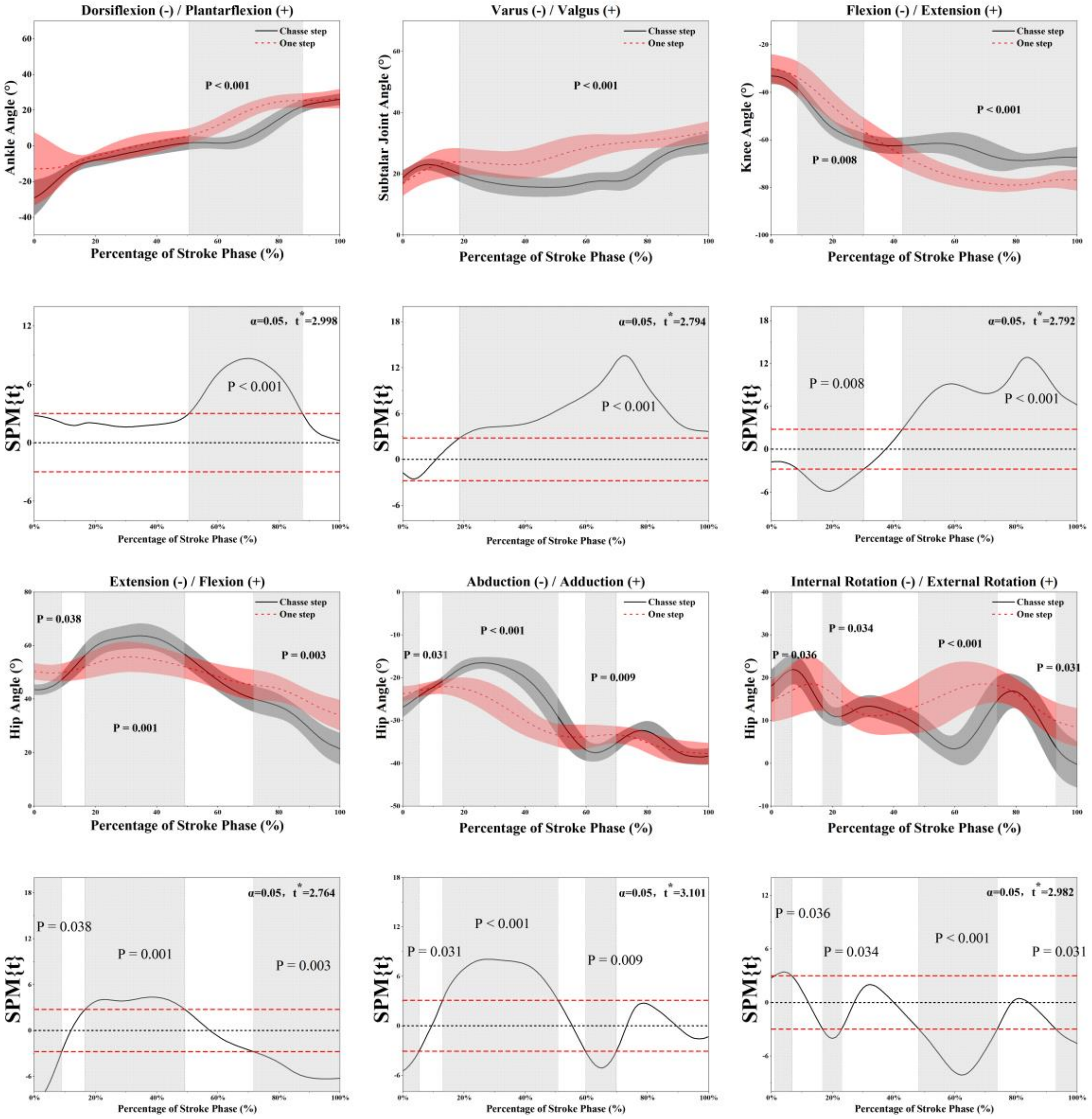

Figure 27. Illustration of the result between the chasse-step and one-step showing the statistical parametric mapping outputs for the lower limb joint angle during the stroke phase.

Note: Grey-shaded areas indicate that there are significant differences ($p < 0.05$) between the chasse-step and one-step.

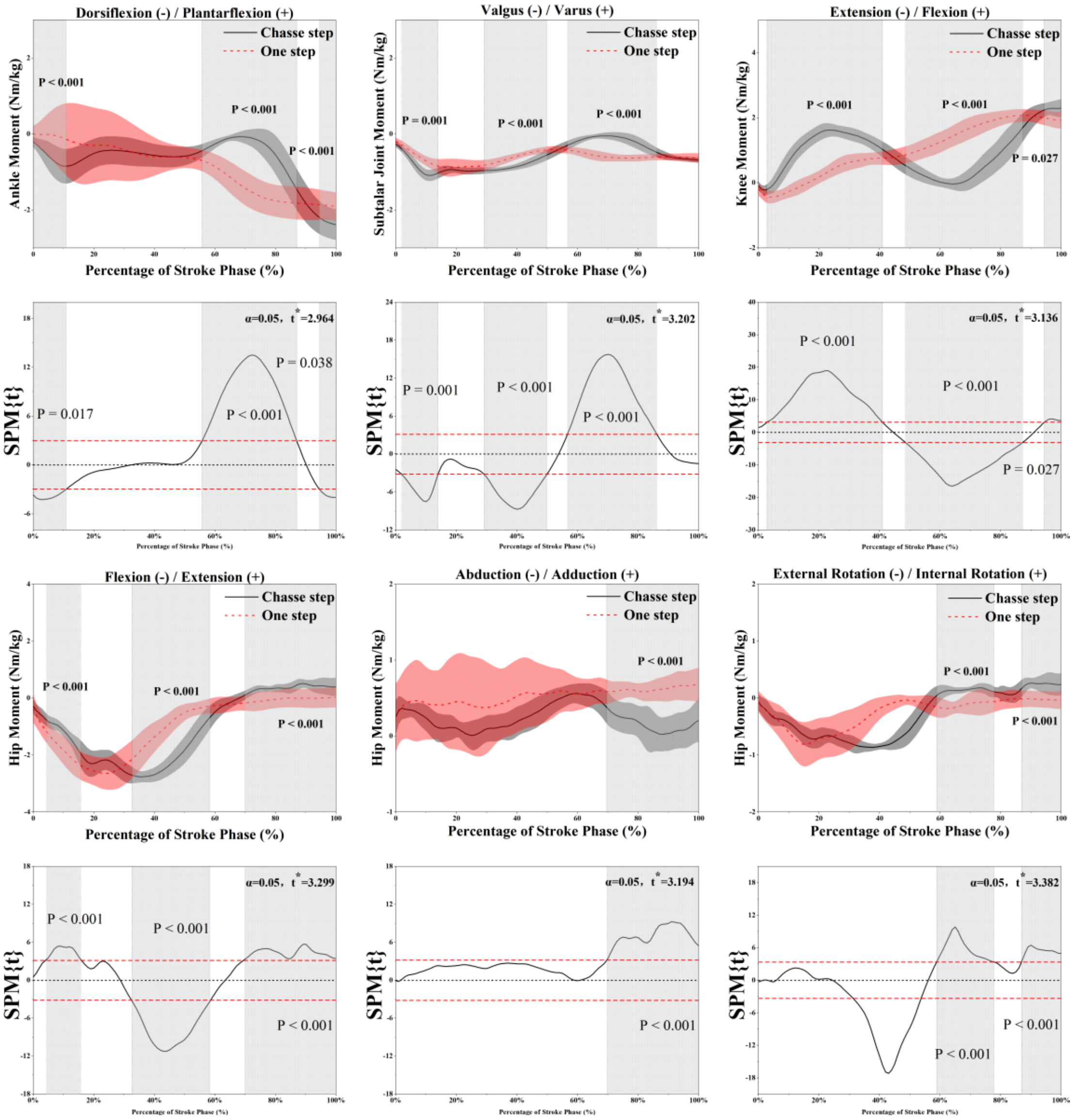

Figure 28. Illustration of the result between the chasse-step and one-step showing the statistical parametric mapping outputs for the lower limb joint moment during the stroke phase.

Note: Grey-shaded areas indicate that there are significant differences ($p < 0.05$) between the chasse-step and one-step.

3.2.3 Kinematics and Dynamics of Lumbar during Stroke Play

Table 14 and Figure 29 show the SPM1d analysis result of the angle and moment in the LAR, LLB, and LF between the CC and LL topspin forehand. In the LAR, the LL showed a significantly higher moment than CC in the 0%–1.75% ($p = 0.045$, $t = 3.331$)

and 3.80%–28.14% ($p < 0.001$, $t = 3.331$) phase, and a significantly higher angle in 3.30%– 22.79% ($p < 0.001$, $t = 3.129$) phase. However, the LL showed a significantly lower moment and angle in the 34.51%–57.98% ($p < 0.001$, $t = 3.331$) and 30.29%–60.43% ($p < 0.001$, $t = 3.129$) phase than CC, respectively. In the LLB, CC showed a significantly higher moment than LL in the 27.93%–49.48% ($p < 0.001$, $t = 3.258$), 55.56%–72.87% ($p < 0.001$, $t = 3.258$), 97.38%–100% ($p = 0.043$, $t = 3.258$) phase, and a significantly higher angle in the 12.29%– 75.30% ($p < 0.001$, $t = 3.125$) and 85.68%–100% ($p = 0.004$, $t = 3.125$) phase. The LL showed a significantly higher moment in the 1.30%– 19.81% ($p < 0.001$, $t = 3.258$) phase than CC. In the LF, the moment of LL was significantly higher than CC in the 6.38%–29.08% ($p < 0.001$, $t = 3.344$) and 90.29%–99.25% ($p = 0.003$, $t = 3.344$) phase, and the angle were higher than CC in the 55.13%–100% ($p < 0.001$, $t = 3.08$) phase. The CC showed a significantly higher moment in the 0%– 2.52% ($p = 0.04$, $t = 3.344$), 37.81%–58.06% ($p < 0.001$, $t = 3.344$), and 63.86%–76.60% ($p < 0.001$, $t = 3.344$) phase, and a significantly higher angle in the 5.26%–10.63% ($p = 0.038$, $t = 3.080$) and 18.40%–39.10% ($p = 0.001$, $t = 3.080$) phase than LL.

As shown in Figure 30, the Rom and peak moment of LLB and LF in CC were significantly higher than LL (Rom: $t = 16.55$, $p = 0$; $t = 12.139$, $p = 0$. Peak moment: $t = -3.396$, $p = 0.002$; $t = 3.412$, $p = 0.003$). The maximum LAR, LLB, and LF in the CC were significantly higher than LL ($t = -2.84$, $p = 0.008$; $t = 13.206$, $p = 0$; $t = -3.307$, $p = 0.003$).

Table 14. The moment and angle results of the SPM1d analysis. (Unit: %)

Variables	**Percentage (*p*)**
LAR Moment	0-1.75 (0.045), 3.80-28.14 (<0.001), 34.51-57.98 (<0.001)
LLB Moment	1.30-19.81 (<0.001), 27.93-49.48 (<0.001), 55.56-72.87 (<0.001), 97.38-100 (0.043)
LF Moment	0-2.52 (0.04), 6.38-29.08 (<0.001), 37.81-58.06 (<0.001), 63.86-76.6 (<0.001), 90.29-99.25 (0.003)
PAR Moment	4.15-30.01 (<0.001), 45.01-80.63 (<0.001),
LAR Angle	3.30-22.79 (<0.001), 30.29-60.43 (<0.001)
LLB Angle	12.29-75.30 (<0.001), 85.68-100 (0.004),
LF Angle	5.26-10.63 (0.038), 18.40-39.10 (0.001), 55.13-100 (<0.001)
PAR Angle	0-1.69 (0.049), 10.29-78.31 (<0.001), 88.90-100 (0.027)

Note: the percentage indicates the process of the stroke play phase.

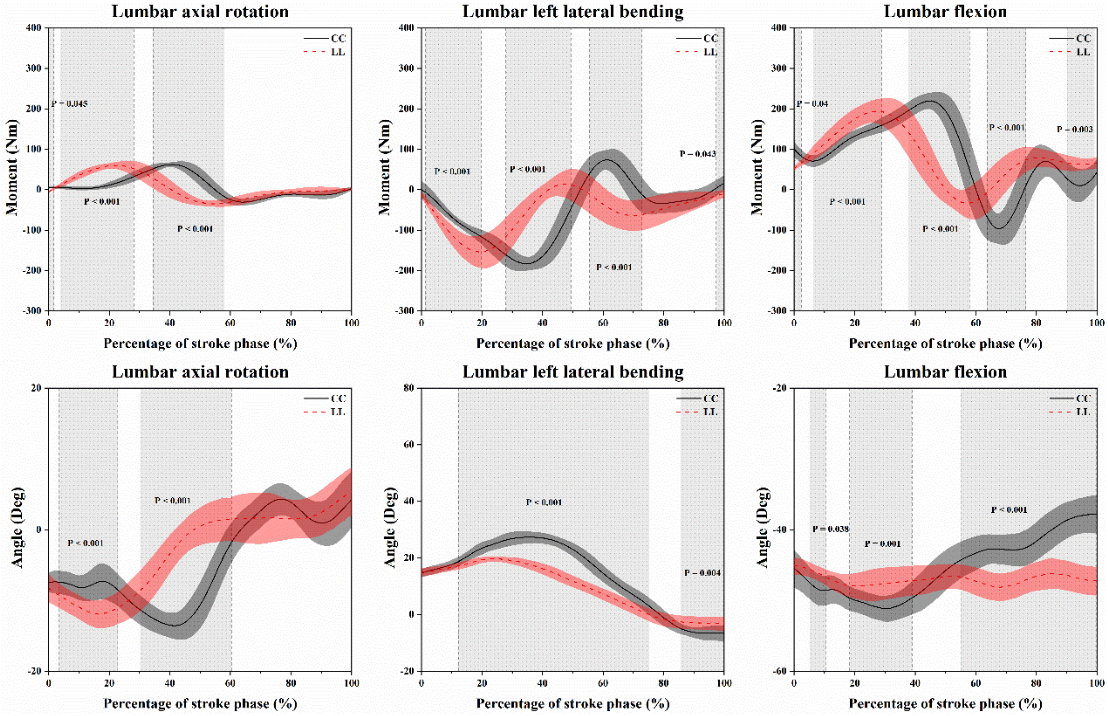

Figure 29. Illustration of the result of the angle and moment in the LAR, LLB, and LF between the CC and LL topspin forehand showing the SPM1d outputs.

Note: Grey-shaded areas indicate that there are significant differences ($p < 0.05$) between the CC and LL.

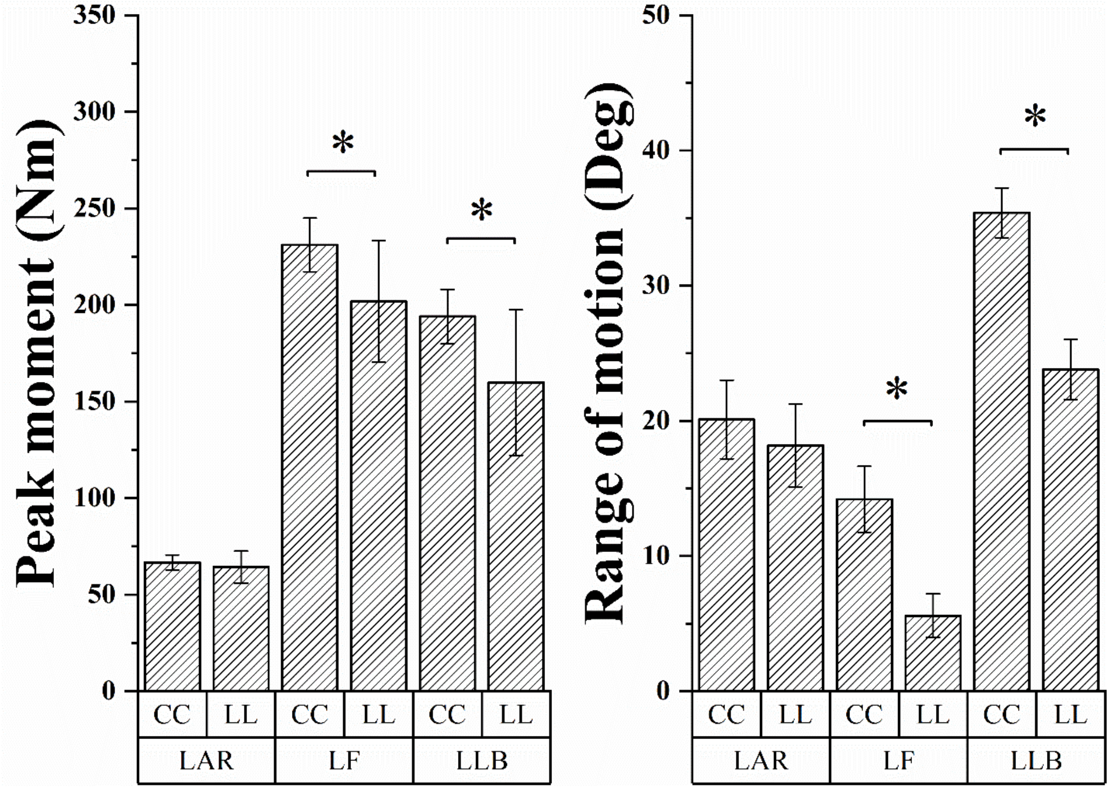

Figure 30. The Rom and peak moment comparison of lumbar movement between the CC and LL topspin forehand.

Note: "*" indicates a significant difference between LL and CC.

3.2.4 Kinematics and Dynamics of the Pelvis during Stroke Play

Figure 31 shows the SPM1d analysis result of the angle and moment of PAR between the CC and LL topspin forehand. The PAR angle of CC was significantly higher than LL in the 10.29%– 78.31% ($p < 0.001$, $t = 2.86$) and 88.90%–100% ($p = 0.027$, $t = 2.86$) phase, but significantly lower than LL in 0%–1.69% ($p = 0.049$, $t = 2.86$) phase. The PAR moment of CC was significantly higher than LL in the 4.15%–30.01% ($p < 0.001$, $t = 3.288$) phase and significantly lower than LL in the 45.01%–80.63% ($p < 0.001$, $t = 3.288$) phase.

The maximum PAR in the CC was significantly higher than LL ($t = -9.627$, $p = 0$), and Rom and peak moment of PAR in the CC was significantly higher than LL ($p = 0$, $t = 12.798$; $p = 0.034$, $t = 2.245$).

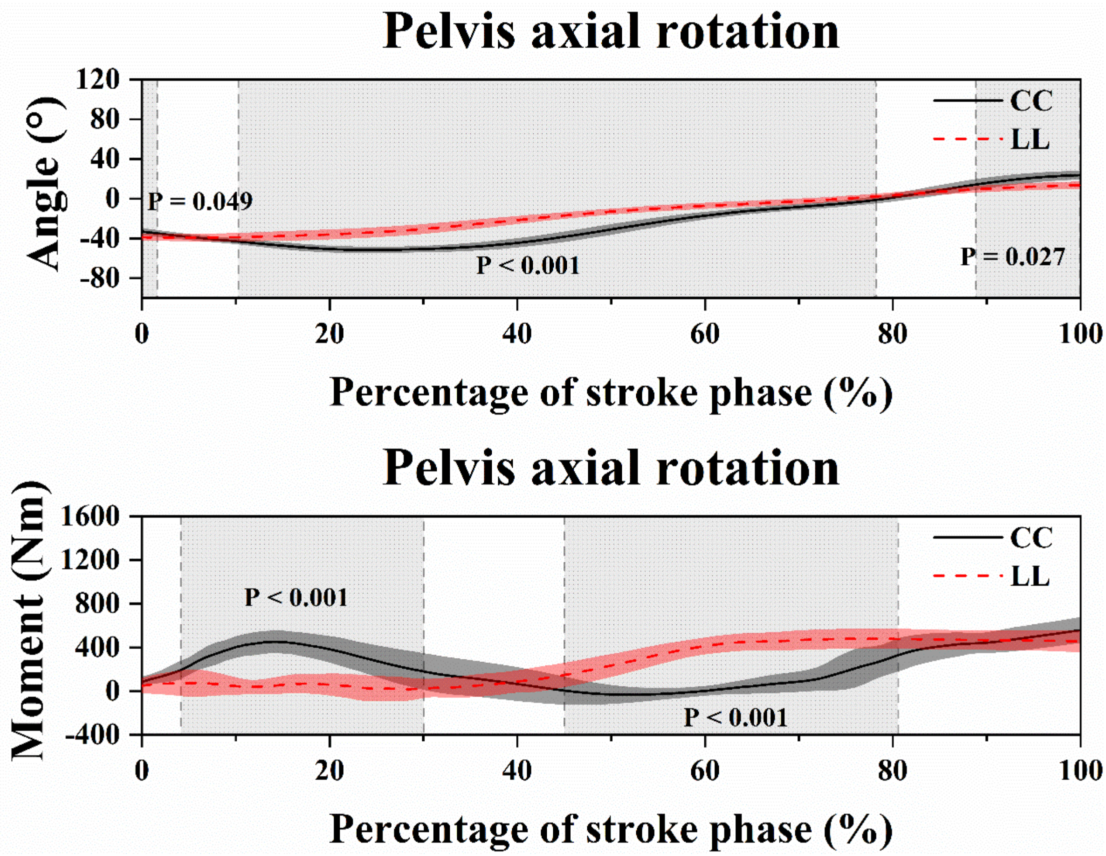

Figure 31. Illustration of the result of angle and moment of the PAR between the CC and LL topspin forehand showing the SPM1d outputs.

Note: Grey-shaded areas indicate that there are significant differences ($p < 0.05$) between the CC and LL.

3.2.5 Muscle Force

Figure 32 shows the SPM1d analysis result of muscle force of the chasse step and the one-step footwork during the stroke. The muscle force of the biceps femoris in the chasse step footwork was significantly greater than the one-step footwork in the 24.42%–50.87% ($p < 0.001$) and 88.04%–100% ($p < 0.001$) stroke phase. The muscle force of the lateral gastrocnemius in the chasse step footwork in 9.52%–47.54% ($p < 0.001$) and 86.89%–99.68% ($p < 0.001$) stroke phase as well as medial gastrocnemius muscle force in 28.55%– 51.56% ($p < 0.001$) and 89.67%–100% ($p < 0.001$) stroke phase is significantly greater than one step. The muscle force of the medial gastrocnemius and the lateral gastrocnemius of the one-step footwork was significantly greater than the chasse step footwork in 63.75%–81.78% and 61.13%–82.29% stroke phase, respectively. Lateral vastus muscle forces in the 8.67%–43.45% ($p < 0.001$) and medial vastus muscle forces in the 8.40%–44.10% ($p < 0.001$) stroke phase in chasse step were significantly greater than the one-step footwork. The rectus femoris muscle force in the 16.54%–34.49% ($p < 0.001$) and 87.14%– 100% ($p < 0.001$) stroke phase in the chasse step was significantly greater than one-step, but significantly less than one-step footwork in the 44.14%–56.65% ($p < 0.001$) and 65.88%–80.87% ($p < 0.001$) stroke phase. The semitendinosus muscle force in the chasse step was significantly greater than one-step footwork during the 29.45%–52.01% ($p < 0.001$) stroke phase. The tibialis anterior muscle force in the chasse step during 14.93%–39.56% ($p < 0.001$), 51.43%–65.58% ($p < 0.001$), and 88.74%–100% ($p < 0.001$) stroke phase was significantly greater than one-step, but significantly less than one-step footwork in the 71.24%–82.61% ($p < 0.001$) stroke phase.

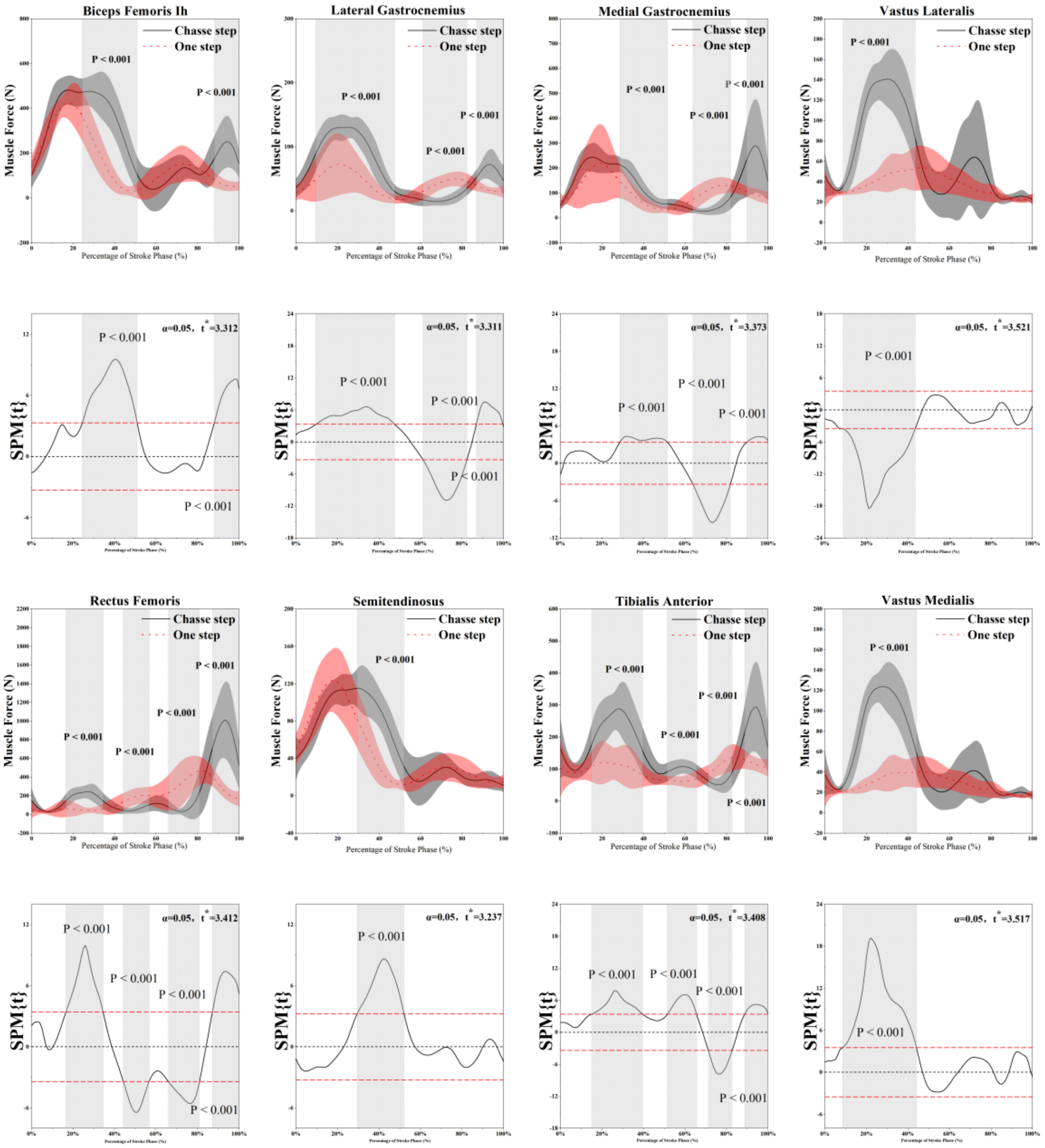

Figure 32. Illustration of the results between the chasse-step and the one-step showing the statistical parametric mapping outputs for the lower limb muscle force during the stroke phase.

Note: Grey-shaded areas indicate that there are significant differences ($p < 0.05$) between the chasse-step and one-step.

3.2.6 Joint Stiffness

Table 15 showed the joint stiffness of the lower extremity in CS and OS during the landing stage in table tennis stroke play. The stiffness of the knee and the transverse plane of the hip in the CS was significantly greater than in the OS (Knee: CS = 0.072 ± 0.011 $N \cdot M \cdot KG^{-1} \cdot Deg^{-1}$, OS = 0.038 ± 0.010 $N \cdot M \cdot KG^{-1} \cdot Deg^{-1}$, $t = 9.487$, $p = 0$; Hip: CS = 0.119 ± 0.037 $N \cdot M \cdot KG^{-1} \cdot Deg^{-1}$, OS = 0.073 ± 0.027 $N \cdot M \cdot KG^{-1} \cdot Deg^{-1}$, $t = 3.934$, $p = 0$).

3.2.7 Joint Reaction Force

The reaction force of the knee, ankle, subtalar, and hip in CS was significantly greater than in the OS. In the knee, the JRF in CS was significantly greater than in OS in the percentage of 30.89-35.83% ($p = 0.035$) and 36.19-100% ($p < 0.001$) but lower than in OS in the percentage of 8.74-14.05% ($p = 0.034$). In the ankle, the JRF in CS was significantly greater than in OS in the percentage of 24.32-30.30% ($p = 0.027$) and 40.76-100% ($p < 0.001$). In the subtalar, the JRF in CS was significantly greater than in OS in the percentage of 24.24-30.15% ($p = 0.028$) and 41.14-100% ($p < 0.001$). In the hip, the JRF in CS was significantly greater than in OS in the percentage of 40.94-100% ($p < 0.001$).

Table 15. The joint stiffness of the lower extremity of CS and OS during the landing stage in table tennis stroke play.

		Angle	**Δ (angle)**	**Moment**	**Δ (moment)**	**Stiffness**	***t***	***95%CI***	***p***
Knee	CS	-60.193±3.988	30.611±4.403	1.850±0.190	2.179±0.222	0.072±0.011	9.487	0.027, 0.042	0
	OS	-54.157±6.863	41.025±5.325	1.280±0.190	1.528±0.355	0.038±0.010			
Ankle	CS	2.589±6.308	28.645±5.943	2.270±0.206	2.676±0.255	0.097±0.023	-0.658	-0.042, 0.218	0.519
	OS	4.85±2.045	22.929±11.908	1.594±0.185	1.843±0.583	0.107±0.055			
Subtalar	CS	25.031±2.657	11.747±4.595	-2.149±0.166	2.580±0.447	0.242±0.078	-1.959	-0.204, 0.012	0.059
	OS	18.155±4.671	9.014±5.938	-1.744±0.225	2.329±0.651	0.338±0.185			
Hip x	CS	40.188±4.275	17.972±4.535	0.974±0.160	1.326±0.141	0.077±0.017	-1.234	-0.082, 0.024	0.227
	OS	36.800±4.817	10.664±2.920	0.493±0.128	0.925±0.193	0.106±0.095			
Hip y	CS	-20.657±3.027	16.076±3.473	-1.084±0.125	1.435±0.272	0.094±0.031	-1.968	-0.051, 0.001	0.058
	OS	-29.809±2.814	13.370±2.020	-1.151±0.102	1.533±0.344	0.119±0.040			
Hip z	CS	1.105±2.501	11.259±2.49	1.178±0.141	1.261±0.200	0.119±0.037	3.934	0.022, 0.069	0
	OS	-4.235±6.400	11.937±4.803	0.644±0.099	0.770±0.143	0.073±0.027			

4 Discussions

4.1 Lower Extremity Injury Mechanism and Prevention

4.1.1 Foot and Plantar Injury Prevention

With the development of biomechanical measurement methods and techniques, biomechanical research of the lower limbs during table tennis has received extensive attention in recent years. The exploration of the lower limb kinetic mechanisms of footwork in table tennis can provide a theoretical basis for the optimization of the lower limb dynamic chain, the prevention of sports injury, and a contribution to the development of table tennis shoes. The purpose of this study was to investigate the differences in lower limb kinetic characteristics between the chasse step and one-step footwork during stroke play in table tennis. The key findings of this study were that: (1) In 6.92–11.22% of the BP, the one step showed greater plantar force than the chasse step, and in 53.47–99.01% of the BP, the chasse step showed greater plantar force than the one step, which means that the one step showed greater plantar force on landing and that the chasse step showed a better force accumulation effect in the BP. In 21.06–84.06% of the FP, the chasse step showed greater plantar force than the one step, indicating better lower limb drive. (2) The one step was observed to have a higher maximum plantar force in the BP and a lower maximum plantar force in the FP compared with the chasse step. (3) The one step showed greater peak pressure in the MR, LR, and LF regions than the chasse step in the BP. In the FP, the chasse step showed a greater peak pressure in the T than the one step. (4) The one step showed lower FTI and greater PTI than the chasse step during the BP, and the chasse step showed greater FTI and PTI than the one step in the FP.

The foot and plantar biomechanical characteristics of forehand topspin have been extensively studied and reported in recent years, and this information is generally considered to be related to lower limb driveability [17, 32, 39] and the origin of the kinetic chain [20]. Peak pressure, plantar force, COP displacement, COP velocity, and contact area, are the basic parameters of plantar biomechanical research. Compared with the one step, the chasse step showed greater plantar force in the 53.47–99.01% process of BP, and a greater plantar force during the 21.06–84.06% process of FP, and a higher maximum plantar force in FP, as well as a greater peak pressure in the T. This, means that the chasse step shows greater complete lower limb extension and drive during the FP. It appears that the greater energy transfer promotes the generation of momentum [16, 57]. As the origin of the dynamic chain, the lower limbs transfer the optimal activation energy from the lower limbs to the upper limbs through the continuous movement of the dynamic chain [17, 12]. Lam et al. (2018) have investigated the biomechanical differences between different footwork during the topspin forehand in table tennis [15]. In their study, the significantly higher peak pressures were in the plantar region of the total foot, toe, 1st, 2nd, and 5th metatarsal during the chasse step and one step compared with the one-step. The chasse step also showed a higher peak pressure than one step in the toe area. This is consistent with the results of this study. However, the MR, LR, and LF observed a higher peak pressure in the one step than the chasse step in this study, and this is not consistent with the results of Lam et al. (2018). This may be due to the different movement distances of the footwork resulting in different momentums resulting in different force values during landing. The chasse step showed higher peak pressure in the T than the one step. This could mean more plantarflexion during chasse step footwork in the FP. This may contribute to a greater range of weight transfer and thus momentum generation [16, 57, 58]. Previous studies have reported on the underlying mechanisms of lower limb energy transfer and racket speed [1, 20, 58]. In this study, the chasse step showed significantly greater plantar force than the one step in the 21.06–84.06% process of FP. From a practical point of view, athletes can enhance the plantarflexion function to bring greater weight transfer, resulting in a greater momentum during the 21.06–84.06% process of FP, thus improving the performance of racket speed. Also from the perspective of sports monitoring, the quality of strokes during one-step footwork can be monitored by analyzing the plantar force curves of players in the 21.06–84.06% process of FP.

PTI is a variable used to evaluate plantar load. This variable describes the cumulative effect of pressure over time in a certain area of the plantar. Excessive values may lead to tissue damage [59, 60]. FTI is a variable that considers the integral of force over time in a plantar area. PTI is the quotient of FTI divided by the contact area, which will provide an average cumulative load per square centimeter. PTI is better associated with plantar tissue injury than FTI [60]. In this study, the one step shows greater PTI than the chasse step during landing in the BP, and larger peak pressure was shown in MR, LR, and LF. This may have resulted in the center of gravity of the body being transferred to the dominant leg when landing, as well as being accompanied by the transfer of energy, leading to more load on the dominant leg during landing. Table tennis players rely more heavily on the movement of the dominant leg [15]. Over-repetition coupled with high plantar pressure may result in injuries in athletes [15, 17, 61]. Therefore, the athlete can reduce the load on the dominant leg during landing by practicing a buffer strategy. In addition, according to the results of this study, the design and material selection of table tennis shoes can be considered to enhance the cushioning capacity of the sole heel area and the stiffness of the toe area. The key findings in this study not only provide information for exploring foot injuries of table tennis players but also provide reference information for the design and development of table tennis shoe soles. There are some limitations in the study that should be mentioned. Firstly, this study simulated the competition environment in the laboratory, which may have some differences from real competitions. Secondly, the experiment did not consider the foot morphology of the subjects, and different foot shapes may show different plantar load characteristics under the same footwork. In the future, biomechanical research related to the lower limbs of table tennis players should include the influence of foot morphology on experimental results. Real-time data and more advanced methods and equipment should be used to collect experimental information during a real competition environment.

4.1.2 Muscle and Lower Extremity Joint Injury Prevention

Chasse step and one-step footwork are highly repetitive movements in table tennis that help athletes reach an appropriate position and area to execute topspin forehands [15, 39, 44, 62]. This study uses an individualized OpenSim musculoskeletal modeling to reveal the joint angle, joint moment, and muscle force characteristics of the lower limb joints during a stroke with the chasse step and the one-step footwork, which further investigate the internal mechanism of energy transfer and biomechanical in table tennis footwork and the risk of possible sports injury. The key findings of this study were that the muscle force of the biceps femoris long head, lateral gastrocnemius, vastus lateralis, vastus medial, rectus femoris, and tibia anterior of the chasse step was significantly greater than the one-step footwork during the early stroke phase (stance). At the end of the stroke phase (push-off), the muscle force of the biceps femoris long head, medial gastrocnemius, lateral gastrocnemius, rectus femoris, and tibias anterior in the chasse step footwork was significantly greater than the one-step. The muscle force of the ankle plantar flexor and valgus muscle groups in the one-step was significantly greater than in the chasse step. Besides, the moment and angle of hip flexion and axial rotation were significantly greater for the chasse step than the one-step footwork, as well as the ankle plantarflexion angle and moment of the one-step footwork were significantly higher than the chasse step footwork. The results of this study were consistent with our hypothesis that both footwork had significant differences in muscle strength, joint angles, and moments. This study can provide theoretical guidance for motion control and injury prevention to table tennis players and coaches.

Overall, the moment and angle of hip flexion and axial rotation were significantly greater for the chasse step than for the one-step footwork, which is consistent with previous studies [15, 23]. Compared to one-step footwork, the chasse step requires a greater distance and is accompanied by a full-body weight transfer. Hip flexion and axial rotation moments add to the racket's maximum acceleration by the kinetic chain that follows the proximal-to-distal segmental sequences [15, 18, 46], and the transfer of whole-body weight creates greater energy transfer, further enhancing the stroke impact. The ankle plantarflexion angle and moment of the one-step footwork were significantly greater than those of the chasse-step footwork, which was inconsistent with previous studies [15, 22], and the possible reason was that the difference in the distance traveled in the one-step footwork and the difference in ball speed resulted in athlete adjustment different motor control strategies of the lower limb. The muscle force of the biceps femoris long-head, lateral gastrocnemius, rectus femoris, and tibia anterior of the chasse step is significantly greater than the one-step footwork during the early stroke phase (stance). At the end of the stroke phase (push-off), the muscle force of the biceps femoris long-head, medial and lateral gastrocnemius, rectus femoris, and tibias anterior in the chasse step footwork is significantly greater than the one-step footwork. This may be because the chasse step footwork is to quickly reach the target area to complete the stroke movement by transferring the weight of the whole body through the large movement of the full feet, which results in a more powerful load on landing, so the lower limb muscles are required to provide more powerful muscle force to maintain the stability of the joints and the quality of the movements, this finding consistent with previous studies [22, 32, 40]. The large-scale transfer of whole-body weight leads to a large-scale transfer in the center of gravity, so at the end of the stroke phase, the lower limb muscles need to provide stronger muscle force to prepare for the next transfer in the center of gravity. A high-quality hitting movement through continuous movement in all directions in a limited area, and the stronger lower limb driving force provide the basis for this rapid and frequent change of directional movement, which is consistent with previous research [43]. Besides, to perform a high-rotation, low-height forehand topspin, the player must bend the knee strongly [21], suggesting important contributions from knee flexor and extensor muscle groups such as the biceps femoris, and rectus femoris. Racquet athletes rely more on the movement of the dominant leg [15]. This can lead to a high muscle asymmetry degree in the extremities of racket athletes, and greater asymmetry of muscle can disrupt the movement rhythm and increase the sports injury risk [15, 63]. The greater knee flexion angle and moment in the chasse step footwork during the early stroke phase (stance) means that is more effective in joint rotation which promotes energy accumulation. Greater knee flexion allows athletes to better utilize energy transfer throughout the kinematic chain to achieve a proper velocity of the racket based on the proximal-to-distal segmental sequences [45]. The agreed result has been reported in a previous study [15]. Also, according to Figure 29, the change in knee moment for the chasse step footwork shows a faster rate of energy transfer, which may also explain the flexibility of the chasse step relative to the one-step footwork. However, attention should also be paid to the risk of sports injuries caused by excessive knee fatigue that may result from prolonged training with the chasse step footwork. The larger loading magnitudes on the ankle and knee joint found in the chasse step and one-step footwork during the topspin forehand would predispose table tennis athletes to overuse conditions such as jumper knee [64] and ankle sprain injuries [65].

Combine Figure 28, 29, and 34, we can observe that the one-step footwork was significantly higher than the chasse step footwork on the parameters of the joint angle and joint moment of plantar flexion and valgus, as well as the muscle force of the plantar flexor and valgus muscle groups. It means that greater muscle force of the medial and lateral gastrocnemius in the one-step footwork provides greater plantarflexion moments and angles. This may explain why gastrocnemius muscle activity is higher when the heel takes off due to plantar flexion during stroke [66, 67]. Besides, the significantly higher knee flexion angle was observed in one-step footwork, greater knee flexion, and greater plantarflexion allow athletes to better utilize energy transfer throughout the kinematic chain to achieve a proper velocity of racket based on the proximal-to-distal segmental sequences [47].

However, there are a few limitations to this study. Firstly, joint forces were not examined in this study. In future research, the joint contact forces of the knee and ankle joints during forehand topspin stroke need to be detected and considered to accurately diagnose and prevent sports injuries. Secondly, this study did not detect the racket velocity, upper limb movement, and the velocity of the ball shot by the coach. In future research, we will consider combining the racket movement with the biomechanical information of the whole-body joints and muscle force information to further explore the key information of table tennis movement. Thirdly, there are only male participants in this research, so the result of this research hasn’t considered the gender factor. Finlay, the sample size of this research has to be mentioned. There was a total of six national-level players who joined this study, finding a large number of athletes of the same level was not easy. Due to the limitation of the sample size, the practical application of the results of this study may require further support from future studies.

4.1.3 Lumbar and Pelvis Injury Prevention

This study simulated the musculoskeletal model in OpenSim to investigate the lumbar and pelvis movement difference between the CC and LL topspin forehand in table tennis. The key finding of this study was the main difference between CC and LL topspin forehand in the lumbar movement was found in the LLB and LF, the Rom, peak moment, and maximum angle of the LLB and LF in CC were significantly higher than LL; the Rom, peak moment and maximum angle of PAR in CC were significantly higher than LL; the moment of LL in the LF and LLB was significantly higher than CC in the early stroke phase. The results of the current study were consistent with our hypothesis, the CC and LL showed a significant difference in lumbar and pelvis movement in the transverse plane. Investigating the difference in lumbar and pelvis movement between the CC and LL topspin forehand could provide guidelines for coaches and players to understand the mechanisms inherent from a biomechanical perspective, especially the information could help beginners build awareness of CC and LL topspin forehand skills more easily for enhance their stroke skill and motor control. The lumbar movement is widely focused, especially in racket sports. The Rom and maximum angle of the LLB and LF in CC were significantly higher than LL in this study. This could be explained by the fact that the target area in CC is the left side of the playing body, and the players need to adjust their bodies to hit the ball correctly. A higher LLB Rom and maximum angle could bring a completed body weight transfer which could benefit the energy transfer from the trunk to the upper limb following the proximal-to-distal segmental sequences in the kinetic chain [15, 32, 46, 54]. A higher LF in CC probably means a more forward shift of the center of gravity in the sagittal plane, furthermore, the shift in the center of gravity will result in greater energy transfer, which may mean greater racket acceleration during the forward swing phase. In previous studies, LBP (lower back pain, LBP) in athletes of racket sports has been thought to be closely associated with lumbar movement [68-72]. The lumbar section as the main core region of the body plays a coordinating role in the compound movement of the upper and lower extremities, however, this is also a major cause of LBP, because in the topspin forehand motor, the LLB, LAR, and LF have occurred simultaneously, the 'coupled movements' could bring more pressure and load to vertebral structures than the single plane movement [73, 74]. Previous studies have shown that 32% of athletes experience pain in the lumbar and spinal column during competition or immediately after training, and 36% of athletes even quit training due to pain [75]. In the topspin forehand, the athlete's unilateral upper extremity needs to hit the ball with maximum force, and this often leads to full body involvement, increasing the impact of the stroke through a large transfer of full body weight. However, the foot on the non-playing side needs to be locked on the ground to maintain dynamic body balance. Extensive repetition of this compensatory movement leads to severe overload of the posterior side of the disc and causes injury. Further, the significantly greater maximum angle and peak moment of LF and LLB exhibited in CC relative to LL may imply a greater risk of injury.

Extensive research on topspin forehand already exists, but few studies have reported detailed information on pelvis movement during topspin forehand stroke. The result shows that the Rom, peak moment, and maximum angle of PAR in CC were significantly higher than in LL. The ROM value of PAR in this study was basically consistent with the study of Bańkosz and Winiarski (2018) [46] and Malagoli Lanzoni et al. (2018) [76] respectively, this indicates that during the topspin forehand, the players follow a steady motor program and execute it repeatedly, which may be gradually fixed and standardized in daily training and practice. Players will make small adjustments to their movement patterns according to the changes in the situation during the match, and finally complete the stroke task. Previous studies have reported the important role of pelvic axial rotation on racket acceleration [41, 77], even trunk rotation is probably the most critical factor in the development of racket speed [78], a higher velocity was observed in CC as compared with LL in tennis [78]. The CC has a longer trajectory than the LL [76], and the target area was on the left side of the playing body, these were the results obtained in players trying to get a racket acceleration during the forward swing phase through full muscle elongation and a greater axial rotation of the lower trunk in CC. The ROM and peak moment of PAR in CC were significantly higher than in LL in the current study, this also could be linked to a more weight transfer that could bring more energy transfer to further enhance the racket acceleration [54], because the playing arm was the endpoint of the body during stroke motor program which follows the proximal-to-distal segmental sequences in the kinetic chain [15, 32, 46, 54]. However, the result of the pelvis movement between CC and LL was different from the study of Malagoli Lanzoni et al. (2018) [76]. This is due to the different calculations, in their study the angle of axial pelvic movement was calculated relative to the table and not based on the player's own body, and the position of the player's feet when hitting the ball was not taken into account, the player's position was different in CC and LL, so the movement information of the pelvis is not comprehensive enough if only the playing table was used as a reference in evaluation. The moment of LL in the LF and LLB was significantly higher than CC in the early stroke phase. This result could support the hypothesis of Xing et al. (2022) [79] in the discussion section. Furthermore, this could probably be explained that LL has a shorter trajectory [76] and less forward swing time compared with CC [79], which results in the players having to pull their muscles as soon as possible in a limited time to gain more elastic energy to complete an attractive stroke. On the other hand, a shorter running trajectory of the ball in LL means a shorter reaction time for the player, which further requires the player to return to the ready position for the next stroke. This could explain why the ROM and maximum angle of LF and LLB in the LL were significantly less than in the CC.

After understanding the differences between the lumbar and pelvis movements of CC and LL topspin forehand, players could enhance the motor control of lumbar and pelvis movements according to the movement characteristics, either by enhancing core strength to improve the explosive power of lumbar and pelvis movements or by flexibility training to enhance lumbar and pelvis synergy, as these modalities are able to enhance the level of energy transfer in the power chain and improve performance. Beginners could quickly understand the role and contribution of the lumbar and pelvis in topspin forehand skills based on the results of this study, thus making it easier to master CC and LL topspin forehand skills. There are several limitations of this study that have to be mentioned [17]: the result of this study was limited to male table tennis players; therefore, the result may not be generalizable to female players [62]; the results of this study were generated in a laboratory environment and the results may be inaccurate in relation to a real game environment, for example, where the player needs to judge the rotation and direction of the next ball, which may result in the player having to adjust their body to ensure they can move to the correct position at all time [80]; the motion time of stroke in each phase and racket velocity should be measured in further studies.

4.2 Topspin Forehand Optimization

The research aimed to describe and compare the lower limb kinematic characteristics of the topspin forehand loop between EA and MA. The key findings of this study were: In time-spending, there were no significant differences between EA as well as MA during the entire playing phases. (1) EA showed significantly less knee and hip flexion in the BP compared with MA, and a significantly larger ankle varus and eversion than MA in the BP and FP, respectively. (2) EA displayed a significantly larger angular changing rate in ankle dorsiflexion as well as varus during the BP with ankle plantarflexion and eversion during the FP. Moreover, EA showed a significantly larger ankle internal and external rotation than MA in the BP and FP, respectively. (3) Compared with MA, EA showed a significantly larger rotation of external as well as internal in the knee in the BP and FP respectively. EA showed significantly larger ankle dorsiflexion and plantarflexion ROM in the BP and FP respectively compared with MA. Between EA and MA during the entire phase, have no significant differences in motion time for significant. However, significantly less time was shown by EA during the BP as well as a significantly larger time period during FP compared with MA. Bankosz and Winiarski (2017) reported that increasing the BP resulted in longer subsequent phases and elongation of the total time [81]. However, Qian et al. (2016) reported that compared with intermediate players, the superior players showed less time during FP [17].

EA showed a significantly larger ankle varus and internal rotation than MA in the BP. This could reinforce the stretching activity of internal rotation, and resulting in the contraction effects enhanced during the FP [58]. According to the theory of the stretch-shortening cycle, the performance of concentric contraction of muscle tendons would be enhanced by the elastic energy stored in the process of eccentric [24, 82]. This means that EA makes a greater preparation of the ankle during BP compared with MA. Compared with MA, EA showed significantly less knee and hip flexion in the BP, as well as a significantly larger ankle varus and internal rotation in the BP. This probably indicates the compensatory mechanism of the ankle joint in the BP. Excessive flexion of the hip and knee joint may result in the next lower limb stretching action's initial speed decrease. More sufficient torsion of the ankle joint can compensate for the flexion of the knee and hip joint, which helps improve the flexibility of the lower limbs. A way that a high racquet speed could generate an impact without undue injury risk was must coordinate all kinetic chain links [24, 83]

In the stage of FP, EA showed a significantly larger ankle external rotation and eversion than MA. Moreover, EA displayed a significantly larger external rotation and internal rotation of the knee in the BP and FP respectively compared with MA. This may result in a greater transfer range of weight to promote momentum generation [57]. Advanced players exhibited more whole-body movements than lower-skilled players by rotating the upper body through effective use of the knee joints in the previous table tennis studies on lower-limb biomechanics [18]. Myers et al. (2008) [84] reported that increased rotational counter-movement of the torso and pelvis at the top of the downswing in golf was associated with increased ball velocity. Compared with MA, EA showed a more sufficient ankle eversion rotation in the FP, and EA more sufficient ankle pedaling and stretching probably means a faster weight transfer effect. Bankosz & Winiarski (2018b) [46] have reported that involved segments of the lower limbs support proximal-to-distal movement sequencing: plantarflexion and rotation of the ankle joint, and rotation in the knee and hip joints. These movements result in the upward and forward velocities of the whole playing upper limb increasing at the contact moment during the follow-through movement. Moreover, Kasai & Mori (1998) [85] have evaluated the technique and performance of topspin shots. They drew attention to the differences between players, depending on the performance level. Players with a high-performance level had significant cooperation from the whole body, specifically the rotation of the trunk and the work of the knee joints. According to kinetic chain perspectives, the speed of the racket and ball in racket sport is considerably influenced by the energy transferred from the lower limb to the upper limbs [15, 86]. An important factor related to optimizing energy transfer in the kinetic chain is joint angular velocity which is expected to increase as skill levels improve [47]. Another key finding from this study is that EA displayed a significantly larger ankle dorsiflexion and varus changing rate during the BP with ankle plantarflexion and eversion during the FP. Qian et al. (2016) [17] have reported a similar result. This finding further reveals that the ankle joint as the starting point of movement plays an important role in the topspin forehand loop. The increased ankle angular changing rate of EA during FP in this study may be related to a more effective lower limb drive for the ball speed to increase [17]. The quality of whiplash-like action was assessed by the sequence as well as the interval of time in momentum transfer between the distal and proximal [87]. The higher angular changing rate of the ankle probably means that compared with MA, EA displayed a faster weight transfer and a shorter time of pedal and stretch in the lower limb. This proximal segment's power transference may play a crucial role in throwing speed, such as the ability that mechanical energy transmits from the trunk to the upper limb to produce a faster racket speed in the athletes [88]. Wang et al. (2018) [43] have reported a similar point, in their study, EA presented a larger angular changing rate of the lower limb joint and less time to hit the table tennis ball, which probably means a higher speed of play. Seeley et al. (2011) [47] revealed that the speed of post-impact increased from slow to medium levels resulting in the velocity of peak ankle plantarflexion and hip extension increased in racket sport forehand. However, during the forehand topspin, the larger knee and ankle loading magnitudes found in sidestep and cross-step footwork would predispose to overuse conditions of table tennis players such as jumper knee [15, 64] and ankle sprain injuries [15, 65]. This requires to development of the ability of rapid muscle response to stabilize dynamic joints during sports activities [89]. We can speculate that a focus on the rapid response ability of the muscles that surround the ankle joint will effectively enhance the ankle dynamic stability as well as decrease the injury likelihood in the ankle during the fast movement of table tennis topspin forehand loop play.

Personalized training is one of the principles of sports training, which aims to adapt training programs, training methods, and training loads to the individual needs of the athlete [17]. Athletic diversity is often attributed to factors such as body anatomy, level of motor skill development, gender, level of technical performance, age, and psychological quality [13]. The consequences of these factors are reflected in athletic performance, forming the movement characteristics of joints and muscles. The EA has a shorter stroke time [16, 17, 20], and they provide rapid hip flexion/extension angular velocity through the rapid work of the muscles around the hip joint [21, 14], which enhances the axial rotation of the lower trunk and increases the acceleration of the racket [41]. The shorter stroke time is beneficial to the athlete with sufficient time to prepare for the next stroke and execute the strategy [1, 20]. This may also explain why EA had significantly greater lower limb joint movement in both the sagittal and transverse planes. Previous studies have shown that when performing high-intensity forehand topspin, the lower body muscles of athletes are fully activated, and the muscle activity is significantly higher than that of other forms of hitting [21]. This demonstrates the involvement and contribution of lower body muscles in the forehand topspin stroke. Therefore, we strongly recommend building strength and explosiveness of the lower limb muscles, as excellent proficiency optimizes the transmission efficiency of the kinetic chain. In addition, long hours of practice in forehand topspin skills are also necessary. Based on the SSC (stretching-shortening cycle, SSC), the elastic energy stored in the muscle-tendon stretching phase can enhance the concentric movement of the muscle, and the training of SSC and strength should be combined to perform which could ensure that athletes are technically competent at each phase prior to progress in strength and complexity. Further to this, strong lower body strength can bring gains to the stability and motor control of footwork and provide support for the stability of the backward phase. He et al. [20] reported the important role of the ankle joint during the forehand topspin, strengthening the muscles around the ankle joint and the subtalar joint can help athletes maximize the important role of the foot as the origin of the kinetic chain.

This information will guide coaches and athletes to attach importance to the role of the ankle joint in the lower limb dynamic chain in the forehand topspin, and the training of the lower limb muscle rapid reaction ability, especially the ankle joint. Some limitations to this study should be mentioned. One of the major limitations of this study was that the athletes performed the action without a match environment. Besides, during the moment of racket-ball impact, there are no variables information in this study.

4.3 Balance Recovery

This research through the CI and CON two ways for professional table tennis players after lower limb exercise fatigue recovery intervention, by measuring the athletes' static balance and dynamic balance ability at six moments change, and exploring the cryotherapy for professional table tennis players, the influence of recovery after lower limb exercise fatigue, from the perspective of biomechanics, further reveals the human body after exercise fatigue recovery mechanism. This study aims to provide powerful theoretical guidance and support for coaches and athletes to choose more effective recovery methods after exercise fatigue. The innovation of this study is that 1) The cryotherapy instrument was used to perform cryotherapy at 0°C on the thighs and calves of subjects after the fatigue of lower limb muscles. 2) cryotherapy instruments can put the accurate temperature control at 0°C, make up for the research blank of this temperature, and provide the theoretical basis for the researchers. 3) Research on table tennis players' static balance ability and dynamic balance ability are few, and this study further enriches the research content in this field. The main results of this study are as follows: 1) Under the intervention of CI and CON, the COP area of athletes showed no difference at six moments. 2) At 48- and 72-hours post-intervention, YBT was significantly improved in both the CI and CON and the recovery effect of the CI was significantly better than CON. Moreover, 24 h post-intervention, the CI showed a significantly better dynamic balance recovery effect than the CON. 3) At the moment of post-intervention, the maximum displacement of COP in the AP and ML of CI was significantly greater than the post-warm-up, showing poor recovery. 4) At the moment of 72 h post-intervention, the maximum displacement of COP of the CI in ML was significantly less than CON, showing a good recovery. The main results of this study are discussed in detail below.

The results of this study showed that the maximum displacement of COP in AP and ML at the moment of post-intervention was significantly greater than the post-warm-up in CI, which indicated that the static balance ability of CI did not recover to the pre-exercise level at the moment of post-intervention. This result has been supported by previous studies. Kernozek et al. [90] investigated that the static COP wobble in the ML was significantly enhanced after cryotherapy was applied to a group of subjects with a lateral ankle sprain. Fukuchi et al. [91] reported that under bipedal standing conditions, cryotherapy increased COP standard deviation and velocity in the ML. The COP displacement velocity in AP and ML was higher after cryotherapy under the condition of one-legged standing. This means that cryotherapy would result in negative effects before more challenging postural control activities. Macedo et al. (2016) [92] explored the effect of cryotherapy on electromyographic response and balance of the lower limb during monopod jump landing, they investigated that cryotherapy increased the amplitude and average velocity of COP.

In many competitive sports competitions, athletes are usually treated with cryotherapy immediately after physical injury [93]. After cryotherapy for an acute knee injury, the athlete can return to training or competition [94]. There is some physiological and clinical evidence that cold compresses can effectively reduce nerve conduction velocity, muscle power, and muscle strength generation [95]. For every 1°C in skin temperature, nerve conduction velocity slows down by 1.5 to 2 meters per second [96], and for every 1°C decrease in muscle temperature, muscle spindle discharge rate decreases by 1–3 pulses per second [97]. The decrease of the static balance control ability at the moment of post-intervention is probably because cryotherapy results in the human body sensors of proprioception loss, which may lead to a change in posture stability [91, 98]. Because the nerve conduction velocity after cryotherapy may be damaged [99], the ability of muscles to control and adjust posture after the body balance is disrupted may also be affected. Cryotherapy has been shown to reduce incoming somatic sensory information from the knee joint. Hopper et al. (1997) [100] found that application of cryotherapy to the ankle resulted in a significant decrease in ankle proprioception, while application of cryotherapy to the knee resulted in less change in knee proprioception, but this subtle reduction in proprioception can lead to a decline in static and dynamic balance on the field. At the moment of 72 h post-intervention, the maximum displacement of COP in the CI on the ML was significantly less than CON, showing a good recovery. Due to the fatigue of lower limb muscles, athletes will suffer from joint relaxation and proprioception decline, which will lead to the decline of joint stability. However, rapid cryotherapy after fatigue can reduce body energy consumption, improve joint stiffness, and activate the central regulation mechanism [101]. Furthermore, at the moment of post 72 h of intervention, the recovery of dynamic balance ability was improved with the elimination of fatigue.

Table tennis is a competitive sport played on a small field, which requires players to run continuously in a small range while playing. At the same time, players need to complete a series of instantaneous explosive movements and change direction quickly and frequently in the process of continuous movement to achieve the purpose of effective hitting [21, 40, 102]. Table tennis is characterized by fast speed, varied rotation, and the small size of the ball [16], which is a great test for players' rapid reaction ability, stride speed, strength, and endurance quality. The center of gravity of mastering transformation is the key point of footwork skill in table tennis, footwork movement balance to keep the body in the trunk and reasonable position, to ensure the stability of the barycenter, to avoid large fluctuations of the center of gravity in the up and down direction, the focus of the substantial guarantee for athletes in a fast-moving high-quality shot provides stable body support. In addition, the balance of the torso provides a guarantee for the athlete to start and brake quickly. Therefore, good posture adjustment ability is not only conducive to reducing the occurrence of sports injuries but also conducive to improving the quality of technical movements of table tennis players. In this study, the YBT performance of the participants at different moments was used to evaluate the effect of cryotherapy recovery on table tennis players' dynamic balance ability. At the moment of 24 h, 48 h, and 72 h post-intervention, the CI has a significant positive effect on the dynamic balance ability of players compared with CON. The YBT performance of players at 72 h post-intervention was significantly better than at 48 h post-intervention. This indicated that cryotherapy began to positively promote the dynamic balance ability of athletes at 24 h post fatigue of lower lime muscle, and the promoting effect lasted until 72 h post-intervention. However, the YBT performance of players at the moment of post-intervention was significantly lower than the post-fatigue in CI. These results indicated that the dynamic balance ability of athletes decreased further after cryotherapy, which is consistent with some previous studies. Montgomery et al. (2015) [103] investigated that 10 minutes CI below the hip joint at 12°C significantly reduced the dynamic balance ability of participants. The study of Kernozek et al. (2008) [90] showed that after cryotherapy on participants with lateral ankle sprains, mediolateral swing variability increased. In the YBT, the farther the subjects touched, the greater their neuromuscular strength, proprioceptive control, and range of joint motion [104]. Any disturbance to the body of these factors can impair balance, and cold stimulation as a disturbance will result in reduced blood flow to the extremities. This redistribution of blood flow may damage neuromuscular and somatosensory components that are important for performing dynamic sensory tasks such as balance and strength [105, 106, 107].

5 Limitations

There are several limitations of this study that have to be mentioned. Firstly, the result of this study was limited to male table tennis players; therefore, the result may not be generalizable to female players. Secondly, the results of this study were generated in a laboratory environment and the results may be inaccurate in relation to a real game environment. Thirdly, the motion time of stroke in each phase and racket velocity should be measured in further studies.

6 Conclusion

Research on the biomechanics of the lower limbs in table tennis has received extensive attention and reports. The key information and findings of the research have been extracted, categorized, combined with practice, and applied for training and participation in table tennis competitions. Based on the results of this study, we have summarized the following key information and recommendations for preventing injury and motor control.

First, the JRF and joint stiffness of the subtalar are large, which indicates that the subtalar joint bears a large impact force during landing, which may cause injury to the foot. The medial and lateral of the rearfoot showed high plantar pressure during the landing stage, and the big toe area and medial forefoot showed high plantar pressure during the forward phase. Strengthening the posterior muscles of the lower limbs can help improve stability during the landing phase.

Second, cryotherapy was not recommended for balance recovery if the competition was on the same day or within 24 hours but it was recommended if the competition was on the next day or after the next day.

Third, compared with the long-line topspin forehand, the cross-court topspin forehand shows a significant violent movement on lumbar left bending and flexion, this maybe could provide information to investigate lower back pain.

The present study could probably provide some support for clinical application. The table tennis coach and the professorial athlete could acquire valuable information to optimize training strategy and enhance motor control during the topspin forehand. Relevant researchers could quickly establish a basic understanding and knowledge base on the lower limb biomechanics of table tennis topspin forehand through this study.

Thesis Points

1st Thesis point: *I provided the first hybrid model (Gait2392 musculoskeletal model in OpenSim together with a Novel Pedar insole plantar pressure measurement system) for detecting lower extremities injury areas.*

First, using experiments I could prove that the highest peak pressures appear in the medial-lateral rear foot and the lateral forefoot during the backward phase if one-step movement is considered [1]. *Concerning the chasse-step, the most dangerous area for possible injury was found to be the toe during the forward phase (as shown in Figure 33).*

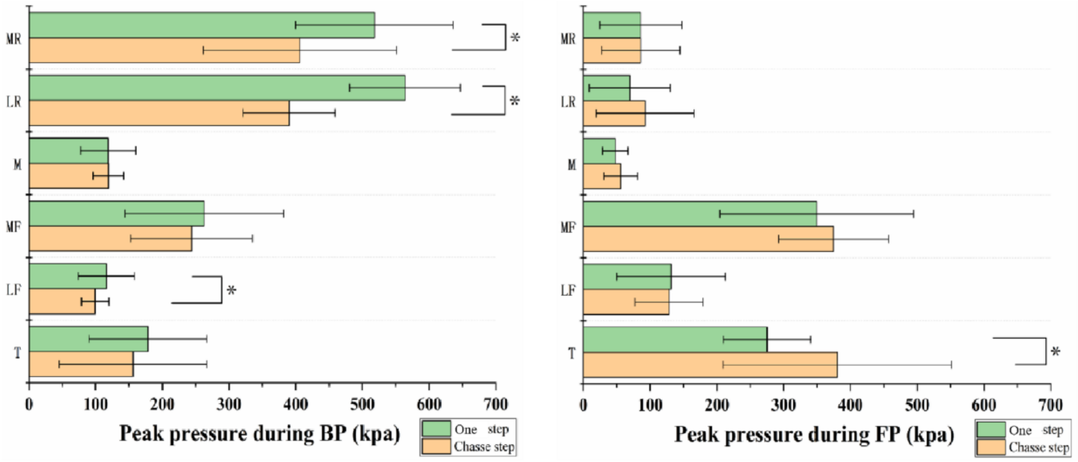

Figure 33. Comparison of peak pressure of each plantar region during BP and FP. Note: The asterisk (*) refers to significance with $p < 0.05$.

Second, the activation and the behavior of eight major muscles (as shown in Figure 34) were characterized in the OpenSim model as a function of the stroke phase and validated by electromyography measurements [2]. *Based on these force characteristics one can deduce that at what percentage of the stroke may the major muscles reach their maximum activity, which can be plausible for injury. These force responses provide valuable information to sports coaches when strategies for muscle strengthening in table tennis are considered.*

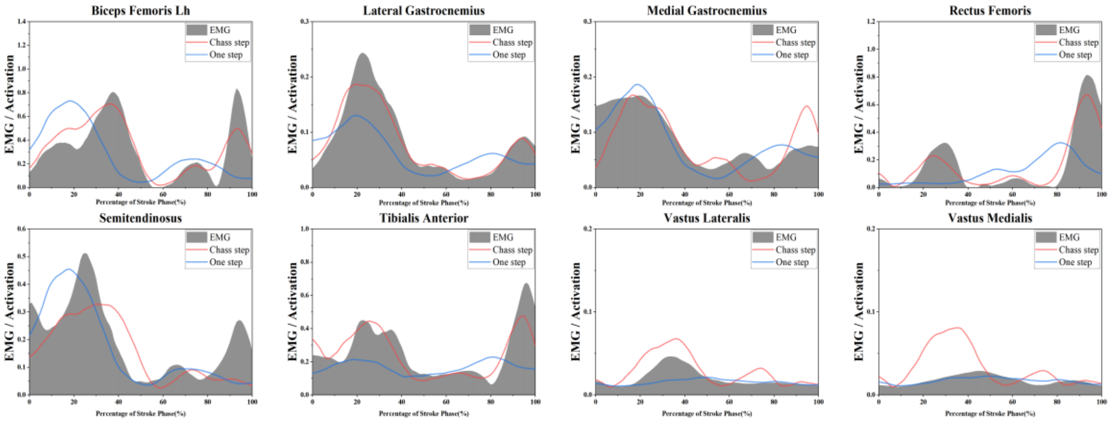

Figure 34. Comparison of lower extrimities muscle sEMG signals and activations from OpenSim Optimization between the chasse step and one-step during stroke in table tennis.

Third, the joint stiffness of the lower limb during the landing stage was simulated and calculated [3] *(as shown in Figure 35), which could investigate the intrinsic mechanical mechanism during the landing process, can report the possible areas where injury may occur.*

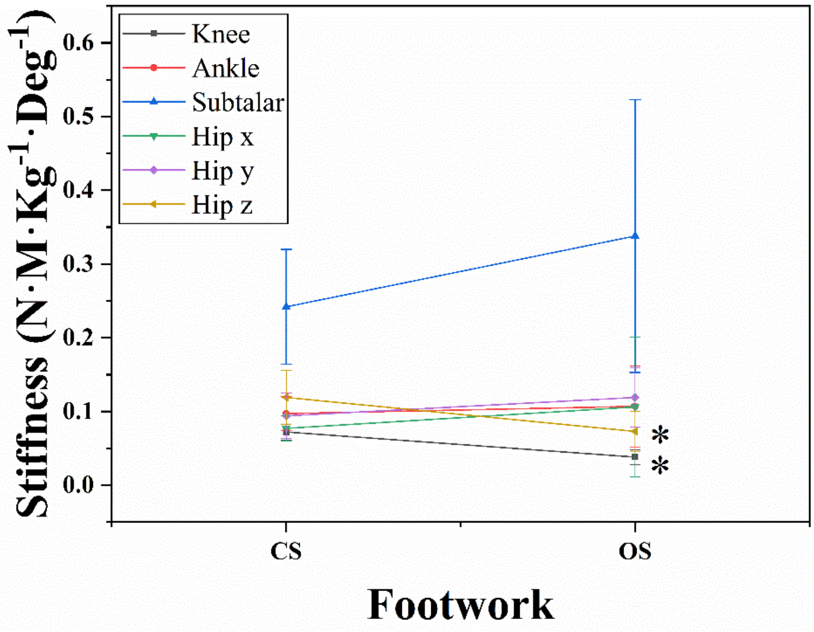

Figure 35. Lower limb joint stiffness during the landing stage in the chasse step and one-step footwork.

Related articles to the 1st thesis point:

[1] **He, Y.,** Sun, D., Yang, X., Fekete, G., Bak er, J. S., & Gu, Y. (2021). Lower limb kinetic comparisons between the chasse step and one step footwork during stroke play in table tennis. PeerJ, 9, e12481. (**Q1,** IF: 3.061)

[2] **He, Y.,** Shao, S., Fekete, G., Yang, X., Cen, X., Song, Y., Sun, D. & Gu, Y. (2023) Lower Limb Muscle Forces in Table Tennis Footwork during Topspin Forehand Stroke Based on the OpenSim Musculoskeletal Model: A Pilot Study. Molecular & Cellular Biomechanics, 19(4), 221–235. (EI, Scopus)

[3] **Yuqi He,** Penghui Zhang, Zixiang Gao, Xiaoyi Yang, Gusztáv Fekete, András Kovács, and Yaodong Gu. Joint reaction force and stiffness during the landing stage in table tennis footwork. The 29th European Society of Biomechanics. 2024. Edinburgh, Scotland

2nd Thesis point: *I have developed and produced portable cryotherapy equipment for the recovery of lower limb fatigue in professional table tennis players*[1,2]*. As shown in Figure 36, I calculated and confirmed that 1) from the 24h post-intervention, the effect of cryotherapy on dynamic balance recovery was significantly better than no cryotherapy; 2) Except for the COP maximum displacement on ML at the 72h post-intervention, the cryotherapy had no positive effect on the recovery of static balance ability; 3) Cryotherapy has a significant negative impact on the COP maximum displacement in ML and AP at the post-cryotherapy, which may lead to the decline of static balance ability. Therefore, it was not recommended to use cryotherapy for balance recovery if the competition was on the same day or within 24 hours. Cryotherapy was recommended if the competition was on the next day or after the next day.*

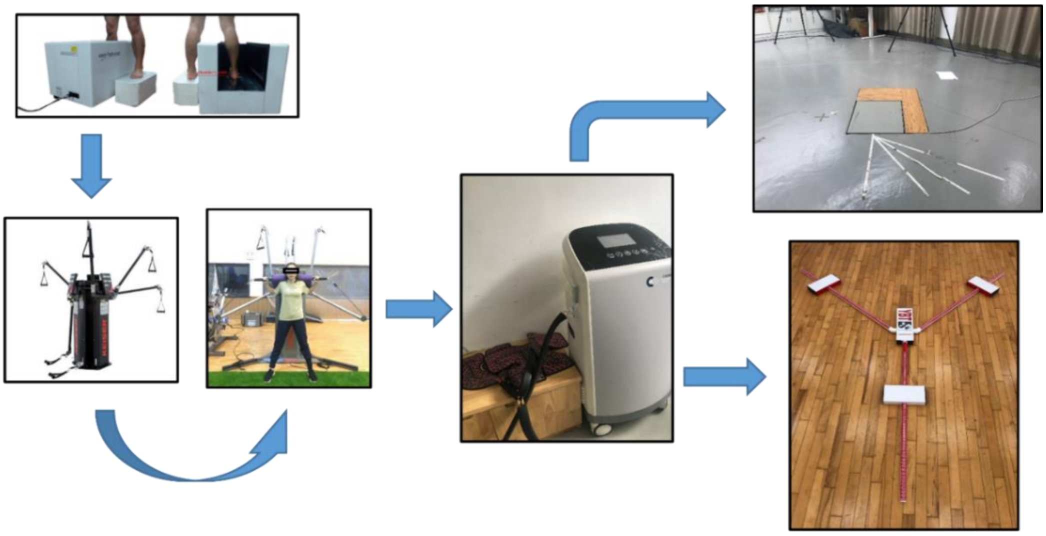

Figure 36. The biomechanics exploration flowchart of the cryotherapy effect on balance ability after lower extremity fatigue.

Related articles to the 2nd thesis point:

[1] **He, Y.,** & Fekete, G. (2021). The Effect of Cryotherapy on Balance Recovery at Different Moments after Lower Extremity Muscle Fatigue. Physical Activity and Health, 5(1). (Scopus)

[2] Lu, Y., **He, Y.,** Ying, S., Wang, Q., & Li, J. (2021). Effect of Cryotherapy Temperature on the Extension Performance of Healthy Adults' Legs. Biology, 10(7), 591. (**Q1,** IF: 5.168)

3rd Thesis point: *As a first researcher in the field, I have given a complete kinematic and kinetic description of the lumbar movement concerning cross-court and long-line topspin forehand using computational simulation and experiments.*

In the kinematic data, I explored for the first time how the lumbar axial rotation, left lateral bending, and flexion movement behaviors as a function of the stroke play phase. It is visible from the results that the cross-court topspin forehand has slightly higher (cca.15%) values throughout the motion (as shown in Figure 37).

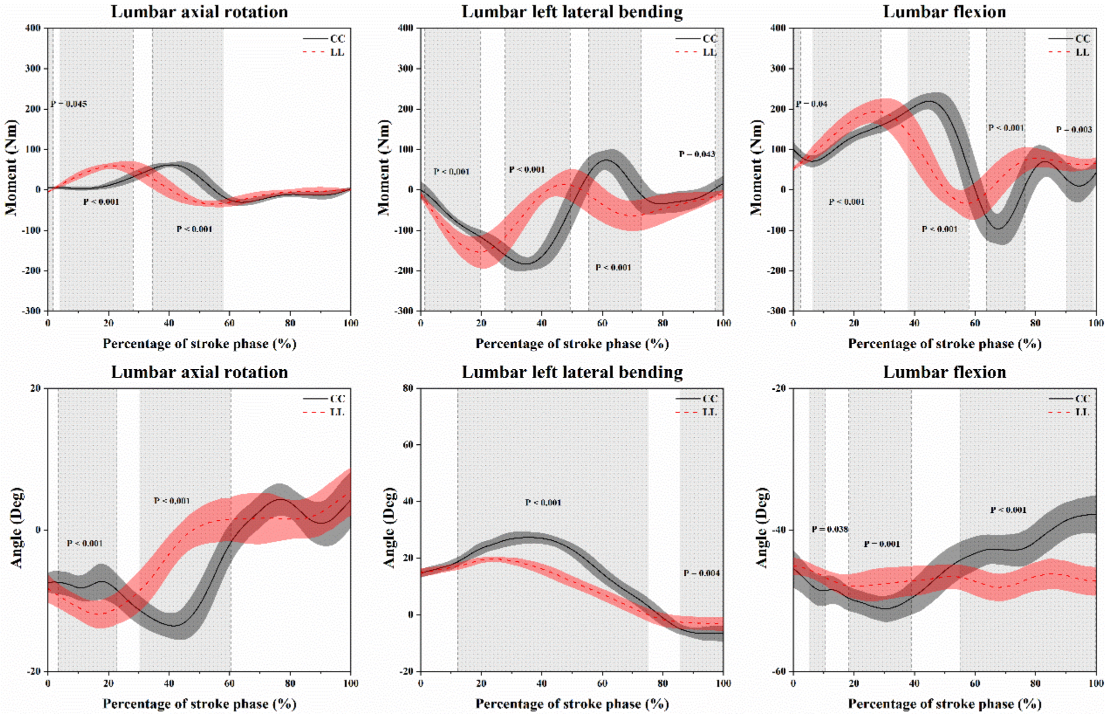

Figure 37. The simulation results of the lumbar angle and moment during cross-court and long-line topspin forehand.

Besides the kinematic data, important findings can be deducted from the kinetic results as well. As shown in Figure 38, the cross-court topspin forehand has a significant effect during lumber flexion (cca.14% higher moment than during long-line topspin forehand) and lumbar left lateral bending (cca.16% higher moment than during long-line topspin forehand). It can be assumed that this increased, and suddenly appearing load on the spine can be the root of lower back pain in the long term.

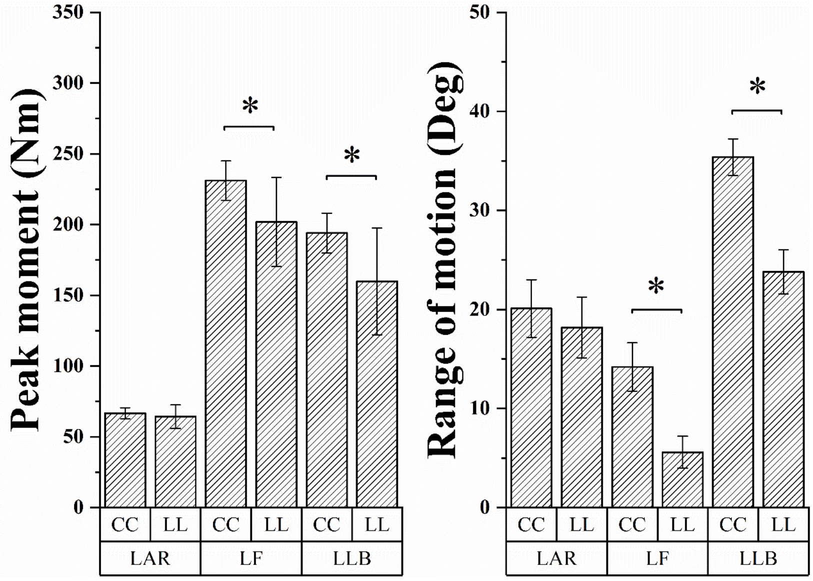

Figure 38. The Rom and peak moment comparison of lumbar movement between the CC and LL topspin forehand.

Note: “*” indicates a significant difference between LL and CC.

Related articles to the 3rd thesis point:

He, Y., Liang, M., Fang. Y., Fekete, G., Baker, J. S., & Gu, Y. (2023) Lumbar and pelvis movement comparison between cross-court and long-line topspin forehand in table tennis: based on musculoskeletal model. Frontiers in Bioengineering and Biotechnology, 11, 1185177. (**Q1,** IF: 5.7)

List of Publications

Referred Articles Related to This Thesis:

He, Y., Liang, M., Fang. Y., Fekete, G., Baker, J. S., & Gu, Y. (2023) Lumbar and pelvis movement comparison between cross-court and long-line topspin forehand in table tennis: based on musculoskeletal model. Frontiers in Bioengineering and Biotechnology, 11, 1185177. (**Q1,** IF: 5.7)

He, Y., Fekete, G., Sun, D., Baker, J. S., Shao, S., & Gu, Y. (2022). Lower Limb Biomechanics during the Topspin Forehand in Table Tennis: A Systemic Review. Bioengineering, 9(8), 336. (**Q2,** IF: 5.046)

He, Y., Shao, S., Fekete, G., Yang, X., Cen, X., Song, Y., Sun, D. & Gu, Y. (2023) Lower Limb Muscle Forces in Table Tennis Footwork during Topspin Forehand Stroke Based on the OpenSim Musculoskeletal Model: A Pilot Study. Molecular & Cellular Biomechanics, 19(4), 221–235. (EI, Scopus)

He, Y., Lyu, X., Sun, D., Baker, J. S., & Gu, Y. (2021). The kinematic analysis of the lower limb during topspin forehand loop between different level table tennis athletes. PeerJ, 9, e10841. (**Q1,** IF: 3.061)

He, Y., Sun, D., Yang, X., Fekete, G., Bak er, J. S., & Gu, Y. (2021). Lower limb kinetic comparisons between the chasse step and one step footwork during stroke play in table tennis. PeerJ, 9, e12481. (**Q1,** IF: 3.061)

He, Y., & Fekete, G. (2021). The Effect of Cryotherapy on Balance Recovery at Different Moments after Lower Extremity Muscle Fatigue. Physical Activity and Health, 5(1). (Scopus)

Lu, Y., **He, Y.,** Ying, S., Wang, Q., & Li, J. (2021). Effect of Cryotherapy Temperature on the Extension Performance of Healthy Adults' Legs. Biology, 10(7), 591. (**Q1,** IF: 5.168)

Yang, X., Mei, Q., Shao, S., Gu, W., **He, Y.,** Zhu, R., & Gu, Y. (2022). Understanding Sex-Based Kinematic and Kinetic Differences of Chasse-Step in Elite Table Tennis Athletes. Bioengineering, 9(6), 246. (**Q2,** IF: 5.046)

International Conference Abstracts Related to This Thesis:

Yuqi He, Zixiang Gao, Gusztáv Fekete, András Kovács, Dusan Mitic, and Yaodong Gu. Lower lime muscle forces in table tennis footwork during topspin forehand based on musculoskeletal. The 28th European Society of Biomechanics. 2023. Maastricht, Netherlands

Yuqi He, Zixiang Gao, Gusztáv Fekete, András Kovács, Aleksandar Nedeljkovic, Dusan Mitic, and Yaodong Gu. Lumbar and Pelvis Movement Comparison between Cross-court and Long-line

Topspin Forehand Stroke: Based on Musculoskeletal Model. The 50th International Society of Biomechanics. 2023. Fukuoka, Japan

He Yuqi, Gao Zixiang, Fekete Gusztáv, Mitic Dusan, and Gu Yaodong. Plantar force comparisons between the chasse step and one step footwork during topspin forehand using statistical parametric mapping. The 40th International society of Biomechanics in Sports Proceedings Archive. 2022. Liverpool, England

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Yuqi He, Penghui Zhang, Zixiang Gao, Xiaoyi Yang, Gusztáv Fekete, András Kovács, and Yaodong Gu. Joint reaction force and stiffness during the landing stage in table tennis footwork. The 29th European Society of Biomechanics. 2024. Edinburgh, Scotland

Other Publications:

He, Y., Lv, X., Zhou, Z., Sun, D., Baker, J. S., & Gu, Y. (2020). Comparing the kinematic characteristics of the lower limbs in table tennis: Differences between diagonal and straight shots using the forehand loop. Journal of sports science & medicine, 19(3), 522. (**Q1,** IF: 4.017)

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Gao Zixiang, **He Yuqi,** Xiang Liangliang, Fekete Gusztáv, Kovács András, and Gu Yaodong. Automatically detecting fatigue gait based on time series bilateral plantar force distribution using deep learning algorithms. The 28th European Society of Biomechanics. 2023. Maastricht, Netherlands

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Reviewer for International Journal Articles:

1. PloS One
2. Computer Methods in Biomechanics and Biomedical Engineering
3. International Journal of Biomedical Engineering and Technology
4. BMC Sports Science, Medicine and Rehabilitation
5. Frontiers in Bioengineering and Biotechnology
6. Sport Sciences for Health
7. Physical Activity and Health
8. PeerJ
9. Frontiers in Physiology
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11. Applied Bionics and Biomechanics
12. Frontiers in Sports and Active Living

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Hirsch Index: 6 (Scopus)
Total independent citations (Scopus): 71

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